Investigating Augmented Reality Visio-Haptic Techniques for Medical Training

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ABSTRACT

It is widely accepted that a reform in medical teaching must be made to meet today's high volume training requirements. Receiving pre-training in a core set of surgical skills and procedures before novice practitioners are exposed to the traditional apprenticeship training model where an experienced practitioner must always be present, can reduce both skill acquisition time and the risks patients are exposed to due to surgeon inexperience. Virtual simulation offers a potential method of providing this training and a subset of current medical training simulations integrate haptics and visual feedback to enhance procedural learning.

The role of virtual medical training applications, in particular where haptics (force and tactile feedback) can be used to assist a trainee to learn and practice a task, is investigated in this thesis. A review of the current state-of-the-art summarises considerations that must be made during the deployment of haptics and visual technologies in medical training, including an assessment of the available force/torque, tactile and visual hardware solutions in addition to the haptics related software. An in-depth analysis of medical training simulations that include haptic feedback is then provided after which the future directions and current technological limitations in the field are discussed.

The potential benefits of developing and using a new Augmented Reality (AR) visio-haptic medical training environment is subsequently explored, and an exemplar application called PalpSim has been produced to train femoral palpation and needle insertion, the opening steps of many Interventional Radiology (IR) procedures. This has been performed in collaboration with IR experts. PalpSim's AR environment permits a trainee to realistically interact with a computer generated patient using their own hands as if the patient existed in the real world. During a simulation, the trainee can feel haptic feedback developed from *in vivo* measured force data whilst palpating deformable tissue and inserting a virtual needle shaft into a simulated femoral artery, at which point virtual blood flow from the real needle hub will be seen. The PalpSim environment has undergone face and content validation at the Royal Liverpool University Hospital and received positive feedback.

An important requirement identified was for a haptics device combining force and tactile feedback to closely simulate the haptic cues felt during femoral palpation. Two cost effective force feedback devices have therefore been modified to provide the degrees of force feedback needed to closely recreate the forces of a palpation procedure and are combined with a custom built hydraulic tactile interface to provide pulse-like tactile cues. A needle interface based on a modified PHANTOM Omni also allows the user to grasp and see a real interventional radiology needle hub whilst feeling simulated insertion forces.

PalpSim is the first example of a visio-haptic medical training environment based on chroma-key augmented reality technology. It is expected that many other medical training solutions will adopt this approach in the future.

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CONTENTS

Statement of Originality

The work presented in this thesis was carried out by the candidate, except where otherwise stated. It has not been presented previously for any degree, nor is it at present under consideration by any other degree awarding body.

Candidate:

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Statement of Availability

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1 Introduction

1.1 Context

Currently, there is unrelenting pressure to update and reform conventional medical practices. Patient safety in particular has been highlighted as a key issue to be addressed by medical processes and technology [1]. These concerns are driving surgical management into innovative minimal access approaches, which in turn are raising further challenges of training the increasingly complex skills required. Safe practice requires the operator to respond correctly to both visual and haptic cues. The operator's deliberations then initiate and inform a range of motor actions, including very fine translational and rotational motions of tools, particularly in challenging anatomy. As the spectrum of available techniques increases, the limited number and availability of suitably trained practitioners becomes a significant problem. Further exasperating this problem, cost minimisation is high on hospital agendas, yet training under the apprenticeship model is expensive as it increases procedure duration [2]. Work time directives in the US and EU are also greatly reducing trainee practitioners' hours of work, further limiting the available training time and thus decreasing the procedural experience a newly qualified surgeon will have as they perform their first unsupervised interventions.

Virtual simulations offer a potential method of providing the pre-training of practitioners, which can alleviate the aforementioned training issues and if correctly designed, can produce reconfigurable simulations that reproduce the look and feel of *in vivo* procedures in a variety of patients. This thesis presents a novel medical simulation solution that integrates haptics technology with augmented reality (AR) in an immersive environment that has broad applicability for many medical training simulation applications. Work performed in collaboration with interventional radiology (IR) specialists has led to an exemplar simulation that addresses femoral palpation and needle insertion, the opening steps of many interventional radiology procedures. In this procedure, described in Chapter 3, a practitioner uses the fine haptic cues felt at their fingertips as they press upon a patient's skin to guide a needle into the femoral artery.

1.2 Hypothesis

Femoral artery palpation and needle insertion can be virtually simulated, effectively substituting existing mannequin-based training methods. Off the shelf visualisation and haptics technologies can be modified to produce a low-cost visio-haptic simulation platform that provides a high fidelity femoral pulse palpation and needle insertion simulation, whilst overcoming the patient variability and simulation durability problems inherent in mannequin-based simulation approaches.

To defend this statement, the thesis will attempt to answer four key research questions:

1. What are the problems with existing virtual simulation technology?
2. Does haptics technology that can be used for effective virtual simulation of a femoral palpation and needle insertion currently exist? If not, can the technology be developed?
3. Is there an ideal visualisation method for a medical visio-haptic training simulation?
4. Can a virtual femoral palpation and needle puncture simulation offer increased functionality over traditional training techniques?

1.3 Thesis Structure and Contributions

This thesis identifies several problems restricting the development of full procedure medical training simulation solutions. An exemplar femoral palpation and needle insertion simulation called PalpSim addresses a subset of the technical challenges identified to advance the integration of visio-haptic training simulations into clinical training programs.

Chapter 2 introduces the prior art of virtual medical training simulations making use of haptics feedback for advanced procedural training. This review first highlights the need for medical training simulations and then introduces haptics and visual feedback. Commercial force and tactile feedback devices and the visualisation hardware available for use in simulation are highlighted in these sections. Medical training simulations making use of haptic feedback to simulate a variety of medical disciplines are surveyed so that conclusions on the emerging trends of the use of haptics and visual feedback in training can be made. A subset of the identified problems is addressed in the following chapters.

In Chapter 3, the need for training through simulation as part of the interventional radiology (IR) training curricula is addressed. The problems limiting current medical simulation are related to femoral artery palpation and needle insertion, a missing functionality in available virtual IR training simulators. Solutions to these problems are presented in the following chapters.

Chapter 4 addresses the visualisation problems presented by current medical simulation approaches. As identified in Chapter 3, current visualisation methods do not permit a trainee to reach down and touch a virtual patient, whilst seeing and feeling the patient below their fingertips, without an undesirable visual occlusion of the user's hands. The constituent parts of an augmented reality (AR) visualisation approach for medical simulation are developed here to overcome the identified occlusion problem. These parts are combined with haptic feedback in Chapter 7 to produce a full visio-haptic training workbench.

An identified lack of affordable, high fidelity tactile devices that can be used in conjunction with force feedback hardware to produce a realistic palpation simulation is addressed in Chapter 5. Multiple tactile solutions, piezoelectric pads, micro speakers and a pin array device have been evaluated for this purpose, before a fourth hydraulically actuated tactile solution has been chosen as the optimal technology, addressing the high fidelity requirements at relatively low cost. This solution, which closely reproduces the fine tactile cues of a femoral palpation, has been designed for integration with a force feedback device to reproduce the haptic resistance of the patient's tissue.

The modification of commercial force feedback devices to produce high fidelity hardware for both palpation and needle insertion is described in Chapter 6. Physical modification of commercial hardware allows fast simulation development at a comparatively low cost when compared to proprietary device development. Should the simulation be commercially produced, a faster deployment and testing cycle may also be achieved. Section 6.2 describes the production of palpation force feedback hardware, making use of two Falcons' from Novint (Albuquerque, USA), low cost commercially available devices. These two 3 force DOF devices are combined with two sets of dual revolute joints and a rigid link to produce a single, high powered 5 force DOF device capable of closely reproducing the forces felt during an *in vivo* femoral palpation. Section 6.3 describes the modification of SensAble's (Wilmington, USA) Omni force feedback hardware. This modification replaces the passive 3 rotational DOF stylus end effector in favour of a real interventional radiology needle hub. A 6 DOF needle hub provides the correct tactile cues as the needle is grasped and 3 force DOF can be asserted on the hub whilst the correct visual cues can be seen through the immersive AR display.

The development of the visio-haptic collocated training environment, PalpSim, including a description of the collocation between the real world visualisation, the virtual world and the force feedback devices, is outlined in Chapter 7. The position and interaction between the simulation objects is then described, along with the structure of the program's multithreaded communication. The calibration of the force and visual feedback is then explained.

Chapter 8 addresses the validation of the PalpSim environment. The results of a face and content validation study conducted in the Royal Liverpool University Hospital's radiology department are described. An evaluation of this study is made, and a further look at the tactile forces felt during palpation is provided. Future validation steps to be conducted are then outlined.

The main contributions are drawn together in Chapter 9, which consolidates the main challenges that exist in medical training simulations. The thesis contributions are described with regard to the four key research questions posed to address the thesis hypothesis. To conclude, future work is then summarised.

1.4 List of Publications

Part of the work presented in this thesis has been published in peer reviewed literature.

Journal Publications

- **T.R. Coles**, D.A. Gould, N.W. John and D.G. Caldwell, "Integrating Haptics with Augmented Reality in a Femoral Palpation and Needle Insertion Training Simulation", *IEEE Transactions on Haptics,* To appear.
- **T.R. Coles**, D Meglan and N.W. John, "The Role of Haptics in Medical Training Simulators: A Survey of the State-of-the-art", *IEEE Transactions on Haptics,* vol. 4, no. 1, Jan – Mar 2011, pp 51-66.

Peer Reviewed Publications

- **T.R. Coles,** D.A. Gould, N.W. John and D.G. Caldwell**,** "Virtual Femoral Palpation Simulation for Interventional Radiology Training" *EG UK Theory and Practice of Computer Graphics, 2010,* Pp. 123-126.
- **T.R. Coles**, N.W. John and D.G. Caldwell, "The Case for Augmented Reality when Training in a Virtual Medical Environment" *Workshop of Medical Virtual Environments at IEEEVR2010.*
- **T.R. Coles**, N. W. John, "The Effectiveness of Commercial Haptic Devices for Use in Virtual Needle Insertion Training Simulations," *Advances in Computer-Human Interaction, 2010, Third International Conference on*, 2010, pp. 148-153.
- **T.R. Coles**, N.W. John, D.A. Gould and D.G. Caldwell, "Haptic Palpation for the Femoral Pulse in Virtual Interventional Radiology" *Advances in Computer-Human Interaction, 2009, Second International Conference on*, 2009, pp. 193-198.

Poster Publications

- **T.R. Coles**, G. Sofia, N.W. John, D.A. Gould and D.G Caldwell**,** "Modification of Commercial Force Feedback Hardware for Needle Insertion Simulation" *Stud Health Technol Inform, 2011,* pp 135-137. Awarded a best poster prize.

2 Current Issues in Medical Simulation

2.1 Chapter Overview

This chapter firstly identifies the need for haptics enabled medical training simulations, as well as highlighting the necessary hardware and software considerations which must be made to produce such a product. A state of the art review of the use of haptics for medical training is then provided. A look at augmented reality simulation emphasises a potential application for medical training. This visualisation technique is developed for use in a femoral palpation and needle insertion simulation described in later chapters. Guidelines for choosing the optimum simulation medium, either mannequin based simulation, virtual based simulation or a hybrid based design are defined, before methods of simulation validation and practitioner skill evaluation are outlined. This chapter concludes with a summary of the reviewed technology and approaches and highlights the trade-offs and choices that must be made whilst developing a medical visio-haptic training simulation.

2.2 The necessity for virtual medical training

Training based on an apprenticeship model has been used effectively by the medical profession for centuries. Here, a "see one, do one, teach one" approach to learning is used where a trainee observes a procedure, practises it under supervision and, when proficient, becomes a mentor themselves. This is also one of the main methods currently used in Interventional Radiology (IR) training, a procedure focused upon throughout this thesis. This type of learning involves the experience of errors, albeit under the guidance of an expert mentor. Yet performing an operation incorrectly through inexperience can lead to avoidable patient discomfort and complications. The latter can prolong a patient's hospital stay or in the worst case scenario, can cause permanent damage or death.

A three year study by HealthGrades (Golden, CO, USA), an American healthcare ratings organisation, found that medical errors resulted in over 230,000 deaths in American hospitals [2]. In a different study based on rates of cancer recurrence in 4700 patients operated upon using keyhole techniques by 29 surgeons in 7 hospitals throughout Europe and N America, Vickers *et al.*[3] report that surgeons require 750 operations to perfect keyhole surgery procedures. It is not acceptable to make mistakes on patients when alternative training methods are available.

As technology has progressed, many different tools and techniques have been deployed to provide added value to the training process, such as using anesthetised animals or cadavers, or by practising on mannequins or fellow students. However, the interactions that occur in an animal's or cadaver's tissues differ from those of living humans due to varying anatomy or absence of physiological behaviour such as blood pressure. Not only are cadavers expensive, but procedures can only be performed once, and a mistake can render the body useless to re-demonstrate a procedure. This type of training also raises ethical issues. Mannequins of varying sophistication are becoming increasingly common to simulate part or all of a patient [4]. However, drawbacks of mannequins include limitations in their replication of physiology and that, at best, they have a limited range of anatomical variability. Barker [5] notes how students resort to training venipuncture upon fellow students due to the plastic mannequin models not providing enough realism. In addition, the aforementioned training methods usually require an expert trainer to be present to instruct trainees on best practice and operation, further increasing costs.

An alternative approach that is making an impact on the medical community is computer simulation enabled experiential training systems, which can train practitioners on a virtual patient, whilst critically analysing skills and providing feedback on the performed procedure without the presence of an expert trainer. This feedback can then be used to refine the required skills until the operator reaches a target level of proficiency before commencing training via the traditional apprentice model upon patients. Simulations can also provide the user with an opportunity to practice difficult cases and gain exposure to rare, but critical procedures that may not normally appear during a resident's training. As the field of simulation matures and becomes sufficiently accurate, simulation could also provide the user with an opportunity to practice difficult cases or to be exposed to those in which patient anatomy is unconventional, before performing the procedure upon a patient. Such "mission rehearsal" can highlight operational and equipment difficulties that would otherwise be overlooked until they are encountered during the real procedure.

Physical models require remodelling to simulate patient variability where a patient's body habitus (related to their quantity of muscle and fat) varies between different subjects. Virtual models offer the opportunity to simply modify the virtual patient using patient specific data from one of the many 3D medical imaging modalities available in the hospital, or to utilise the skills of the numerous well trained medical illustrators who are capable with 3D modelling packages. This is a significant advantage of computer simulation over that of a cadaver or fixed anatomical models. However, when producing a realistic training simulation, the virtual patient must be displayed to the practitioner in such a way that they believe the simulation replicates a real situation so as to achieve "suspension of disbelief" [6]. Cadavers and mannequins have physical presence which a simulation lacks. Overcoming this lack of presence is a challenge addressed in the following chapters of this thesis.

Of the human sensory modalities (visual, auditory, touch, smell and taste), the two cues most frequently used in virtual simulation and those addressed within this thesis are vision and touch. It is thought that smell and taste will be included in the future to heighten the suspension of disbelief a trainee can experience. An example of this could be to introduce the smell of a theatre using the ScentPalette from EnviroScent (Ball Ground, GA, USA). Sound is also a very important cue for the correct learning of certain procedures using high speed power tools such as burr-based bone and tooth drilling. The following review highlights a small number of simulations using auditory cues to alert a user to a fault, or to give guidance to the trainee. The touch and visual cues are of most interest in the following review and thesis. An introduction to the simulation of touch is now given.

2.2.1 Haptics

The term haptic is used to describe the sense of touch. The bidirectional sense of touch is based around two types of interaction, tactile and force/torque otherwise described as the cutaneous and kinaesthetic senses. Tactile cues are felt from receptors within or close to the skin, allowing humans to detect if a surface is smooth or rough, hot or cold, as well as conveying pain and information about surface vibrations. The kinaesthetic sense, described from here on as force/torque feedback, is felt from receptors within muscles, tendons and joints and can provide information about the weight and inertia of an object a person is holding, and the forces/torques exerted on the body through user-object surface contact. These receptors also allow a person to know where their hand is in space, even with closed eyes (proprioception).

Haptics solutions are less mature than visual display technologies. The exact function and thresholds of the various haptic receptors within the body are little understood when compared to the human visual system and, as such, this is an active field of research. A comprehensive reference of the perceptual thresholds of the hand has been written by Jones and Lederman [7]. Both tactile and force/torque feedback can be crucial to the success of carrying out a medical procedure.

The term 'force feedback' is often used in place of 'haptic feedback'. However, these terms are not interchangeable as force is only a small part of bidirectional real world interaction. In a general case of proprioceptive feedback, where a person interacts with a simulated scene, both force as well as torque must be experienced. This requires 6 force/torque DOF feedback, but this is not typically provided in simulation due to the higher cost of manufacturing devices that can provide torque as well as directional force feedback. During this thesis, it will be made explicitly clear if torques are included when referring to feedback and devices.

In 1965, Ivan Sutherland correctly predicted that the sense of touch would be added to virtual environments [8], allowing the user to feel virtual objects [9]. Burdea and Coiffet [8] in reference to Batter and Brooks [10], note how this became reality in 1971 and that many of today's haptics devices still use this same robotic arm-like arrangement.

The human force/torque perception operates at a far higher rate than our visual system. The latter can be fooled into seeing continuous motion by displaying 25 to 30 interlaced images per second. However, providing artificial haptic feedback to a user requires a significantly faster rate of "haptic image" update (around thirty times faster). This requires a significant amount of computational power for even simple models and has been a limiting factor in the development of haptics, only becoming a viable technology for desktop simulation within the last fifteen years [11].

The typical "Haptic devices" as they are sold commercially provide a mechanical I/O device with which a user interacts. The device will track one or more end effectors in physical space and provide force and/or torque feedback in a bidirectional interaction between a virtual environment and a user. Devices that provide tactile feedback are more commonly referred to as "tactile devices" but these are not widely available. Force feedback technology is explained in section 2.3. Section 2.3.1 describes the commercially available force/torque products and Table 1 provides a list of these devices together with their capabilities. A review of tactile interfaces which can be used in conjunction with these force/torque devices to provide a full haptic experience follows in section 2.4.

2.3 Force / Torque feedback devices

There are many commercial force/torque feedback devices ready to purchase and install off the shelf as simply as installing a new peripheral such as a web cam. Commercial force/torque feedback devices vary greatly in the degrees of freedom they offer, the size of their workspace, the force and torque they can apply, the shape of the end effector and maybe most significantly, in price. Different types of actuation used in haptics devices include: shape memory metals, magnetic, piezoelectric materials, electro-rheological fluids, DC electric motors and pneumatic as well as hydraulic actuation. DC motors are by far the most common method of actuation as they offer a good balance of force, weight, back driveability and cost.

There are many desirable properties of force feedback devices that will help to make a device more natural to use and enable optimal haptic interaction with a medical (and other domain) virtual environment (VE). Some of these properties are conflicting and so the advantages and disadvantages must be carefully considered in order to make an informed decision about the device of choice. For example, a device that is stiff will usually be made of metal and therefore have a larger mass. This in turn can have an undesirable higher inertia than a lightweight plastic device.

The term “Degrees Of Freedom” (DOF) relates to the number of transformations that can be applied to the end effector of the haptic device. A solid object in the real world has six DOF: three translational DOF commonly labelled x, y and z, that are used to describe the dimensions of a force feedback device’s “workspace”, or the space in which the end effector can be manipulated; and three rotational DOF (torque) around the x, y, and z axes, which are sometimes referred to as pitch, yaw and roll. Force/torque feedback devices advertise capabilities in excess of six DOF, such a device could provide all six spatial DOF and an additional one DOF scissor interface at the end effector. When the translation DOF are actuated, the device is said to provide force feedback. When the rotation DOF are actuated, the device is said to provide torque feedback.

The shape of the end effector of a force feedback device is important to produce a meaningful interaction with the environment and the grasp used to hold it directly influences the force/torque that can be applied. Grasping geometry can be classified as a precision grasp or a power grasp [13], with the user able to perform more dexterous or higher power tasks respectively. Most commercial force feedback devices come equipped with generic end effectors, shaped like pens, balls and tubes and some of these can be seen in Figure 2-5. Thimble interfaces into which a single fingertip can be inserted are also available on some high end devices. It is increasingly more common for medical simulations to use modified end effectors, in order to increase the face validity of the simulation. For example, a syringe shaped end effector can provide the extra validity needed to help a trainee nurse to suspend disbelief. On the other hand, such a modification may increase the cost of the simulator with no significant increase in training effectiveness in comparison to using an off the shelf stylus end effector. There are also examples of two commercial devices being combined to provide extra degrees of force feedback (force DOF) for a particular task. For example, Figure 2-1 shows Simquest's burr hole drilling simulation hardware in which two Novint Falcon devices are configured to give 5 force DOF to a single drill handle. This approach requires specific design engineering expertise to develop the solution. An example

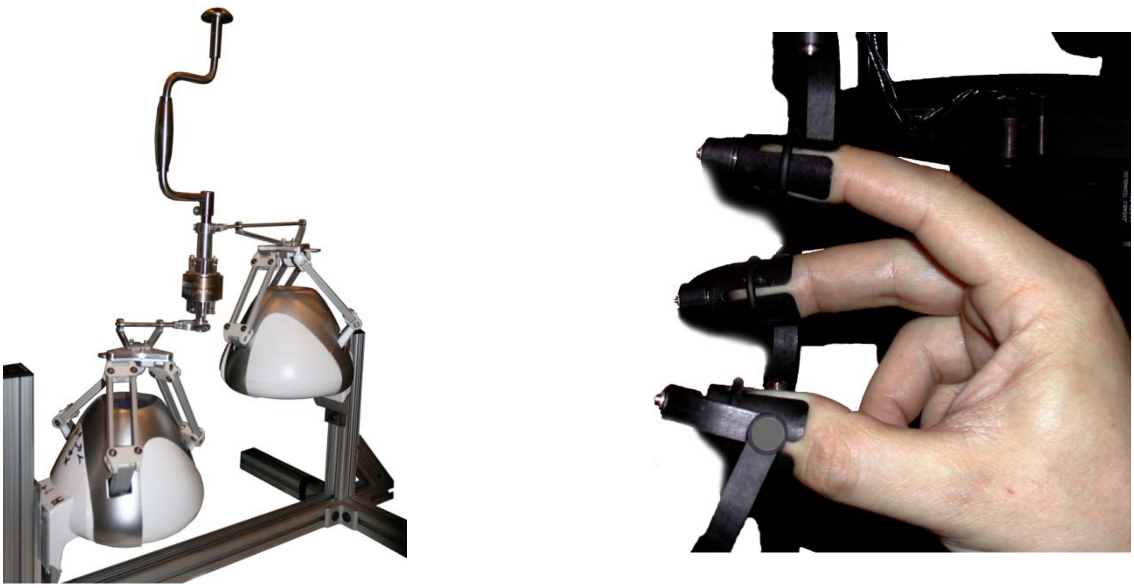


Figure 2-1 Left: SimQuest's burr hole drilling simulation hardware. Two 3 force DOF Falcon devices arranged to give 5 force DOF feedback to a single drill handle. Right: McKnight's three fingered haptic setup using three PHANTOM Premium devices to simulate grasping [12].

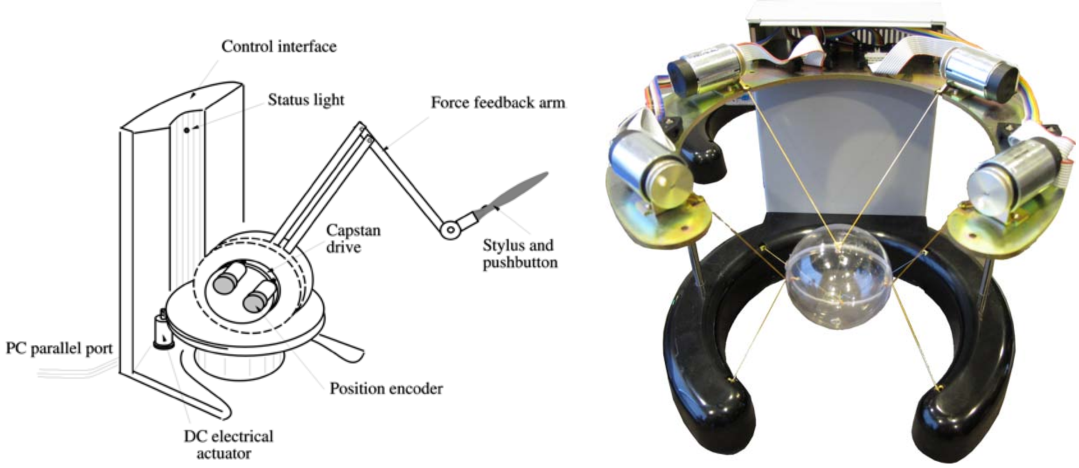


Figure 2-2 Left: Sketch of SensAble's Desktop 6 degree of freedom and three degree of force feedback linkage based device. Right: SPIDAR-G, 6 degree of force feedback tension based device [14].

of three thimble devices used together to simulate multi-finger grasping feedback can also be seen in Figure 2-1.

Force/torque devices used in medical simulation are typically grounded. However, haptics devices can be worn by the user or mounted upon mobile platforms to increase the range of motion provided. There are two broad mechanical categorisations of force/torque feedback devices: linkage based and tension based devices.

The most commonly used are linkage based devices, where one or more solid links connect an end effector to the devices base. These links are actuated using motors at the joints or by actuators situated within the device's base, Figure 2-2. The device's rigid links provide a robust method of tracking and transmitting force to the end effector. However, as the workspace of a device is increased, so does the weight and inertia of its links as they are lengthened. The force output of the motors must increase to overcome the increased link weight and the lever effect, and higher accuracy position encoders are required to maintain the equivalent functionality and accuracy of a device with a smaller workspace.

The second category of device, tension based devices, scale much better, see Figure 2-2. These devices use multiple flexible wires to suspend an end effector between actuators. The actuators, fitted with position encoders, maintain a slight tension on the suspending cables to provide accurate tracking until a force is to

Figure 2-3 Mounting the PHANTOM Premium upside-down for increased range of motion in suturing procedure. Image courtesy of R. Webster.

be applied. The direct cable connections allow a stiff contact to effectively be simulated but approximately only half of the workspace can be used to provide accurate force feedback. Outside this it is difficult to achieve force output without distortion (a pull towards the centre of the workspace). Theoretically, the number of cables used in the device plus one dictates the possible degrees of force feedback which can be provided (between 1 and 6 DOF) although a larger number of cables may increase the force fidelity.

The required refresh rate to provide realistic force/torque feedback is commonly accepted to be at least 1000Hz. However, this refresh rate is widely debated. According to Burdea [15], a minimum refresh rate of only 300Hz is acceptable. Conversely, a study by Booth *et al.* [16] using SensAble's Premium 1.5 to deduce the minimum acceptable haptic refresh rate, suggests that "a minimum acceptable refresh rate must lie within the 550-600Hz range". The necessary rate of update is dependent upon the stiffness of the surfaces to be simulated. A stiff contact between objects is better simulated by higher refresh rates, whereas lower refresh rates are satisfactory for softer objects. Additional methods can be applied to simulate touching stiffer objects such as combining force with vibrations at the end effector to reproduce the small vibrations felt upon object contact [17]. Typically, a trade off must be made between the accuracy of the haptics effects produced and the computation speed/haptic refresh rate required in the application. Batteau [18] presents experimental results demonstrating the unnoticeable haptic latencies of most humans. This time between action and realisation can be harnessed to improve the haptic response through further calculation without reducing the fidelity of the simulation.

Most linkage-based devices will have an irregular shaped workspace due to mechanical restrictions in the armatures, whereas many applications can only effectively use a cubed workspace as reaching the devices workspace limits can break the haptic illusion. Therefore, the workspace of a device is usually designed for the range of movement of a joint. For example, the extension and rotation of a finger, wrist, elbow or arm from the shoulder. Aside from translational workspace restrictions, the rotation (orientation) DOF of the end effector will also be limited in some respect, as to provide force feedback to the end effector, it must be attached by a link. This unavoidably restricts the rotation capabilities of the end effector. To maximise the usable workspace of devices, some developers have chosen to mount commercial force feedback hardware upside down. A suturing application which does this is shown in Figure 2-3 [19].

Other technologies aside from linkage and tension based designs for force/torque devices have been investigated. One such technique which has recently become a commercially available product is magnetic levitation [21], now produced by Butterfly Haptics (Pittsburgh, USA). Currently the magnetic levitation device can produce high force/torque feedback in a small workspace. Another technology in academic development is focused ultrasound radiation pressure [20] this technology promises haptic feedback without the use of any mechanical links, see Figure 2-4.

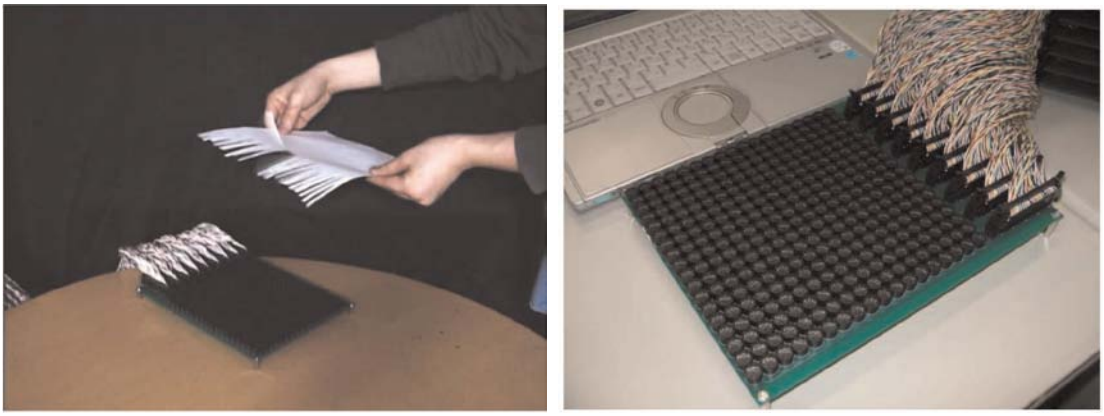


Figure 2-4 A non-contact tactile display based on the radiation pressure of airborne ultrasound provides haptic feedback through air [20].

2.3.1 Commercial General Purpose Force/Torque Hardware

In some cases it may be desirable to custom manufacture force/torque hardware to produce a device tailored to a specific task. However, expertise in hardware development is usually required to do so [22] and a long lead time on its production may be required before significant progress can being made in the development of software, unless developers test the simulation using other methods. Alternatively, many medical simulations, both commercial and academic, have been produced using commercially available force/torque devices. As commercial devices have undergone both testing and safety approval and are already in production, simulation development times can be reduced. To better tailor the generic devices to simulation of specific tasks, a variety of training simulations have made minor modifications to these existing devices. A brief summary of the commercial force/torque hardware designed for general, non task-specific use follows, some of which can be seen in Figure 2-5. Table 1 presents these device's characteristics as described by the manufacturers, although the methods of measurement between vendors may not be standardised. A subset of the device performance measures described Hayward and Maclean [23] is given, as unfortunately not all of the measurements are commonly provided by device manufacturers.

Of the tension based category, SPIDAR devices have been used for one [24] and two handed force simulation [25] and the latest SPIDAR-G, a 6 DOF haptic device, has been used for patient rehabilitation [14] although these are not commercial devices. However, devices based on this design are made by two companies. Mimic (Seattle, USA) manufacture two devices called the Mantis and Mantis Duo providing 6 DOF input and 3 force DOF to each force end effector. These are one and two handed devices, produced with a pen shaped end effector as standard, although this is interchangeable. A large scale tension based device, the INCA 6D is produced by Haption. The low visual occlusion of these devices can be a significant advantage over linkage devices when placed in front of a display. Another interesting, but non commercial, tension based force feedback design is a portable backpack device. The device has three wires used to exert a force on a thimble end effector [26].

Of the linkage based category, SensAble Technologies (Wilmington, USA) hold a large market share in the generic force feedback device market. The devices in their PHANTOM range, first developed by Massie and Salisbury [27], are the Omni, Desktop, and Premium 1.0, 1.5 and 3.0. Each device is fitted with a pen shaped end effector with at least one button. An optional thimble can be fitted to the Premium devices. The PHANTOM devices offer 6 DOF input and 3 force DOF with some of the Premium devices offering 6 force/torque DOF. An additional seventh degree of freedom pincer grip attachment is available for the Premium's. The Omni is the company's lowest cost device, and until recently was the cheapest device to provide 3 force DOF. The workspace of the Desktop and Omni devices was designed under the pretence that a small wrist-centred workspace is sufficient [27]. The Premium devices range in workspace size, with the 1.0 also offering a wrist centred workspace, the 1.5 offering a forearm centred workspace and the 3.0, movement from the shoulder. The Premium devices offer high force output, precision position sensing and a stiff interaction. Free space feels relatively frictionless.

Force Dimension makes three devices, the Delta, Omega and Sigma.7. The Delta haptic device was developed by the VRAI group from the Swiss Federal Institute of Technology (EPFL) [28] and was commercialised by Force Dimension in 2001. The 3 or 6 DOF Delta device offers a large workspace, high force output and gravity compensation. The wrist centred Omega can be purchased with an additional actuated seventh DOF but lacks torque feedback. The Sigma device, designed to control medical robots also provides a wrist centred workspace and all 7 degrees of high force/torque output. A lower quality, but far cheaper replica of the 3 DOF Omega device is available from a company called Novint Technologies Inc. (Albuquerque, USA). This device, called the Falcon, is designed for the computer games market and retails at around £200 (GBP). The reduction in cost has resulted in a device with a reduced stiffness, increased friction during free movement and a lower peak force output. Despite this, it is proving to be an appealing alternative as a low cost force feedback device for simulation.

Quanser Inc. (Ontario, Canada) who market five force feedback devices, specialise in real time system control. An armature based device that resembles that of SensAbles devices, called the Mirage F3D-35, offers high powered 6 DOF and 3 force DOF capabilities. Three pantograph based devices provide 2, 3, and 5 force DOF with the three and five degree devices using two pantographs to provide the extra degrees of freedom. The 3 DOF device was originally designed by DiMaio during his PhD [29]. Quanser's latest device, the HD2, offers 5 high force/torque DOF and can be seen in Figure 2-5. A sixth device currently undergoing patent requests was used in a needle insertion simulation [30], but no specific details have been released.

In addition to producing a tension based haptic device, Haption offer four other devices ranging in workspace sizes from wrist centred, to whole arm. Designed primarily for the manufacturing market, little work into medical simulation appears to have been carried out with these haptic devices.

The CyberForce force feedback system, previously from Immersion Coop. and now CyberGlove Systems LLC (San Jose, USA) was initially developed for telerobotic applications in the US army. The device is marketed as the world's first desktop whole-hand and arm force feedback device and provides 5 force DOF to the fingers (one point force per finger) through actuators and tendons. In addition, it offers 3 force DOF applied to the user's wrist and 6 DOF position sensing using a device developed by SensAble. The system is modular and a unit called CyberGrasp can be used to provide force feedback to the fingertips only if required. This is a high cost force feedback solution.

MPB Technologies (Montreal, Canada) market two force feedback designs. The Cubic is a 3 force DOF device with a parallel interface and the Freedom 6 is a 6 force/torque DOF device with the optional addition of a scissor interface (the Freedom 7). The Freedom 7 was originally developed at McGill University [31].

A high powered 3 force DOF device called the HapticMASTER, has been developed by FCS Control Systems [32] and now owned by Moog Inc. (East Aurora, USA). The device can exert a peak force of up to 250N, much more force than necessary in most surgical manipulations, but a modified version of this technology is being used by Moog in their commercial dental burr drilling simulator where stiff contacts must be correctly simulated.

In a 2009 state of the art review on the use of haptics in medical simulation published as part of this work, it was written "Other technologies for haptics devices have been investigated (e.g. Lorentz magnetic levitation [21]) that promise better haptic interaction fidelity in the future, but have not yet been incorporated into medical simulation solutions." The Maglev 200 has since been commercialised and a demo needle insertion simulation, that felt limited by the device workspace, has been developed to demonstrate the devices capabilities. The device simulated stiff contacts well and it is thought the device is well suited to dental simulations which require high force in a small workspace. The Maglev's small workspace may limit the achievable face validity in larger scale medical training simulations.

Figure 2-5 Commercial Force Feedback Hardware. 1^{st} line – Manufacturer, 2^{nd} line - Device Name

2.3.2 Force/Torque Summary

Choosing a commercial force feedback device for a specific application is not as simple as deciding upon the workspace required and selecting the single suitable device in this category. Even the largest workspaces have multiple devices available. The requirement to have 6 force/torque DOF feedback may lead to the device having a larger than required workspace. Budget restrictions can also limit the functionality that can be provided and often devices providing only 3 force DOF must be used where more force DOF would be preferable. An analysis of the task to be simulated should determine if the trade-off is valid. The force/torque capabilities of the device and the resolution of both position and rotation sensing also need to meet the requirements of the task. If a procedure requires millimetre translational precision whilst manipulating tools, a device with a coarser resolution than this would not be appropriate. Also the risk of providing too high fidelity of force/torque feedback can be as much of a problem as providing too little. A medical procedure where this scenario occurs is laparoscopic surgery. Here the tools enter the body through tight introducers that severely limit the interactions felt during a procedure. Providing too little or too much feedback may lead to negative training.

Use of commercial haptic devices will enable easier replication of a simulator after its development. The production cost may be lower if performing modifications to an existing device and such a device can be tested with already available software drivers. Production of a custom haptics device is an expensive and complex process only to be attempted by the experienced. In addition to the products listed in Table 1, there are some haptics devices available for specific medical procedures. Mentice (Gothenburg, Sweden), is widely known for their minimally invasive procedure training solutions (MIST and VIST). Since acquiring Xitact (Morges, Switzerland), who specialised in the manufacture of medical force feedback interfaces, Mentice now market the Xitact IHP for the emulation of endoscopic instruments and the Xitact CHP for the simulation of interventional procedures such as cardiology, peripheral interventions and interventional radiology. Also at the low fidelity, low cost end of the force feedback market, Logitech (Fremont, CA, USA) license and market many force feedback devices such as gaming joysticks (which have been used in some medical simulations). A now discontinued device, the 2 force DOF Logitech Wingman Mouse [33] released in 1999, also showed promise as a low cost force feedback device. At least one needle insertion simulation used this device [34].

Company	Devices	Degrees of Freedom	Degrees of Force Feedback	Workspace mm	Max Force Nm / Torque mMm	Stiffness N / mm	Price £x1000
SensAble Technologies www.sensable.com	Omni Desktop Premium 1.0 Premium 1.5 Premium 3.0	6 6 6 6 6	3 3 3 6, 3 6, 3	160 x 120 x 70 160 x 130 x 130 127 x 178 x 254 191 x 267 x 381 406 x 584 x 838	3.3 / 0 7.9 / 0 8.5 / 0 8.5 / 515 22 / 515	1.02 1.7 3.5 3.5 1	1.7 9 15 20 – 43 45 – 60
Force Dimension www.forcedimension.com	Omega 3, 6, 7 Sigma 7 Delta 3, 6	3, 6, 7 7 3, 6	3 7 3, 6	160 x 160 x 110 190 x 190 x 190 360 x 360 x 300	12 / 8.0 20 / 400 20 / 200	14.5 NA 15	12 - 20 52 19 – 36
Novint http://home.novint.com/	Falcon	3	3	101 x 101 x 101	~ 9 / 0	NA	0.2
Immersion Corp www.immersion.com	CyberForce CyberGrasp	6 5	3 5	304 x 304 x 495 Finger	8.8 / 0 12 / 0	NA NA	38
Haption www.haption.com	Virtuose: 6D Desktop 3D15-25 6D35-45 6D40-40 INCA 6D *	 6 6 6 6 6	 6 3 6 6 6	 129 x 120 x 120 500 x 644 x 350 1080x 900 x 600 400 x 400 x 400 Variable	 10 / 500 15 / 0 35 / 3000 100 / 10000 40 / 5000	 2.5 2 2.5 NA NA	 25 21 72 102 68 *
Mimic www.mimic.ws	Mantis	6	3	325 x 270 x 260	15.2 / 0	5.5	8
Quanser www.quanser.com	Mirage F3D-35 HD2 Pantograph: 2DOF 3DOF 5DOF	6 6 2 3 5	3 5 2 3 5	400 x 200 x 300 530 x 300 x 500 270 x 240 270 x 240 480 x 250 x 450	25 / 0 19.7 / 1725 10.1 / 0 10.1 / 255 9 / 750	2 10 3 3 10	30 51 – 59 17 21 42
Moog FCS Robotics www.fcs-cs.com/robotics	HapticMaster	3	3	1000x 400 x 360	250 / 0	10	37
MPB Technologies www.mpb-technologies.ca	Cubic 3 Freedom 6S F7S	3 6 7	3 6 7	330 x 290 x 220 170 x 220 x 330 170 x 220 x 330	2.5 / 0 2.5 / 150 2.5 / 150	NA 2 2	NA 21 25
Butterfly Haptics www.butterflyhaptics.com	Maglev 200	6	6	24 x 24 x 24	40 / 3600	50	30

Table 1 ***Degrees of freedom (DOF)*** *– Sensed degrees of freedom,* ***Workspace*** *measured in millimetres (note: methods manufacturers use to measure a devices workspace may vary),* ***Max Force/Torque*** *Force measured in Newton's, Torque in mili-newton-metre (mNm)* ***Stiffness*** *– Device stiffness N/mm as quoted by device manufactures (will vary significantly through workspace).* ***Price*** *in GBP is displayed in multiples of one thousand. An approximation at the time of writing based on conversions from multiple currencies, price ranges are given where device specifications are variable. * Haption's INCA 6D device price is dependent upon size of work space (large – greater than 2m)*

2.4 Tactile Devices

Tactile information is conveyed by compressing, stretching or vibrating and by varying the heat at the skin surface. Pasquero [35] provides in-depth information about the human tactile sense and a comprehensive list of 13 different tactile technologies. Note that the current limited understanding of human tactile receptors means that the design and optimisation of tactile devices is a slow iterative process. Of the developed tactile devices, most are large and lack the portability necessary to be used in combination with force feedback devices for a true haptic interaction, and these large devices are not reviewed here. To be useful for medical training purposes, a realistic feeling of touch, identical to that felt during the actual procedure, must be simulated. It may be useful to simulate heat, conveying information on the patient's temperature. This could be done with a temperature controlling glove [36]. No medical training simulation is yet known to incorporate this cue.

CompuTouch AS (Asker, Norway) have produced tactile devices that are small enough to be attached to a fingertip, see Figure 2-6. These tactile displays have a tilting metallic plate interface that can be controlled by electromagnetic coils within the device. Various tilting combinations can produce the illusion of touching complex surfaces.

In an approach similar to that first taken by Caldwell *et al.* [37], another small portable tactile device consisting of a 3 by 2 array of pneumatic balloons has been developed by Culjat *et al.* [38], see Figure 2-6. The device has been designed to add tactile information to the controllers of the Da Vinci surgical system from Intuitive Surgical, Inc. (Sunnyvale, CA, USA). The conveyed tactile information is suitable for training purposes.

The term vibrotactile refers to a vibration sensation that is more global than directed tactile feedback. Vibrotactile devices are now common place in games consoles to alert a user to an action such as being shot or driving a car over a rough surface and in mobile phones to alert the owner of a message or call when in silent mode. These devices comprise of a motor with an off-centred weight connected to the shaft. Some simulation solutions may find force feedback devices too expensive and opt to use vibrotactile displays to convey information such as operator mistakes or contact between two objects. A simulation adopting this approach is being developed for ultrasound scanning training [39]. The project uses the Nintendo Wii Remote controller, which incorporates 3D tracking and vibrotactile technology as a virtual ultrasound probe.

2.4.1 Tactile Summary

The tactile sense is an important cue and as research provides methods of producing tactile stimulation at an affordable cost and in small enough devices to be mounted upon force feedback hardware, the technology will become more widespread. Currently, there are very few commercially available portable tactile devices that can be used in conjunction with force feedback and as such tactile feedback is not commonly provided in visio-haptic medical simulation. Chapter 5 of this work attempts to address this and evaluates four methods of tactile actuation to simulate a femoral artery palpation: piezoelectric pads, micro speakers, a commercial pin array device from Aesthesis (Salford, UK) and hydraulic actuation.

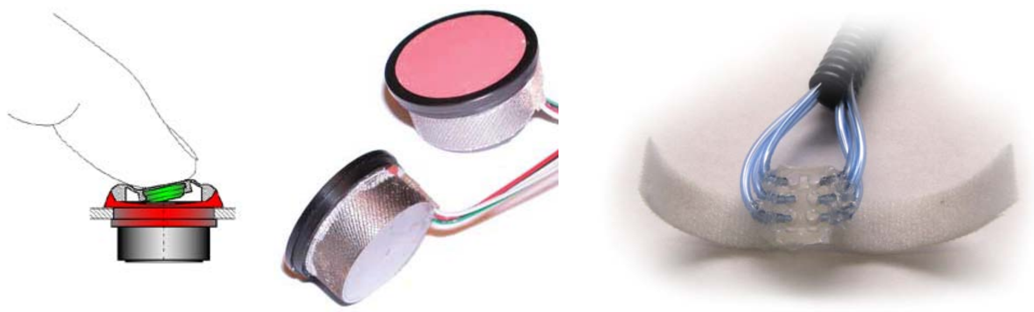


Figure 2-6 Left / centre: Compact tactile interface from CompuTouch AS (Asker, Norway). Right: A pneumatic balloon tactile interface [38].

2.5 Visualisation

As haptic feedback alone does not provide enough information to produce an immersive medical training simulation, invariably visual and sometimes auditory feedback is incorporated. Common methods of providing visual feedback are briefly explained below, together with consideration of how a haptics device can be optimally integrated with the various display types.

An LCD monitor is the default display that comes with any computer today and is sufficient to use when simulating some medical procedures. Many minimally invasive procedures, for example, require the practitioner to view video or information on a screen in a 2D format (e.g., fluoroscopic images of an interventional radiology procedure or ultrasound images), for which an LCD display would be sufficient. However, during a virtual ultrasound, a practitioner cannot look down at the simulated patient or at their hands, as they will only see haptic hardware, thus breaking the virtual illusion, Figure 2-7 - left.

Perhaps the simplest solution to this problem is to introduce a mannequin that represents a real patient. A force feedback device can be mounted under or above a mannequin and, for minimally invasive surgery (MIS) where long tools are inserted through two or more portals into the body, this approach is perfectly acceptable for tool manipulation training (e.g. [41] Figure 2-7 – right). However, if

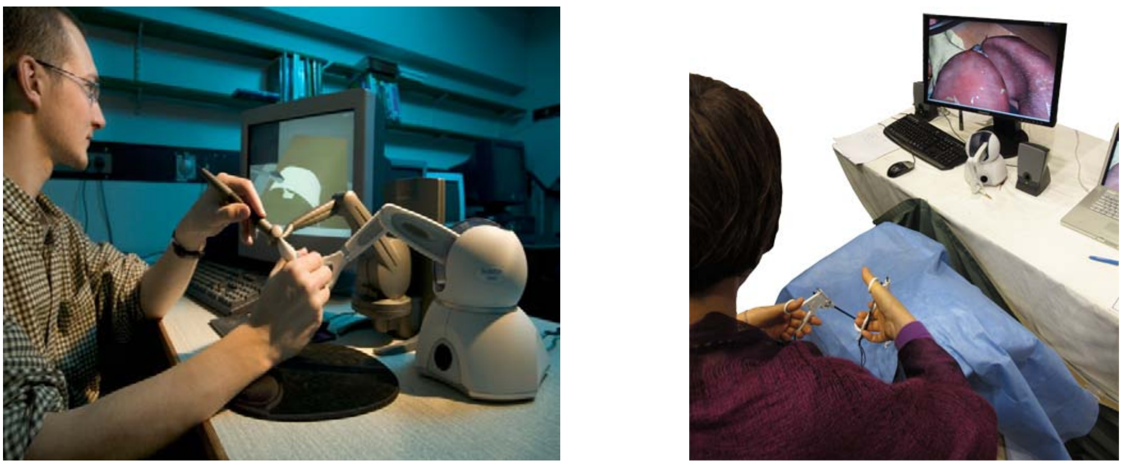


Figure 2-7 Left: BIGNePSi Bangor University's Ultrasound needle insertion simulation using a standard LCD monitor for visualisation of the patient interaction prior to using an Immersive workbench [40] Copyright: G.Davis, Menai Bridge. Right: Laparoscopic adjustable gastric band simulator [41].

training involves identifying an incision site and inserting a trocar (a sharp-pointed surgical instrument, used with a cannula to puncture a body cavity) for MIS tool access, a mannequin approach proves unsuitable as virtual force feedback for both procedures cannot be simulated. In addition, varying the simulated patient habitus involves producing many different mannequin structures.

A computer generated virtual patient will not have the physical restrictions of a mannequin, particularly if it can be displayed in three dimensions (3D). The binocular stereo component of depth cue information is often exploited to reproduce 3D visual effects, provided that the user is not stereo blind (possibly as many as 20% of the population are [42]). Stereoscopy projection is performed by displaying two images with a calculated binocular disparity to the left and right eyes individually [43].

Time sequenced (or active) stereo displays project a right eye, then left eye image in quick succession. In synchronisation, shutter glasses (Figure 2-8) worn by the viewer, occlude the right eye at such a time that only the left eye image is displayed on screen, and vice versa. The glasses contain liquid crystal and a polarising filter. The lens is transparent, but when a voltage is applied it becomes dark. The brain processes these two separate images and a stereo image is perceived. When performed at 120Hz [44], the result is a seamless stereo image. Lowering the refresh rate below 120Hz increases the chance the user will see the flickering effect, a particular problem of older and low-cost versions of the technology. Time sequenced stereo can be used in both large multi viewer displays and desktop single viewer displays but even though this consumer market has seen a

Figure 2-8 Infrared shutter glasses and controller.

boom in the last two years, active stereo glasses (e.g. RealD (Beverly Hills, USA), NuVision (Oregon, USA), XpanD (Pasadena, USA), Nvidia (Santa Clara, USA)) are still high in cost compared to static polarising glasses.

Polarisation is a common method of large scale 3D projection, which simultaneously projects two orthogonally polarised images onto a screen. Glasses with polarising filters aligned to these two images are then used to split the image to the appropriate eyes. The glasses are relatively cheap in comparison to time sequenced shutter glasses, and they offer an adequately robust visualisation as long as the filters and image remain aligned. The rapid growth in consumer 3D television has led to the production of many desktop stereo LCD's using polarisation technology (e.g. Zalman (Seoul, Korea), iZ3D (San Diego, USA), MiraCube (Incheon, South Korea), 3DInfotech (Irvine, USA)). One example made by Zalman uses linear polarisation and polarises alternate horizontal pixels of the LCD display. This leads to a loss of screen resolution, with the number of visible pixels per eye halved. The display has a small optimum viewing area that the user must remain within so as not to observe cross talk, i.e. one eye starts to see parts of the image intended for the other eye. A popular screen in the stereo gaming market is the 22" iZ3D screen. This is a linear polarised screen combining two stacked LCD units that project a full resolution images to each eye. A higher cost circular polarisation technique can also be used, allowing the user to tilt their head without losing the stereo effect (e.g. MiraCube).

Planar (Beaverton, USA), 3D Infotech (Irvine, USA) and Omnia (Madrid, Spain) also market polarised displays which, unlike conventional polarised monitors, use two LCD displays to project separate images to a single half mirrored screen suspended in-between the monitors. This approach allows a high resolution image to be rendered for each eye whilst static polarised glasses can be used. Collocation of the perceived stereo image and haptic device cannot be achieved however, this approach has been implemented by SimQuest (Silver Spring, USA) during testing of their OpenSimSurg simulation with two Omni force feedback devices from SensAble technologies, see Figure 2-9.

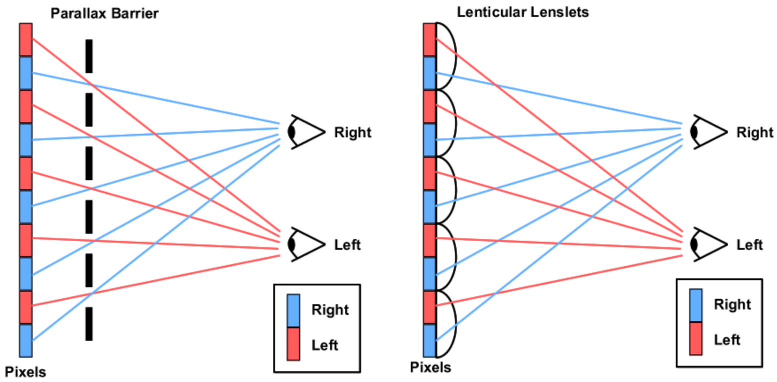


Figure 2-9 Left: Haptic interaction using a Planar display (Picture courtesy of SimQuest). Centre / Right: Methods of passive autostereoscopic display. Images from www.3d-forums.com, last accessed 06/11/2010.

Autostereoscopic displays [45] simultaneously present left and right eye images to the corresponding eyes without the need of glasses by splitting the light at the projection plane. The most basic implementations halve the resolution of the resulting image, splitting alternate pixels to each eye using a barrier mask, or lenslets, see Figure 2-9. This approach will only work for fifty percent of head positions due to the nature of the projection and the unknown position of the eyes. Active autostereoscopic systems use head tracking to adjust the optimal view to follow the position of a viewer's eyes. Varying eye widths and slow tracking can cause problems with this technology. An alternative approach is to increase the projection complexity to include multiple simultaneous views. This technology is produced by numerous commercial suppliers, e.g. Philips 3D Solutions (Eindhoven, Netherlands), NewSight (New York, USA), 3D Super (Rochdale, UK), Spatial View (Toronto, Canada) and SeeReal (Munsbach, Luxembourg).

Head mounted displays can be worn on a user's head to provide either a two or three dimensional view of a virtual world. These have not widely been used in medical training simulations. HMDs typically provide a low resolution image and are expensive due to the high cost of technology miniaturisation. The displays can also be bulky to wear although modern hardware is beginning to overcome these issues. A specialised subset of HMD's for augmented reality allow the user to see through the display to the world outside, whilst enabling virtual objects to be integrated into the visualisation. This can be achieved with the addition of video cameras in front of eye level screens. Example products are manufactured by Sensics (Columbia, USA), Trivisio (Kaiserslautern, Germany), nVisor (Reston, USA) and Vuzix (Rochester, USA), see Figure 2-10. To achieve a realistic visualisation, the user's head movements must be accurately tracked either visually, finding markers extracted from the live camera feed, or via an external tracking device. Errors in tracking cause a significant problem during this visualisation approach and dependent upon application, millimetre misalignments can break the augmented reality illusion. In most displays, a parallax between the camera and the user's eye also introduces an error in movement, and tasks performed close to the users eyes can be disorientating, although through the addition of mirrors this can be overcome. The field of view

Figure 2-10 Left: 920AR Consumer level augmented reality HMD from Vuzix. Right: Fixed position binocular augmented reality concept product from Virtual Proteins

of such displays is typically much narrower than that of the human eyes giving a feel of tunnel vision. A fixed position augmented reality concept product, designed in part for medical simulation, from Virtual Proteins BV (Eindhoven, The Netherlands) appears to offer the same capabilities as video-see-through displays but restricts the user's freedom of motion by fixing the display to a stand. This restriction would alleviate the device weight and simplify real and virtual world alignment. However, a promotional medical visualisation demo which includes haptics, does not demonstrate the displays augmented reality capabilities at the time of writing.

In a large number of medical interventions, the practitioner will stand over the patient who lies beneath them. As such, the most frequently adopted display for medical training simulations (as seen in the following review) is the immersive virtual workbench (first proposed in [46]), Figure 2-11. Using active stereo, the virtual workbench creates the illusion of 3D objects lying below the horizontal projection plane. The computer image is normally projected onto a semi-transparent screen that allows the user to see both their real hands below the mirror and projected virtual objects in collocation [47] if accurate eye tracking methods are employed. Commercial semi-transparent displays are marketed by

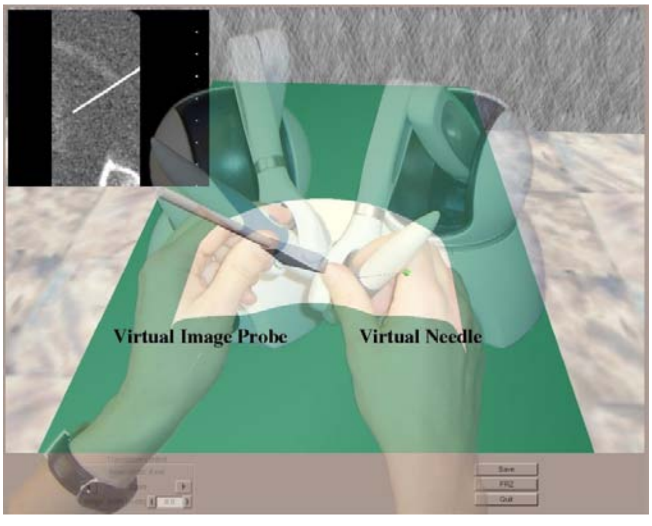


Figure 2-11 Left: Immersive workbench, Right: An artist's impression of a user's eye view of visio-haptic collocation using BIGNePSi: Bangor University's Ultrasound needle insertion [40].

Reachin (Stockholm, Sweden), SenseGraphics (Kista, Sweeden), and Immersive Touch (Chicago, USA). As Figure 2-11 shows, the arrangement of the LCD or CRT monitor and mirror, provides free space below the projection plane in which tools (haptic devices) can be placed. Like in standard time sequenced stereo approaches, the flicker of the shutter glasses can cause some users eye strain, but the display's biggest shortcomings are the unrealistic occlusion of the real world space below the screen and the focal conflicts it causes users [48]. Only around 50 percent of the image projected onto the horizontal plane is reflected to the user's eyes with the other half originating from below the mirror so that the user can see their own hands. In a situation where a virtual patient is believed to be lying underneath the mirror, the image is projected on top of the user's hand, causing an unrealistic viewing illusion. A focal conflict is also caused, as the focal distance of the patient image is closer than that of the user's hands. To see a clear virtual image, the user's focus must fall upon the horizontal projection plane, but to see their hands, they must adjust this focus through the glass to their hands below where they perceive the patient to be lying. Because of this, some users find it hard to settle their focus at a single distance, as neither the projection plane at one distance nor the users hand at another captures all of the information in focus they expect to see.

2.6 Haptics Libraries & Modelling

Several Application Programming Interfaces (APIs) have been produced to aid in the construction of haptics rendered virtual environments. These APIs can implement common methods of modelling forces, provide physics simulation, offer different methods of collision detection and interface with most of the products listed in Table 1. However, they can be slow to support new advances and so it is often preferable to develop the core simulation routines separately. Licensing methods can also vary. SensAble Technology's OpenHaptics API is a commercial C++ library, but it is free for academic use. OpenHaptics provides cross platform support and, with respect to programming, it resembles the OpenGL graphics library. It only works with SensAble's force feedback devices, but these are the most popular products today.

Chai3D [49], an open source library, includes both graphics (using OpenGL) and force feedback components and is written by academics in C++ to be platform independent. It is a comparatively light weight API, but it allows extensions to be easily added (such as ODE physics engine support), and also offers support for a range of commercial force feedback devices.

The H3DAPI, is a haptics development platform including graphics support. It is available under either an open source or commercial license dependent upon usage. According to the development requirements X3D, C++ or Python can be used. The API is maintained by SenseGraphics and provides support for Force Dimension, Novint, Moog FCS Robotics and SensAble force feedback devices. A scenegraph architecture is used to reduce the complexity of environment definition.

SensAble's devices are the most widely supported of all the haptic manufacturers, and some additional APIs that provide singular support for these are XVR by VRMedia (Pisa Italy), and OpenSceneGraph (through an additional sub-library called osgHaptics).

ReachIn market two commercial haptic API's that support various device manufacturers. One, the self titled "Reachin API" is compatible with C++, VRML and Python with visual components rendered using OpenGL. The second is HaptX, a haptics-only engine designed for the games market. Haptik [50], like HaptX, also provides a basic abstraction layer for force feedback hardware. It is an open source library allowing a wide range of devices to be accessed through a common interface.

The VirtualHand API, formerly from Immersion and now from CyberGlove Systems LLC, is a C++ simulation development API for hand interaction. It supports CyberGlove's gloves, as well as their CyberForce system and various hand tracking hardware. MHAPTIC [51], is another hand interaction simulation environment catering for two handed manipulation. It is not freely available.

Specific to medical applications, OpenMAF [52], is an open source framework for computer aided medicine and is based on the VTK toolkit. Haptic feedback is not the main focus in this project but is provided through SensAble's OpenHaptics interface. SPRING [53] is an open source, real-time soft-tissue simulation platform developed by Stanford University. SPRING's main focus is minimally invasive surgery and a limited number of force feedback devices are supported. SOFA [54] is a framework aimed at real time medical simulation and the development of new algorithms. Haptic support for SensAble's range of devices and Mentic's Xitact IHP force feedback device is provided. Mass-spring and FEM deformation models, fluid models and a large array of collision detection features are also provided. GiPSi [55] is an open source framework for developing human organ level surgical simulation. The structure of the API is designed to use more general models than those used in SOFA whose models must be tailored toward specific methods. ESQUI [56] is a platform independent laparoscopic surgery framework, although it is intended that the system can be applied to any surgical simulation. Using XML style scene descriptions, the ESQUI platform advocates the Simulation Reference Markup Language (SRML) as a standard for information exchange between simulators. One commercial laparoscopic haptics device is supported at the time of writing. VSS [57] is also a framework in development for virtual surgery simulation offering a cross platform object oriented system with support for both haptics, GPU processing and semi-automatic segmentation.

2.7 Deformable modelling

One of the hardest problems in medical simulation is the modelling of tool and/or hand interaction with soft tissue. The collision detection parameters must constantly change as the surface of the soft tissue deforms. This deformation is due to the actions of many different material layers within the tissue whose properties are little known and typically too complex to model in real time. Simplifications are usually made to enable a sufficiently rapid response time. Bodily functions such as respiration, changes in blood pressure and contraction of muscles will also result in tissue deformation. Moore *et al.* [58] provide a broad overview of deformable models for a wide range of disciplines. A more detailed survey by Nealen *et al*. [59] focuses on deformations for computer graphics animations where visual fidelity is the main goal, whilst Meier *et al*. [60] survey deformation techniques for real-time surgical simulation. There is little overlap between these two surveys, highlighting the different challenges faced between animation and surgical simulation. These include strict real time behaviour, acceptable accuracy in modelling highly complex tissues and the capability to cut models for surgical simulation. A comparison of techniques demonstrates the trade-off between computational efficiency and realism. More recently Famaey *et al*. [61] provides a detailed review of the key continuum mechanical models for surgical simulators for minimally invasive surgery.

2.8 Haptic Devices in Medical Simulators

Haptics has primarily been used in medical simulators to enhance training applications, and a variety of different medical specialties have been covered. This survey first reviews palpation simulations, as this is the first task performed in many procedures. The review then moves on to needle insertion simulations, often the next step once a palpation has located a site for needle puncture. A needle puncture is required to anesthetise the patient before more invasive procedures, in order to obtain blood and biopsy samples and to introduce tools in minimally invasive procedures such as interventional radiology. The simulation review moves on to Laparoscopic simulation, currently the most widely validated discipline of virtual training simulations. The field of endovascular simulation, the subsequent step after a palpation and needle insertion in an interventional radiology procedure is then surveyed, after which the varied simulation approaches used for knee arthroscopy training are presented. The survey focuses on the haptics hardware and visualisation method used in each approach and is used to draw conclusions on an optimum method of virtual palpation and needle insertion simulation in Chapter 3, the exemplar surgical simulation developed in this thesis. Another application area where haptics devices are used for procedure training is dentistry (e.g. [62]). However, this speciality along with others is outside the scope of interest of this thesis.

2.8.1 Palpation

Palpation is where a practitioner presses upon an area of interest with their fingers to locate landmarks beneath the patient's skin and feel for the presence or absence of anatomic and/or physiological features or abnormalities. This could be for patient assessment or guidance for an intervention. Palpation commonly requires multi finger, multi contact tactile manipulations, a challenging task for a medical simulator to implement, and so is usually ignored. When included, the manipulation is usually greatly simplified.

An early example of palpation simulation is that of a knee palpation using a Rutgers Master force feedback glove interface [63], work that was later extended to simulate the palpation of subsurface tumours [64] [65]. In this work, the visio-haptic feedback was not collocated and a low quality 3D hand avatar was displayed to visualise the haptic interaction. Later work from Rutgers University did not use their glove interface but adopted a thimble approach. This simulator for prostate cancer diagnosis [66] used a PHANTOM Premium device mounted upside down to modify the device's workspace. However, only one finger received force feedback and no tactile feedback was provided.

A training simulator for palpatory diagnosis in osteopathic medicine, the Virtual Haptic Back (VHB) [68] [69] uses two PHANTOM Premium 3.0 devices with thumb thimble interfaces. *In-vivo* measurements of the back compliance have been recorded to improve simulation accuracy [70]. This is the only virtual palpatory simulation to date that offers virtually simulated collocated visual and haptic feedback, with an investigation into the optimal visualisation method for this simulation performed [67]. Three visualisation methods: a standard LCD placed behind the two haptic devices, a head mounted display and an immersive mirror display were evaluated as part of the VHB simulation development process using three evaluation criteria: interface realism, comfort and ease of use. The results of these tests are presented in Figure 2-12. For realism, the HMD was rated to give higher fidelity feedback than the LCD display, but it did not rate as highly

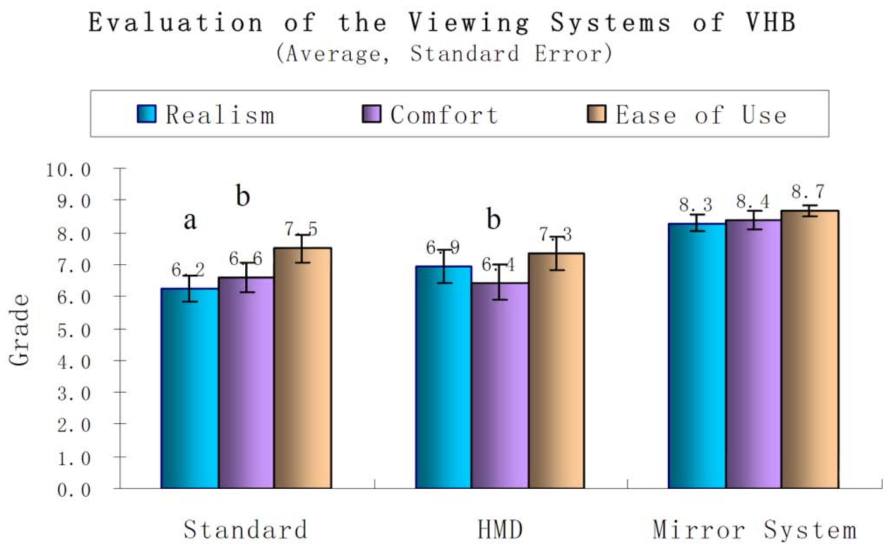


Figure 2-12 Virtual haptic back. Viewing technology evaluation results [67].

as the Immersive workbench. It was predicted that this was because the user could not see their fingers whilst wearing the HMD and its limited field of view (FOV) of 27 degrees unrealistically limited the amount of virtual world the users could see at one time. In their study the HMD also scored worse in both the "comfort" and "ease of use" categories with the immersive display scoring highest. The relatively compact HMD was rated as uncomfortable to wear. The immersive display was praised for its intuitiveness, although the aforementioned occlusion limitations of such a display were cited as problematic.

Several palpation simulators use the PHANTOM Desktop. Stalfors *et al*. [71] designed a remote diagnosis (telemedicine) simulator for malignancy in the head and neck area using a PHANTOM Desktop stylus with a VRML model created from computed tomography (CT) data. Chen *et al*. [72] proposed a model for index finger palpation of the upper leg, simulating interaction with skin, muscle, ligament, and bone using a PHANTOM Desktop device, although no specific medical application of this palpation is planned. They discussed varying deformation models for palpation and use a contact model based on Hertz's theory from contact mechanics [73]. Using a stylus interface offers low face validity unless a tool is used during real world palpation.

A custom thimble for an Omni has been fabricated in a brachial pulse palpation simulation [74]. This thimble transfers the force to a single finger but requires an additional two fingered grip to counter the un-actuated torque encountered, see Figure 2-13.

Figure 2-13 Left: A brachial pulse palpation simulation [74] Right: Haptic Interface Robot (HIRO) [75]

Immersion filed for a patent in 2001 for a haptic interface for palpation of a pulse [76]. This haptic interface closely resembles their design of the Wingman Force feedback mouse [33] with minor amendments. The simulation has not yet appeared commercially.

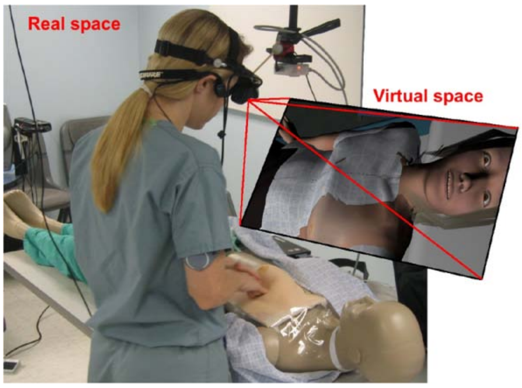
A five fingered breast palpation simulation [75] developed by Gifu University uses the Haptic Interface Robot (HIRO), see Figure 2-13. This simulation has recently been demonstrated using the updated HIRO III robotic hand. A finite element soft tissue model is used to simulate the deformation of the breast tissue. Five magnetically attached thimbles, one for each finger, attach to the fingers of the robotic hand and provide an individual 3 DOF point force to each finger. This simulation is rendered on a computer monitor with no visio-haptic collocation and no declared intention to simulate an intervention such as a biopsy. Only force feedback is provided. A different breast palpation simulator [77] uses the stylus of a PHANTOM Premium, a 6 force DOF feedback device. A third unpublished (video only) [78] breast palpation simulation was performed by Stanford Robotics lab as part of their multi point haptic interaction research. In this simulation, a two finger pinch attachment for a Premium device is used. The breast tissue is deformable and although when compared to a pen, the interface offers a more intuitive method of touch, the pinch interface does not appear to effectively represent a palpation.

In the field of veterinary medicine, a rectal palpation training simulator for bovine fertility examinations has been developed at the Royal Veterinary College (London, UK) using a PHANTOM Premium 1.5 force feedback device with a thimble interface, see Figure 2-14. The device is housed within a fibreglass model of a cow [79]. Other work from the same group includes a horse ovary palpation simulator, HOPS [80], and more recently a simulation of feline abdominal palpation [81], also in Figure 2-14. The latter simulation uses two PHANTOM premium devices to provide force feedback to one finger of each hand on either side of a cat mannequin's torso. The mannequin provides simulation context and a tactile stimulus as the trainee's fingertips brush past the mannequin's fur. The

Figure 2-14 Left: A rectal palpation training simulator for bovine fertility examinations [79] Right: a simulation of feline abdominal palpation [81]

developers state that the high fidelity force feedback devices are necessary to convey the haptic cues required. Despite the high cost, all three of the simulations are used in the veterinary curriculum at the college, with the Premium devices swapped between the simulations accordingly. Later work by another group has looked to improve the virtual bovine rectal palpation by adding deformations to the rectal model [82]. This work appears to be in its early stages and uses only an Omni stylus for the palpation.

The direct practitioner/patient contact in palpation requires simulation of both force and tactile feedback. Although commercial force/torque feedback can be simple and inexpensive to incorporate, the lack of commercially available tactile devices limits current solutions for palpation simulation. A breast palpation simulation which allows the user to palpate a visually virtual patient whilst using a mannequin to provide passive haptic feedback has been produced by Kotranza *et al.* [83], see Figure 2-15. This simulation overlays a virtual human avatar on top of a mannequin and during simulation a female avatar's gown can be removed to expose the breast to be palpated. This image, viewed through a HMD, is collocated with the real world mannequin offering passive haptic feedback as a trainee reaches down to palpate the virtual visual avatar. The breast phantom contains pressure sensors, and as it is palpated, audio cues are provided to impersonate the patient. During simulation, the practitioner cannot see their hands through the HMD and the virtual patient's facial expressions do not change, although the goal of the virtual simulation was to improve "interpersonal touch" during medical



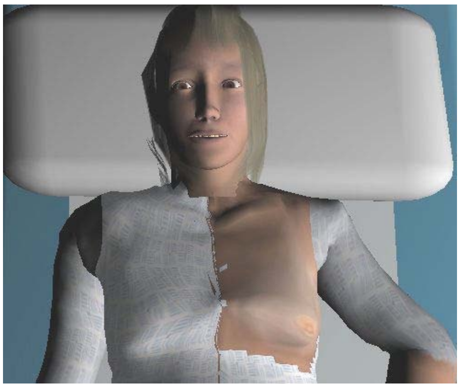


Figure 2-15 A breast palpation simulation using passive haptics and a HMD visualization [83].

simulation. Interpersonal touch, where patient expression and reaction is a "critical component of non-verbal communication" does not appear to be addressed in this simulation, as the only communication is verbal. As no consideration was given to a need for patient variability, no interventions such as biopsy are to be simulated and the simulation relies solely on imitated verbal responses to guide a successful palpation. Guidelines outlined in Chapter 2.12 of this thesis suggest that the simulation may be better simulated using a full body force sensing mannequin and audio speakers.

2.8.2 Needle Insertion

For realistic simulation of a needle puncture, 6 DOF and 6 force DOF feedback should ideally be simulated, but many simulators opt to reduce costs in order to allow for a wider adoption of the training simulation.

The Mediseus Epidural simulator (Medic Vision) was a commercial example of a needle insertion simulation using a commercial force feedback device. In this simulation, a SensAble PHANTOM Omni is encased inside the toaster shaped system, which offers a modified syringe end effector pre-inserted into a fixed insertion point, see Figure 2-16. This transforms the 3 positional force DOF to one positional and two orientation torque DOF. To further reduce costs, Medic Vision had investigated replacing the Omni with a Novint Falcon device, which would have reduced the cost of this component by 80%. However, Medic vision ceased trading before this modification went into production. The epidural simulation software can be run from a laptop. A vocal response is given if the user makes mistakes and an objective report is produced to provide feedback for the student [84]. A user cannot pick their desired point of entry into the model, nor is the mannequin model representative of a complete human back, so this must be visualised on an attached monitor. An alternative epidural anaesthesia simulation is EpiSim from Yantric Inc (West Newton, MA, USA), which was originally developed by MIT [85], see Figure 2-16. This uses a similar hardware setup to the Medius Epidural, but replaces the small box form factor with a mannequin model and pre inserted needle. The simulator takes a high fidelity rather than low cost approach, first using a SensAble PHANTOM Desktop before changing to use a 6

Figure 2-16 Left: The Mediseus epidural simulation by Medic Vision. Right: The EpiSim epidural simulation by Yantric Inc

force/torque DOF Premium device to provide better simulation fidelity [86]. An LCD screen is used to display fluoroscopic x-ray images of the lumbar region and a three dimensional model of the needle, spine and tissues. Simulation of variable tissue thicknesses accommodates for patient variability, whilst the mannequin model offers a visually static representation for the fixed position puncture.

Novint have produced multiple custom medical grips for the Falcon. In one commissioned project, a grip incorporating a real syringe and fluid was developed, Figure 2-17. Although the company have considered producing a 6DOF end effector as a consumer product, no such grip has yet been released, possibly due to a desire to keep device costs low.

A Falcon force feedback device has been used as the master interface to guide a robotic MRI prostate needle procedure [87]. The ball shaped end effector has been modified to look like a biopsy needle and provide a more intuitive interface, see Figure 2-17.

Chinese acupuncture is another field in which needle insertion simulation is being employed for training purposes. A simulation by Heng *et al.* [88] uses two computers to split the computation workload and is displayed in stereo upon a mirrored immersive workbench with a force feedback device below. A PHANTOM Desktop device was used during testing but this was changed to an Omni in the presented simulation. The original Omni end effector was unclipped and a real

Figure 2-17 Left: One of Novint's custom force feedback grips for needle insertion simulation. Centre: A Falcon used as a master to guide a robotic MRI prostate needle brachytherapy procedure using modified end effector [87]. Right: Chinese acupuncture, original Omni end effector unclipped and a real acupuncture needle is taped to the jack connector [88].

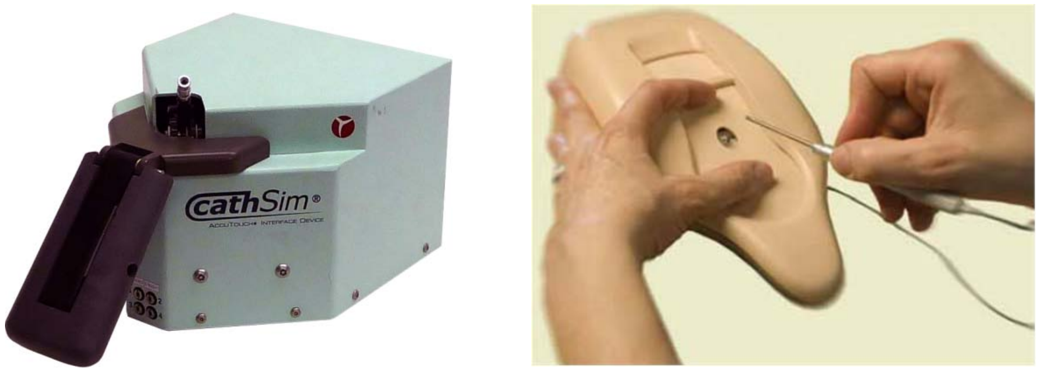


Figure 2-18 Left: Immersion Medical's CathSim AccuTouch System for intravenous access simulation Right: Laerdal and Immersion's Virtual IV intravenous access simulation replaced the CathSim. This new simulator looks more like a regular mannequin. The needle insertion point is clearly visible.

acupuncture needle was taped to the jack connector in its place to provide a more realistic haptic interface.

An initial step in many interventional procedures is the insertion of a needle or trocar, as an introducer for other tools. Most current commercial simulators for minimally invasive surgery (MIS) are built with the introducer already in place, e.g. the Mentice's Procedicus VIST - a simulator for vascular interventional surgery where forces can be applied to a real catheter and guidewire. Other commercial simulators for interventional procedures use a similar approach to reduce simulation complexity and cost. Immersion Medical produced a now discontinued intravenous access simulation device named the CathSim AccuTouch System [89] [90], Figure 2-18. It contained a needle carrier with 3 DOF movement, and 1 force DOF feedback. Movement consisted of pitch, yaw and depth of insertion. Force could be applied along the depth of insertion vector, either whilst inserting or retracting the needle. This 1 force DOF allowed the forces felt as a needle passed through different tissues to be recreated. The CathSim has now been replaced by the Virtual IV intravenous access training simulator, Figure 2-18. First developed by SimQuest, the device closely represents a mannequin, with a small fixed position area for palpation below a clearly visible pre-defined hole into which a needle can be inserted. The device was acquired by Laerdal (Stavanger, Norway) and is produced by Immersion.

A simulator for percutaneous vertebroplasty (a minimally invasive procedure performed to bind spinal fracture components) has been developed by the

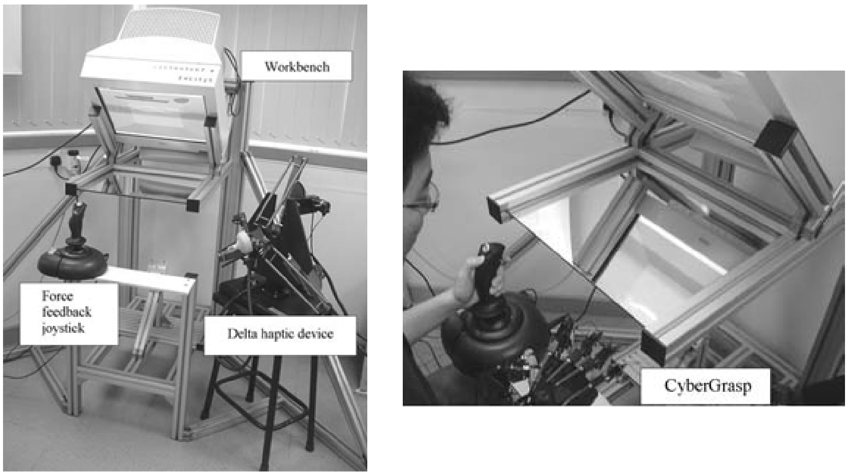


Figure 2-19 A simulator for percutaneous vertebroplasty [92]. A force feedback joystick, a Delta haptics device and a CyberGrasp glove are all used during the simulation.

National University of Singapore [91], see Figure 2-19. It is claimed that the simulation provides advanced feedback, requiring an array of high cost technology: a force feedback joystick, a Delta haptics device and a CyberGrasp glove. This minimally invasive procedure requires the practitioner to deliver cement from a needle at a specified critical rate to bond the fracture components. An immersive mirror display is used for stereo visualisation during simulation. The researchers have applied biomechanical models to model the bone needle insertion [92]. Future work aims to produce a low cost version of the simulation.

A trainer for catheter insertion has been developed at the Centre for Advanced Studies, Italy [93]. Using a head tracked stereoscopic viewing system and a PHANTOM device, the solution was reported to be “sufficiently representative of a real catheter insertion” by a surgeon in the field but not validated. The soft tissue component of the simulation uses an incremental viscoelastic model [94]. Forces are applied using a lookup table.

Simulations of ultrasound-guided needle puncture using two Omni devices have been implemented by Forest *et al.* [95] (Figure 2-20), and Vidal *et al.* [40]. The latter simulator, which is developed in part at Bangor University, is viewed on an immersive workbench. One of the Omni styluses is replaced with a custom, ultrasound probe shaped end effector, and the second Omni is used for the virtual needle. The simulation uses the graphics processing unit (GPU) to generate ultrasound-like images from CT data. This work is currently being developed and enhanced with the intention of producing a commercial biopsy simulator [96]. Using the default Omni stylus as the needle proxy impacts upon the obtainable simulation face validity and Bangor's simulation is now using the Omni needle modification described in Chapter 6 to overcome this.

A commercial haptic ultrasound training application, ScanTrainer for endovaginal scanning has been developed by MedaPhor (Cardiff, UK), see Figure 2-20. The forces to be simulated as the long scanning probe is inserted have parallels with needle insertion. This simulation is implemented using the H3D software and a single Omni force feedback device. Another endovaginal simulation VEUSim [97] is in development at Drexel University.

A spine needle biopsy simulator for training and task planning [98] uses a mannequin to transform three translational force DOF to one position and two orientation force/torque DOF much like the Mediseus Epidural simulator. However, the 3 force DOF Premium device is placed outside the mannequin structure allowing the attached needle end effector to be inserted into the mannequin rather than being pre-inserted. A single fixed insertion point for needle puncture is permitted. It is thought that that approach would slightly increase the face validity of the simulation. A 3D visual user interface at the side of the mannequin allows the trainee to follow the needles movement toward a target lesion. It is not clear from the literature what validation studies have been carried out.

Lumbar puncture simulators have been developed over the last 15 years and

Figure 2-20 Left: HORUS a simulation for image-guided needle puncture [95]. Right: ScanTrainer from MedaPhor, an endovaginal ultrasound training simulation.

demonstrate well how technology advances are aiding continual simulation improvements. An early (1994) lumbar puncture simulation produced using a custom haptics device [99] suffered from low bandwidth actuators that caused problems simulating stiff objects, and a large graphics delay that only allowed the procedure to be performed slowly. Moving forward 6 years, a later simulator used a PHANTOM 1.5 device, a mannequin and a more powerful computer with OpenGL support [100]. With a goal to produce a lumbar puncture simulator that was "effective, not cost prohibitive, relatively simple to maintain, and truly usable", results were said to be "encouraging" with one of the future goals to accommodate for patient variability. More recently, a lumbar puncture simulator using a 6 force/torque DOF Premium force feedback device has been produced [101]. The six degrees of force/torque feedback allows for accurate simulation of all possible forces/torques felt whilst inserting a needle and importantly if a user releases the needle whilst it is inserted, it will remain in the correct position and orientation within the simulated tissue, see Figure 2-21. This simulator, viewed on an immersive stereo mirrored display, calculates needle tip resistance using CT density data and in addition models the forces acting on the needle shaft. This involves restricting rotation and transversal motion, as well as increasing needle friction as depth increases. The tissue model is static with both a 2D and a 3D stereo view provided during the simulation. Testing was performed by users with varying medical experience, who concluded they could tell the difference between different tissues. The PHANTOM Premium's 6 force/torque DOF were reported to facilitate "realistic needle behaviour" although the standard stylus end effector is used to convey this feedback.

Needles are also used in suturing for wound closure. The goal of one simulation [19] was to develop a realistic and economical haptics suturing simulator, refer back to Figure 2-3 (page 18). However, the Premium 1.5 device used in this simulation puts a high price on the required hardware. The simulation is displayed upon a stereo-enabled mirrored display and the force feedback device is mounted upside down modifying the range of motion of the stylus. This configuration modification can be used to increase the usable workspace for specific tasks. A mass spring model was implemented to represent the deformable skin tissue. This simulation is now intellectual property of Verefi Technologies Inc (Elizabethtown, PA, USA).

A device with 7 DOF positional capabilities and 4 force/torque DOF was developed by Hing *et al.* [102] to simulate a needle insertion. The forces and tissue deformation involved in a needle insertion and removal were collected from a porcine specimen that was then used to validate a finite element force feedback model. There is no visual feedback, and needle insertion and withdrawal velocity are unaccounted for. Work towards simulation of realistic tissue deformation is ongoing.

DiMaio developed algorithms to simulate visual and force output during needle insertion into deformable tissue [103] [104]. The model calculations are performed upon mesh nodes in the tissue model. If nodes are in contact with the needle, they're constrained. A rigid needle will constrain node movement along a single axis. Nodes may either stick to the needle or slip. As the physical behaviour of the needle within the tissue is not known, the model for sticking and slipping has been produced experimentally. The original haptics device used in the simulation [29] has been commercialised by Quansar, and is marketed as the Planar Pantograph Mechanism. It is a 3 DOF device allowing planar translation and unlimited rotation about a single axis. Quansar are also involved in providing a new force feedback device for Verres needle insertion [30]. The device has a syringe shaped end effector but the technical details are unpublished.

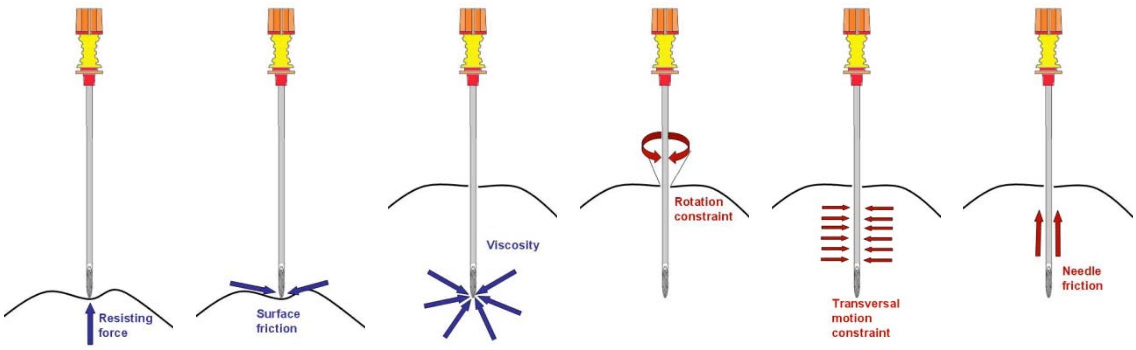


Figure 2-21 6DOF needle forces and torques as simulated by Faber *et al.* Images from[101]

2.8.3 Laparoscopy

Laparoscopic surgery, or minimally invasive surgery (MIS), is a surgical procedure performed through small incisions, using long thin tools to perform a procedure within the body. A surgeon's view of the procedure is occluded by the skin and as such a camera is inserted into the patient along with the tools. Tool manipulation is unintuitive as the surgeon has to move the tool handle right to move the tool tip left etc. The force/torque feedback is limited by the tight trocars through which the MIS tools enter the body. Orientation within the patient is also difficult to master and identifying the anatomy from a restricted camera angle is problematic. A practitioner needs training before performing an operation and several simulators are commercially available for this purpose.

There are more commercial simulators available for training in laparoscopy than for any other medical speciality. Procedicus MIST, a simulator sold by Mentice was one of the first on the market. It was originally developed and sold by Virtual Presence (London, UK) as the Minimally Invasive Skills Trainer. The current version uses the Xitact ITP and IHP hardware devices. The ITP performs only tracking whereas the IHP also provides axial force and pitch, yaw, and roll torque feedback. The simulation software is modular, allowing basic skills to be tuned using an abstraction of the real *in-vivo* task. For example, procedures such as suturing and knot tying are trained using a series of simple geometric shape manipulation tasks. Many validation studies have been performed on the product

Figure 2-22 Left: The Xitact IHP instrument tracking and haptic hardware for laparoscopy simulation. Right: Core skills training using Mentice's MIST simulator.

such as that from Taffinder *et al.* [105]. A complete list of studies can be found on Mentice's website [106].

The LapVR, formally owned by Immersion Medical (San Jose, USA) and now by CAE (Montreal, Canada), is a laparoscopic simulator that offers basic skills modules including the handling of geometric objects and also includes training in more realistic environments. Tasks include camera navigation (see Figure 2-23), cutting and procedural tasks of laparoscopic cholecystectomy. Each user's progress through the curriculum can be tracked for evaluation purposes. A custom haptics device originally developed by Immersion is used.

LAP Mentor, a product from Simbionix (Cleveland, OH, USA) originally used the Xitact LS500 [107] hardware (which combined a computer and monitor along with two laparoscopic interfaces and a camera tool). The latest version uses new haptics hardware from Mimic Technologies. Simbionix also market a lower cost, portable version of this system designed to be run with a laptop - the LAP Mentor Express. The device supports an expanding set of modules including training of knot tying, suturing and gastric bypass along with decision making and teamwork tasks.

The SurgicalSim Education Platform (SEP) is produced in a partnership between SimSurgery (Oslo, Norway) and Medical Education Technologies (Sarasota, FL, USA). Its basic skills module allows port placement, camera navigation, tissue

Figure 2-23 CAE's LapVR hardware, a commercial example of Laparoscopic simulation. Right: Lap Mentors camera manipulation training module.

manipulation and suturing exercises. In addition to these basic operations the simulation also includes gall bladder and embryo removal. Teamwork and decision-making are also trained. The interface to the simulator provides no haptic feedback and uses electromagnetic trackers embedded in the handles of specially built laparoscopy tools. The tool shafts are inserted into an elastomeric sheet that represents the skin access portal. The deformable tissue interaction software is licensed to other medical simulation companies.

LapSim (Surgical Science AB, Goteborg, Sweden), is a laparoscopic simulator available with a choice of either the Xitact IHP or ITP interfaces. The simulation has two laparoscopic tool interfaces and a single monitor. The standard basic skills module deals with procedures such as suturing. Various add-ons are available such as the gynaecological module. The simulation has been the focus of many validation studies [108] and links to these can be found on Surgical Sciences website.

Simendo (SIMulator for ENDOscopy) marketed by DeltaTech (Rotterdam, The Netherlands) provides a control interface without force feedback capabilities. Marketed as a simple simulator that does not try to tackle the complexities of a real procedure, DeltaTech recommend training on pigs for real world training. The simulation is designed to train the practitioner’s basic tool skills only, much like the Mentice MIST, where during tasks geometric shapes must be manipulated. The simulation does not require a high specification computer due to its simplicity and has undergone validation [109].

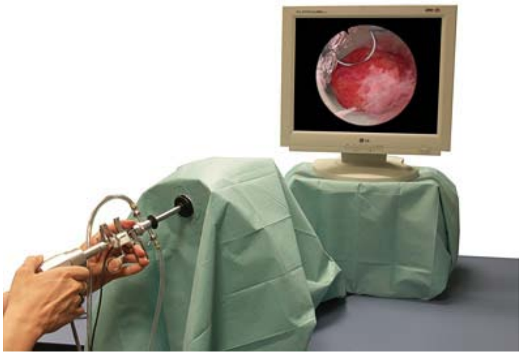
SkillSetPro (Verefi Technologies Inc.) is a laparoscopic training simulator combining camera navigation, suturing and basic skills training software modules. The user interface is Verifi’s custom hardware derived from a SensAble Omni, which is embedded into a mannequin’s torso. The simulator incorporates trainee feedback and performance measurements that are “easily recorded and viewed”. Teamwork and collaborative surgery can also be trained with the systems Head2Head module. Validation has been performed, but not as extensively as some of the other laparoscopic simulators [110].

The VSOne system (VEST System One) produced by Select IT VEST Systems (Breman, Germany) is a virtual endoscopic surgical trainer using a custom built haptics interface and the KISMET simulation software developed by Forschungszentrum Karlsruhe (Karlsruhe, Germany). The KISMET software supports real time interaction with deformable objects.

The VirtaMed (Zurich, Switzerland) HystSim is used for teaching hysteroscopy procedures and has been licensed by Simbionix. This system is the result of many years of work at ETH Zurich which entailed extensive attention to both physical behaviour and visual appearance [111]. The simulation has undergone face validation [112].

An academically produced laparoscopic adjustable gastric band simulator is the first produced for this procedure. The simulation, seen in Figure 2-24, uses two PHANTOM Omni devices to provide force feedback to the tools. Although not used here, SensAble produce a commercially available stylus modification for this type of simulation that can be attached to their Desktop device.

The Haptica ProMIS (Boston, MA, USA) system contains target simulations of a number of laparoscopic procedures that are used in combination with physical



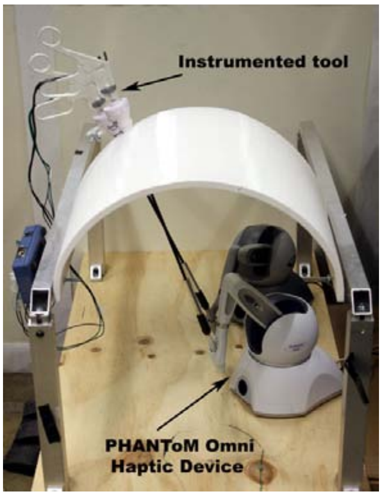


Figure 2-24 Left: HystSim from VirtaMed. Right: Laparoscopic adjustable gastric band simulator hardware [41]

models and digital video cameras to produce an augmented reality display. This system does not include a haptics device but rather uses the real interaction of surgical tools with physical surrogate anatomy to provide haptic cues.

For effective simulation, tools that penetrate a fixed point of a mannequin structure can be attached to devices providing 3 force DOF. If simulations were to include trocar insertion, a 6 force/torque DOF device would be required. The frictional forces exerted on the laparoscopy tools as they pass through a trocar must also be considered as these may well prevent more subtle haptic cues from being detected. No tactile feedback is present in laparoscopic surgery apart from at the tool/hand interface.

2.8.4 Endoscopy

A clinician feeding an endoscope into a patient will experience resistance between this flexible tool and the patient's body. There have been several examples where endoscopes have been used with haptics in a simulator to give an appropriate physiological response and accurate tool behaviour, e.g. [113] a bronchoscope force feedback device, VIRGY endoscopic [114], and [115]. Commercial products for endoscopy include GI and URO Mentor from Simbionix and Endoscopy AccuTouch from Immersion. Trifan and Stanciu [116] provide an up-to-date and comprehensive endoscopy simulation review. The decision of how many force DOF are needed for an endoscopy simulator is similar to that discussed in the previous section.

2.8.5 Endovascular Procedures

There are different disciplines of IR: one focusing on the peripheral vascular system; procedures focusing on the brain, possibly as therapy for strokes called interventional neuro-radiology, and those procedures focusing on the heart, interventional cardiology.

Many endovascular procedures start with a needle insertion into the vascular system, but current commercial simulators skip this step to reduce the simulation complexity and hardware cost. A guidewire and catheter are then manipulated within the vascular anatomy to navigate to the position of interest. This is a 2D visual (using fluoroscopic guidance) and tactile process, sensing small axial forces and torques at the fingertips whilst manipulating the wire. The acute training of wire guidance and the response to the fine forces felt whilst advancing a wire is crucial for efficient IR procedure training. An over exertion of force can have serious consequences and correct training to prevent this must be included in any IR wire guidance training simulation.

Two early IR simulators were the Dawson-Kaufman IR simulator, designed by HT Medical for practicing angioplasty [117] and the daVinci/ICard IR simulator [118] [119].

VIST (Mentice AB, Sweden), see Figure 2-25, is sold as a simulator for various endovascular procedures. VIST was developed from an interventional cardiology

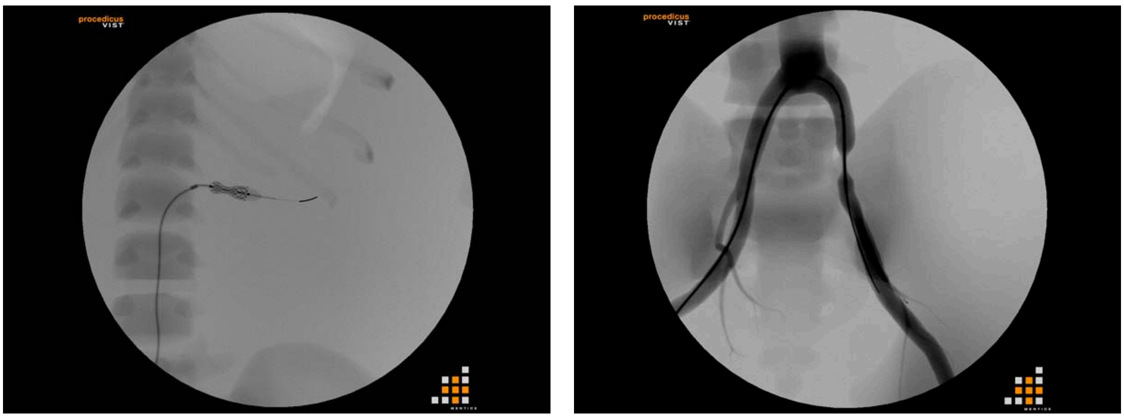


Figure 2-25 Visualisations from Mentice's VIST. Left: Stent placement. Right: Fluoroscopic dye highlights the femoral arteries.

simulation called ICTS (Interventional Cardiology Training System) [120]. The development of ICTS started at MERL, the Mitsubishi Electric Research Lab (Cambridge, MA, USA) in collaboration with the CIMIT group [121] and a number of interventional radiologists. This work was subsequently brought to completion by Virtual Presence (Sale, UK) under contract to Guidant (Brussels, Belgium) and ultimately commercialised by Mentice. A more recent simulator using a hydraulic pulse generator for palpation and an adapted Vascular Surgical Platform (VSP) haptics device from Mentice for catheter and guidewire manipulation is being developed by the CRaIVE consortium in the UK [122], see Figure 2-26. This simulator is aimed at training the Seldinger Technique for catheter insertion, which covers the initial steps of introducing a guidewire and catheter into the patient. A construct validation study is currently in progress.

Anderson [118] extended work from the daVinci simulation towards an IR simulator for cerebral vascular, peripheral vascular and cardiac applications in collaboration with Kent Ridge Digital Laboratory in Singapore and the Johns Hopkins Medical Institution [123]. An earlier paper [124] describes NeuroCath, the cerebral vascular track of the simulator. The latest developments of NeuroCath are given in [125]. The system interface is a mannequin structure and the simulator's focus is guidewire manipulation. Currently, real cardio vascular IR instruments

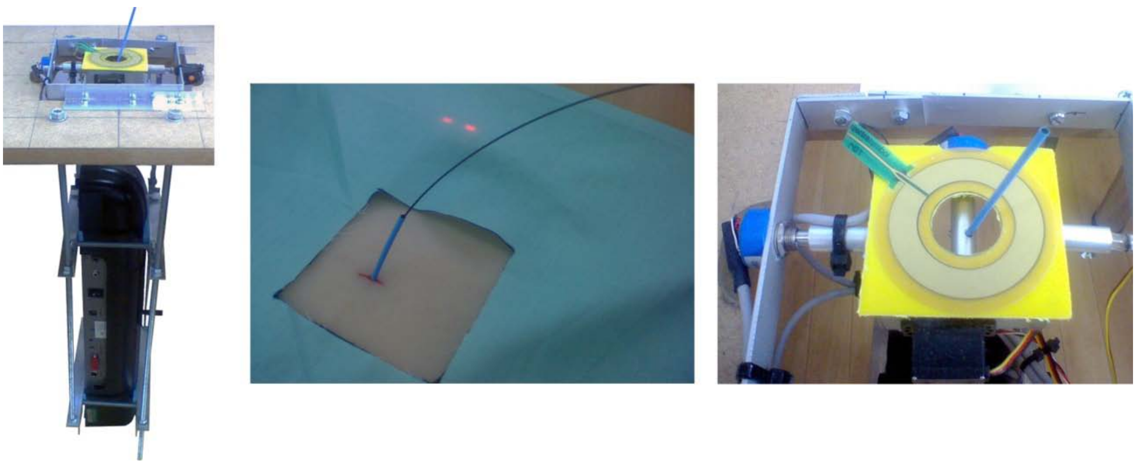


Figure 2-26 Custom haptics based needle holder from Bangor University. A needle can be held at a chosen orientation through which a guidewire and catheter are fed into a Mentice VSP haptics device (black lozenge shaped box). Fake skin covers a yellow disk shaped pressure pad which senses the users finger position. Commercial off the shelf haptic devices cannot be used for guidewire simulation as specialised hardware is required to provide force and track the wire/catheter. Such devices typically include optical motion sensors combined with force feedback mechanisms to allow a guidewire and catheter to be used simultaneously whilst monitoring depth of insertion and applying forces to each tool as appropriate.

can be inserted into the guidewire interface, which tracks and provides force feedback to the user in conjunction with visual feedback. A vascular model and potential field catheter navigation method for the simulation is discussed in [126].

Modelling the response to a guidewire as it is manipulated within the vascular system is a complex research topic as both structures are deformable. Alderliesten *et al.* [127] test the reliability of their catheter simulation by comparing the simulated results to those of real wire manipulation in a phantom model. The reproducibility of guidewire propagation was also assessed. A straight and curved tip wire was modelled with a series of rigid segments. The static friction of the wire against the side of the vascular system (which had been previously ignored) was also considered [128]. The latest simulator from the SIM [129] group, EVE, is a neuroradiology training simulation [130]. Some features of the simulator include interactive fluid dynamics of blood flow [131], volumetric contrast agent propagation and real-time collision detection and collision response [132]. Current efforts are aimed toward integrating performance assessment and user guidance.

Other simulations of interventional procedures under fluoroscopic guidance include Simbionix's ANGIO mentor for interventional endovascular procedures. This includes two smaller portable versions of the product: the Mentor Mini and Express which can be run on a laptop and use the compact Mentice-Xitact endovascular interface device. Mentice also sell a compact version of VIST which uses this same interface. The CathLabVR, simulator from Immersion, uses custom haptics hardware.

The HERMES project [133] created a training system for coronary stent implants. The system uses a custom haptics device [134] and a finite element method for soft tissue modelling of the artery [135]. The CathI (Catheter Insertion System) [136] simulation adopts a mannequin approach. The mannequin is laid on an operating table to provide a simulation environment as close as possible to that of a real procedure. SimSuite (Denver, CO, USA) also produce a commercial training simulation.

2.8.6 Arthroscopy

A knee (or shoulder) arthroscopy procedure requires a small camera and specialised instruments to be inserted into the knee. In this procedure, the practitioner's tools will interact with both hard knee bone and the soft surrounding tissue. Simulating realistic hard contacts is a significant problem [17], especially when combined with the need to simulate soft contacts next to the hard objects.

Arthroscopy simulators include the commercial product insightArthroVR, created and manufactured by GMV (Madrid, Spain), see Figure 2-27. The device uses an LCD monitor for visual feedback combined with two SensAble Omni devices with modified end effectors. The tips of the end effectors are manipulated within a knee or shoulder mannequin depending upon the procedure performed; giving the simulations good face validity. The simulator has undergone a recent validation study of face and content validity using a questionnaire style evaluation, and construct validity judged with a time metric on a small subject number [137]. A larger subject group needs to be studied along with a longer term transfer of skills study to draw any concrete validation conclusions.

Mentice sold an arthroscopic simulator called the Procedicus VA that originated from work at Prosolvia. It was acquired following the dissolution of Prosolvia and the subsequent creation of Mentice where the product was refined and commercialised. This simulator, which saw extensive use and publication as the first commercial arthroscopy simulator, is now being re-engineered for future re-introduction.

Another commercial simulator is available from Touch of Life Technologies (ToLTech) (Aurora, CO, USA), see Figure 2-27. This simulator has undergone extensive development under sponsorship and guidance of the American Academy of Orthopedic Surgeons. Two separate monitors are used, one for the virtual mentor and the second displaying the procedure. Force feedback is provided by two SensAble PHANTOM Desktop devices with modified end effectors. SensAble's force feedback device's have also been used in academic arthroscopic simulations [138], [139]. Heng *et al.* [140] provide illustrations showing that an off the shelf PHANTOM Desktop can't be directly used in their simulation as its 3 force DOF are not the correct degrees for the simulation. They have developed their own 4 DOF device which offers three degrees of force feedback.

Force feedback hardware has also been incorporated into a knee mannequin. Examples include KATS [142] which has undergone validation studies [143], OrthoForce [141], and the early work at MERL that used voxel-based haptic simulation approaches [138] coupled with a powered gimbal linked to a SensAble PHANTOM [144]. As humans can distinguish between the high frequency vibrations that occur when two objects come into contact [145], Tenzer *et al.* [17] have tried to recreate the vibrations felt during the arthroscopy procedure to enhance the tactile fidelity of the OrthoForce device. Although the device has only been tested on a small group, the results appear to be positive. Commercial haptics devices can be used for arthroscopy simulation if heavily modified. Frequently custom devices are used. The tactile information involved whilst palpating the knee without tools has not been simulated.

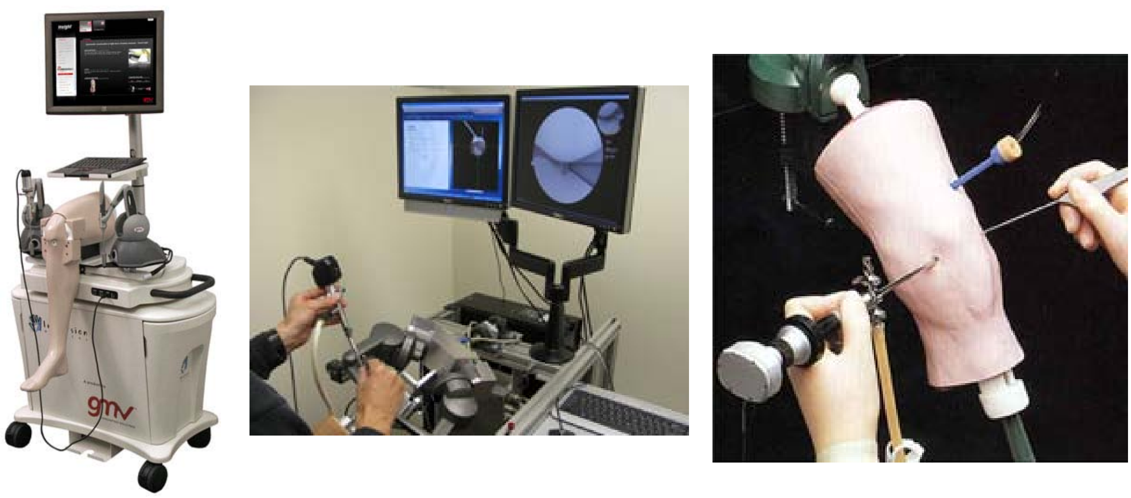


Figure 2-27 Left: insightArthroVR from GMV uses two PHANTOM Omni devices to provide force feedback. Centre: ToLTech arthroscopy simulator using two PHANTOM Desktop devices. Right: OrthoForce [141] using custom force hardware and including vibration feedback.

2.9 Augmented reality

Augmented reality describes a cross modal interaction where interactions with objects in the real world influence interactions with objects in a virtual world, combined in a single visualisation, be it visual or haptic. Commonly, this takes the form of visual collocation where virtual objects are overlaid upon a real world scene such that the virtual objects appear to be part of the real world view; substituting, highlighting or modifying real world objects, or providing additional information about the real world.

2.9.1 In Surgical Applications

The shift from open surgery, where layers of a patients healthy tissue are cut away to reach an area of interest, to MIS in which the use of modern imaging techniques enables a practitioner to guide tools to the point of interest with minimal damage to surrounding tissues, causes the surgeon visualisation problem as direct line of sight during the intervention is lost. The main focus of augmented reality in medical applications has been to restore this direct line of sight by moving the medical images viewed on 2D visualisation system beside the patient to co-locate the visualisation “in situ”. This in situ visualisation typically allows the practitioner to see through a patient’s skin and tissue as if the layers on top were transparent or had been cut away like in an open procedure. Successful

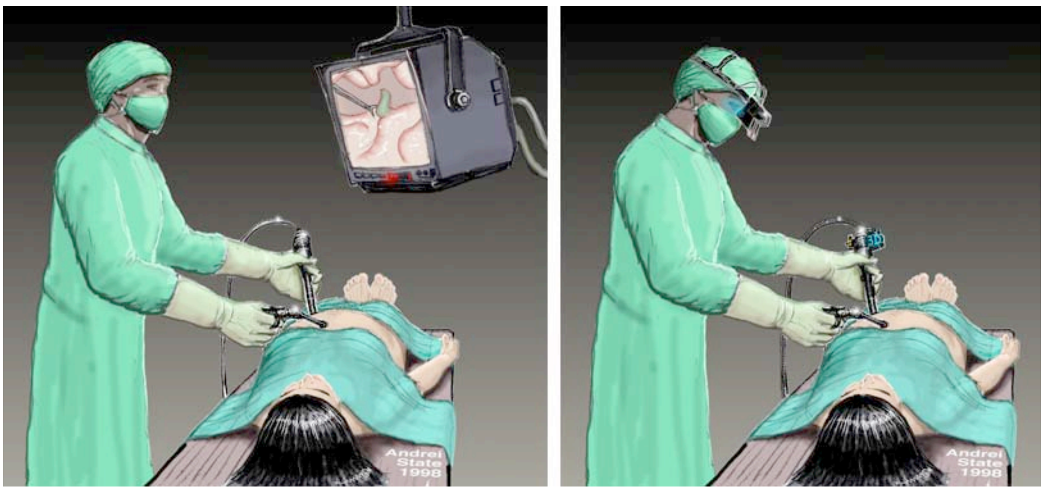


Figure 2-28 Artists impression of regular and augmented reality visualisation for laparoscopic surgery. Taken from Fuchs *et al.* [146]

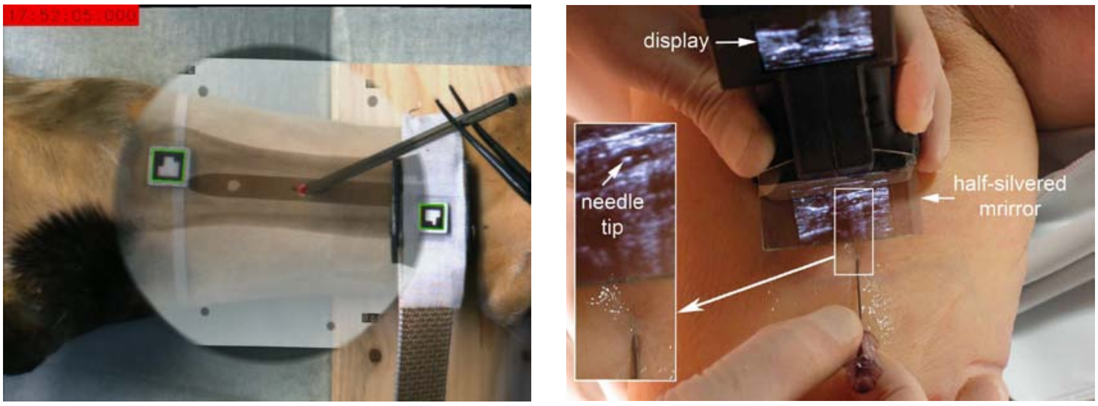


Figure 2-30 Left: Augmented reality guidance of intramedullary nailing displayed on an LCD monitor [151]. Right: The Sonic Flashlight from Insituvue (Pittsburgh, USA). Operator's point of view as the device is held in one hand as the needle in the other is guided with the aid of the reflected ultrasound image of the vein. Image taken from [152].

application of this technology would allow MIS tools to be guided more intuitively to their target, considerably reducing the cognitive load a surgeon faces whilst performing a procedure. Developing this idea, a HMD system for augmented reality ultrasound was proposed in 1992 [147], an approach which was later used in ultrasound guided needle biopsy [148]. Fuchs *et al.* then went on to augmented reality guided laparoscopic research in 1998 [146], see Figure 2-28. There are various other applications of in situ visualisation using a HMD, for example, to assist port placement in laparoscopic surgery [149].

Not all approaches use a HMD for AR visualisation, a standard LCD display is used to overlay x-ray images over a real world video stream for needle guidance [153]

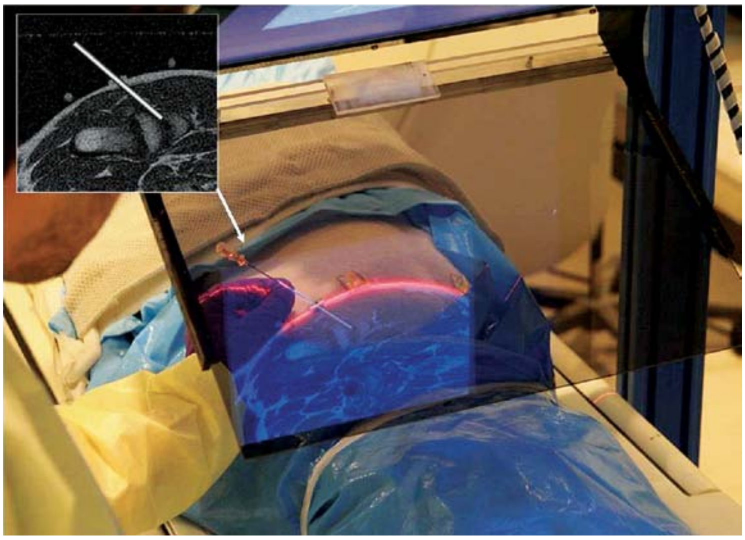


Figure 2-29 Close up view of the MR image overlay system in a porcine trials to guide needle insertions into the joint space of the shoulder. The plan on the targeting image is shown in the inlay [150].

and for intramedullary nailing [151], see Figure 2-30. A small portable application of the immersive mirror display technology currently undergoing commercialisation by Insituvue (Pittsburgh, USA), is used to aid ultrasound guided needle punctures, overlaying a real time ultrasound image directly over the patient's skin [154], see Figure 2-30. On a larger scale, an MRI image guided needle insertion is an example of the use of a full size immersive mirror display for in situ medical visualisation [150], see Figure 2-29. A full review of this field is out of the scope of this thesis.

2.9.2 In Medical Training

Augmented reality (AR) has not been widely used in medical training applications, although visual AR anatomy training applications have been produced. One such example is BARETA [155], a training application using handheld virtual anatomy and a clipping plane to interactively explore anatomy, see Figure 2-31. A procedural training application using augmented reality has been designed to train ultrasound guided needle placement [156]. Developed at Leeds University, this simulation tracks a mock ultrasound probe as it is passed over a foam mannequin body and needle as it punctures the foam structure. An ultrasound image is then generated with respect to the positions of these two tracked objects in relation to the mannequin. The virtual nature of the ultrasound images, generated from CT scans, can simulate the different internal structures of many patients, although the external shape of the model will not change as the simulated patients habitus does, and anatomy specific haptic feedback is not felt as the needle is inserted.

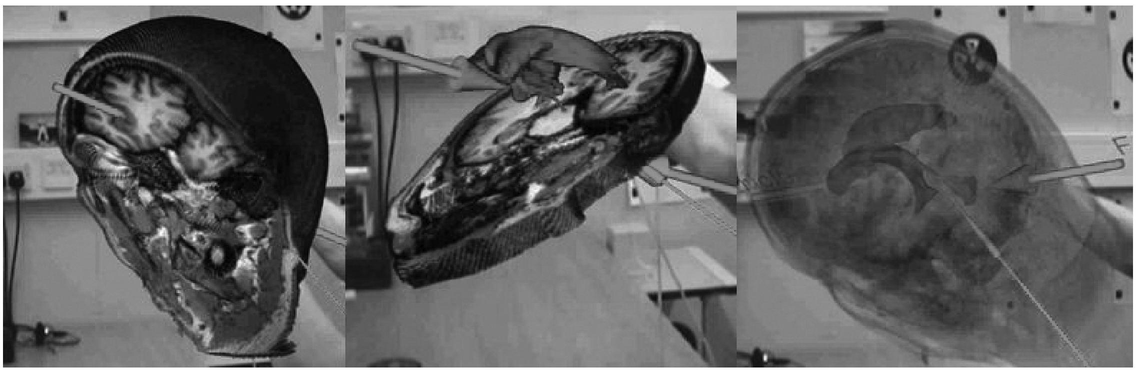


Figure 2-31 BARETA, Bangor universities augmented reality education tool for teaching anatomy.

A feasibility study researching the use of augmented reality and haptics for the simulation of bone fracture repositioning [157] was conducted in candidature for a PhD at ETH Zurich. The study, which used a single helmet mounted camera and HMD for visual feedback in combination with a PHANTOM Premium 1.5 device, did not produce a medical virtual environment. However, algorithms to reduce visio-haptic calibration errors which occurred due to HMD tracking complications and system lag were developed. Two non-medical test applications were produced, one a haptic ping-pong game and the other allowing a user to manipulate a deformable virtual cylinder with the haptic stylus's end effector, see Figure 2-32. The simulations suffered from a visual parallax discrepancy as the camera was mounted on the helmet rather than in front of the user's eyes, which reduced the simulations immersion into the AR environment. It also suffered from low quality AR images, system lag and occlusion problems. To further understand the problem of system delay, a paper by the group [158] investigates the influence of visual and haptic delays on stiffness perception and uses a new grounded HMD display to support the weight of the higher quality device. This visio-haptic system appears to have a large computational overhead, requiring 4 separate computers to run the simulation effectively, each dedicated to either tracking, haptics, graphics or simulation.

Although augmented reality is commonly associated with visual augmentation of the real world, a recent publication has presented work on haptic augmentation of

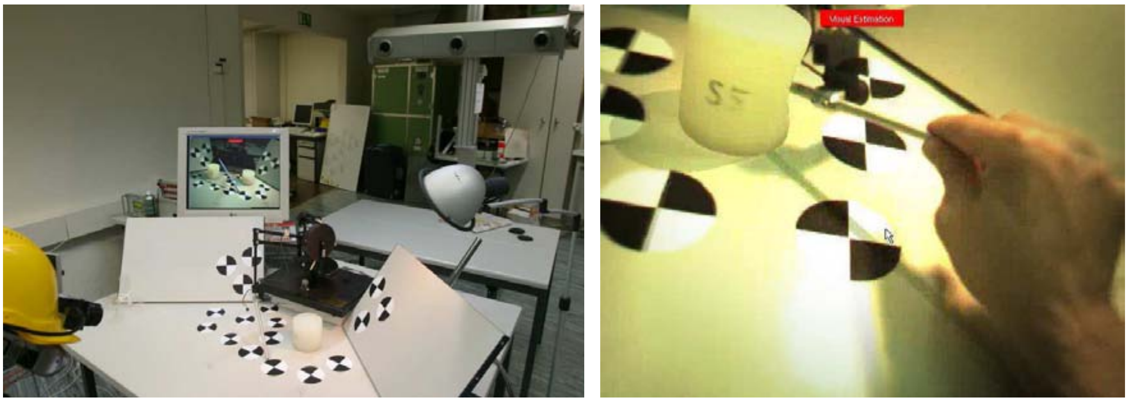


Figure 2-32 HMD augmented reality. Left: The major components of the system (Optotrak in background, the PHANTOM, the lamp, the landmarks placed on three different planes to cover the user's field of view involved in the system) Right: The stylus interface used to manipulate virtual objects. Images taken from [157].

real world deformable objects [159]. Researchers from Pohang University, Korea have fitted a PHANTOM Premium 1.5 haptic end effector with a force sensor. As the user touches a real world object with the sensor, the haptic device is used to provide an additional force feedback over that felt from the object to modulate how stiff the real object appears. This modulation of the real surface could allow for virtual tumours to be simulated within real objects, although the pen interface currently used may not offer a realistic palpation interface.

2.10 Simulation Evaluation and Validation

The fundamental perceptual issues whilst performing MIS and other procedures are not fully understood. Evaluating simulations of such procedures is therefore non-trivial.

Simulations can be evaluated through assessment of the benefit they provide, this can be performed by assessing the validity of the simulation as defined by Morthy *et al.* [160].

- **Construct validity** is the extent to which a test measures the trait that it purports to measure. One inference of construct validity is the extent to which a test discriminates between various levels of expertise.
- **Content validity** is the extent to which the domain that is being measured is measured by the assessment tool—for example, while trying to assess technical skills we may actually be testing knowledge.
- **Concurrent validity** is the extent to which the results of the assessment tool correlate with the gold standard for that domain.
- **Face validity** is the extent to which the examination resembles real life situations.
- **Predictive validity** is the ability of the examination to predict future performance.

Formal validation studies that focus on the use of haptics in medical simulation are scarce, with most simulations only assessing the simulation's face validity as further validation is extremely time consuming and its long term effects are difficult to prove. The evidence presented in this review suggests that surgical simulations incorporating haptic feedback provide a richer training experience than those that do not. The most complete validation studies to date have been performed upon the available Laparoscopic simulators (e.g. [161] and [162]). One study supports the use of force feedback during a tissue characterisation task in a MIS setting [163]. It concluded, “subjects are more comfortable characterizing tissues when both vision and force feedback were provided”. Another study conducted using Immersion’s (now CAE’s) LapVR suggests that for more advanced Laparoscopic tasks, the addition of force feedback in a simulation results in faster task completion [164]. However, a recent review of haptic feedback in conventional and robot-assisted laparoscopic surgery [165] concluded that there is no firm consensus on the importance of haptic feedback in laparoscopic simulators.

Often skills transferability is reported by demonstrating an improvement in the tasks performed by trainees who use the simulator over those who do not [166]. One evaluation of an endoscopic sinus surgery simulator [167] argues that a significant difference in performance between experts and novices demonstrates a similarity to real world performance. It also advocates rating and comparing anonymous videos of procedures performed by simulation trainees and a control group. A complete evaluation of transfer of skills where one control group does not use a simulator, another uses the simulator without haptic feedback and the third uses the complete haptic simulation has yet to be carried out. This has partially been addressed by Morris *et al.* [168] who demonstrated that recall following visio-haptic training is significantly more accurate than recall following visual or haptic training alone, although haptic training alone is inferior to visual training alone. However, whether the latter would be true for an interventional radiologist, where reacting to haptic cues is a vital part of a successful procedure, has not been investigated.

Another trend is to use data acquired from empirical *in vivo* force measurements [169] rather than using a purely mathematical model. The measured forces can then also be used to compare simulation output against real world data and reduce reliance on the validation of a simulator's fidelity and accuracy by a subjective “it feels right” approach. Such a study has not yet been reported.

A good overview of the issues that need to be considered when assessing a surgical simulator has been outlined by Gallagher and Ritter [170]. The need for multidisciplinary collaboration to build an effective simulator is advocated. Other important points made are that it is advantageous to deconstruct tasks into simple steps, to have repeatability of procedures which will facilitate learning from mistakes, to provide objective feedback, and that it is necessary to integrate simulators into the education curriculum. There is often a trade-off between the fidelity of the simulation and its cost, and it is not always necessary to achieve ultra high fidelity in order to provide a training benefit. The current published evidence clearly demonstrates that VR simulation can improve intraoperative performance. The work surveyed demonstrates that a good use of haptics has an important role to play in achieving this goal.

2.11 Validation through simulation

Assessment is an important part of the training curricula to monitor trainee's progress and to identify the point at which trainee's are ready to progress to the next stage and perform *in vivo* procedures. Gallager *et al.* [171] provide a hypothetical model of attention for both the novice and expert surgeon, proposing that the attention required by a trainee whilst learning a procedure will exceed a humans cognitive abilities, see Figure 2-33. Through pre-procedural training it is thought that this load can be reduced below the trainee's cognitive limit at which point, if their skills are adequate, training could proceed under the apprenticeship model *in vivo*.

The ability of a simulator to identify the competency of a user is the simulators "construct validity". If it is not possible to identify the difference in performance between a first time user and expert then the simulators validity should be questioned. A common approach to measure competency in simulators is to measure time against error rate. Error rate is of higher importance in a surgical context than the time taken to complete the simulation (within reason). An expert should perform a procedure without error in a comparatively short time, whilst a trainee will make errors (of which they should be notified) and if they do complete a procedure without error, they will mostly likely take longer than an

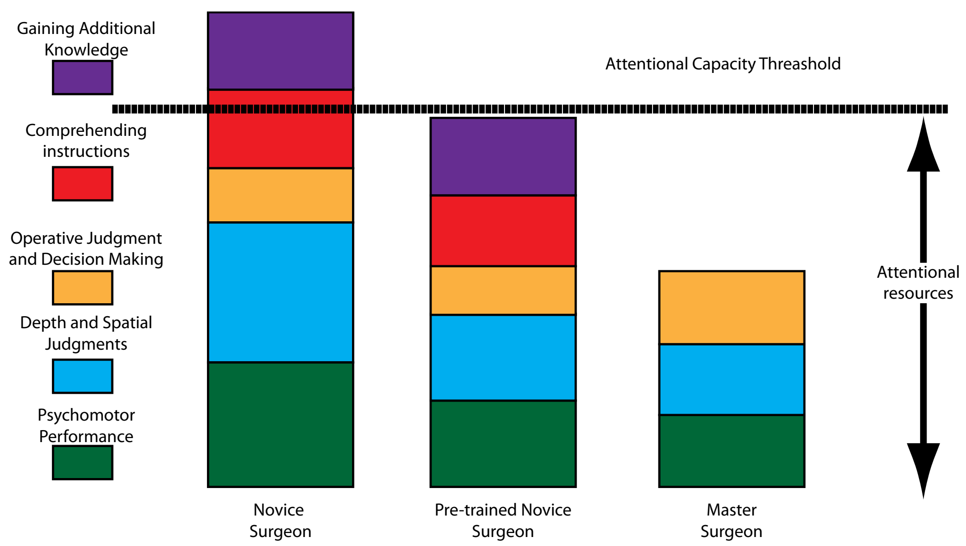


Figure 2-33 A reproduction of the the hypothetical model of a surgeons cognitive load before, during and after training as proposed by Gallager *et al.* [171]

expert. Additional metrics that have been used for laparoscopic simulators include tracking the motion of a tool's tip to record the smoothness of motion and path length [172] and occasionally the tracking of a users limbs [173]. An expert's tool tip will generally travel a shorter distance to perform a certain procedure or deviate less from a central focus point. Their limb movements will also be more rapid and controlled. The advantages of motion metrics over time / error rate are questionable [174] and a more in depth analysis is usually required. A different approach for surgical performance analysis looks at a user's tonic accommodation (a stable parameter that is adopted by the eye) in a laparoscopic procedure [175]. It is thought that a surgeon's performance can be affected by their tonic accommodation value where surgeons must perform procedures looking at a monitor upon which there is a visual representation of the operating field. This could have implications for VR simulation as a virtual operating environment is projected upon a monitor at a different focal distance to the objects in a real world environment. A better understanding of these problems will inform the development of approaches that are optimally suited to the properties of the human perceptual system. Examining visuo-motor performance within the framework of sensory integration is a new area of research that may provide a powerful way to evaluate surgical systems and simulations in the future.

Currently there is no standard to evaluate surgical performance, but as these are introduced, surgical accreditation through simulation is likely to become a part of everyday training. In much the same way a pilot is not allowed to fly commercial airliners if they have not undergone regular simulation training and accreditation, a surgeon will not be allowed to perform surgical procedures until they have performed them virtually. Interestingly, if a simulation has predictive validity, a surgeon's fundamental abilities can be monitored and theoretically a potential trainee's ability to learn could be quickly analysed, leading to an early selection of promising candidates from a pool of potential surgeons.

2.12 Choosing the Optimum Simulation Medium

This review has shown no simulation hardware stands out as the optimum haptic feedback device. This choice is very task independent. Likewise the choice of simulation medium, be it mannequin, semi, or fully virtual is task dependent. Applying technology where it is not needed may increase cost and inadvertently lower training fidelity. Through careful consideration of cost, reusability and a simulation medium's ability to create suspension of disbelief, three guidelines outlining where a simulation medium is best used have been devised:

- If the simulated patient is to be touched, but no surgical interventions (e.g. needle insertion) are to be performed and there is no variability in patient habitus and anatomy, a mannequin can be used.

- If the simulated patient's habitus is to be kept constant, the surgical tool entry positions are to be pre-defined and visual feedback is via a standard 2D monitor, a mannequin housing for virtual haptic tools can be used in combination with a separate display.

- If variable patient habitus is to be simulated and/or a tool entry position may be arbitrarily chosen, a full virtual reality approach must be used. Augmented reality, as presented in Chapter 7 of this thesis, should be used to provide high fidelity visual feedback.

A mannequin based simulation can be satisfactory only if no penetration of the mannequin's skin is required. In simulations where the skin is punctured, leaving markings, the skin must be replaced after each simulation to retain the simulators face validity. Development of self sealing materials can offer some improvement to this approach. Changing the habitus of the simulated patient and the anatomy beneath the skin is not possible without buying new parts and manually interchanging them. Cost will increase rapidly as the number of anatomical variations increases.

If the habitus of the patient is to be kept constant but interventions must be performed within the patient, assuming the tool sites are pre-defined (training intervention site location cannot be trained in this situation), a mannequin based tool housing can be used. The virtual patient anatomy can change within the statically simulated body habitus. The visual cues in such procedures (e.g. laparoscopy and IR) can be displayed upon an LCD monitor at the side of a patient as occurs in practice. Current implementations using this approach are part procedure simulations.

If the patient habitus is variable, it is not possible to use a mannequin for simulation visualisation due to its static form. In this situation, virtual simulation excels. A virtual environment can be used to project a virtual representation of a patient that can have any habitus or anatomy. If the patient's habitus is static but the simulation must accommodate interventions, for example, a needle puncture or cut, the simulation must be virtual if it is to be cost effective. As previously stated, replacing a mannequin after each simulation is expensive and as inefficient as training upon a cadaver. One mistake can render a cadaver useless to re-demonstrate a procedure. Conversely, a virtual model which is punctured, cut and deformed can be reloaded at no cost at all and the user is free to practice the procedure as many times as necessary to achieve a reliable outcome. The forces felt during the simulation can be conveyed through haptic force feedback devices held within the user's hand in the case of tool manipulation or through strategically placed haptic end effectors which can be used to restrict a user's hand as it comes into contact with the virtual patient in the case of palpation. In addition to these advantages computer based simulations can compute quantitative metrics of performance. Practitioner assessment through computer based simulation is discussed in section 2.11.

2.13 Discussion and Conclusions

The particular requirements for haptics within a surgical simulator varies with the application, but the common trends and issues identified above can be summarised as follows:

- How to use haptics and still have an affordable simulator? The cost of multi-purpose force feedback devices has greatly reduced, but custom devices often needed by surgical simulators are still expensive.
- The availability and development of tactile interfaces is still in its infancy.
- There are always technology questions to consider, with real time response essential.
    - How many force DOF are needed? Is a device offering 3 force DOF sufficient as devices providing 6 force DOF or more are expensive?
    - What computational power is needed? Is a dedicated processor required for the haptics pipeline?
    - Is the haptic hardware's force/torque range sufficient? Will the range cover the whole pathology and patient variability that the simulator will encounter?
    - Is the precision of the haptic hardware high enough? What resolution is necessary during the procedure?
- Multipurpose haptics devices are by far the most commonly employed, but do they compromise the fidelity of the simulation particularly when compared to custom built haptics devices? However, software support for multipurpose devices is good with several haptics libraries now available. In many cases new and novel algorithms are also being implemented to improve performance and fidelity of simulation.

- What is the objective of the simulation? Clinical skills or tool training? A higher fidelity is typically needed for the latter. In both cases, a more successful simulation is provided if a detailed task analysis has taken place.

- There is a marked lack of validation studies that can report on the benefit (or otherwise) of using haptics in a surgical simulator. The question of appropriate simulator metrics for the use of haptics remains open.

Inevitably, compromises are made when incorporating haptics into a medical simulator but, nevertheless, the technology has a growing and important role to play in the medical domain. In particular, this has been the case for minimally invasive procedures and also when a tool such as a needle or surgical drill needs to be simulated. Open surgery and procedures where practitioners must grasp soft tissues directly with their hands remains a research challenge.

In a 2008 publication by economist R.M. Scheffler, the cost of training a new physician is estimated to be $1 million (USD) [176], approximately £540,000 GBP (as of the publication date, Sept 2008). The true advantage of providing an effective training simulation is hard to assess. In monetary terms, effective simulation can reduce the time wasted in extended procedures due to inexperienced practitioners practicing during valuable operating time; the guidance needed from expensive experienced practitioners; the medical errors (costly by requiring corrective procedures and through compensation claims) and the need of expensive cadavers. On an ethical level, a simulation that prevents medical errors that would have resulted in a patient death or disability should make the simulator indispensible, but in the real world of business this is not an adequate enough justification unless it is possible to prove to providers and insurers that a simulator will reduce risk and thus save money. Unfortunately, a direct linkage between simulator training and improved patient outcomes is difficult, if not near impossible to prove [177].

Procedure variability between practitioners in a single department, as well as between different hospitals will cause conflicting requirements. A procedure carried out by an expert should not be deemed incorrect because they don't perform the procedure in the same way as the control experts. To incorporate this variability into any measurement and assessment approach, it may be advisable to limit definitions of incorrect methods of performing a procedure (for example, defining an area not to be penetrated by the needle) to the locality of specific anatomical structures. Such measures would be turned off when a simulation is used to allow the surgeon to try high risk manoeuvres in a safe environment and discover new techniques. Any limitations of a simulator should be made clear to avoid incorrect training that could ingrain bad habits (negative training). An extreme example of this could be a simulation that allows the removal of the virtual patient's heart without killing them. For such an extreme example, it is clear that the practitioner would not believe the simulator is correct and perform this operation in real life. However, more subtle simulator inaccuracies may be more difficult to distinguish as unrealistic. Although many simulations aim to recreate realistic representations of anatomy and physiology to develop skills that can be transferred to the patient [178], this may not be the most effective training method available. Validation studies of the MIST task trainer prove that manipulating simple geometric objects, i.e. not anatomically realistic, is a very effective tool for training basic MIS hand/eye coordination tool skills [179].

The ability to record metrics in a simulation offers advantages for trainee evaluation. A whole array of data is available to the programmer to process for evaluation and therefore needs to be carefully thought out. The metrics to be recorded should be defined in a task analysis preceding the simulator development and not as an afterthought. If the data recorded is the correct information and it is accurately correlated with expert judgement of performance, the possibility to use simulation for accreditation exists [178].

The availability of effective simulators will ultimately be defined by cost. The cost of force/torque feedback devices is slowly decreasing, but the volume of sales needed to reduce cost significantly is currently beyond the scope of medical simulation. SensAble Technologies currently make the most popular choice of force feedback interfaces for medical applications. Their PHANTOM Premium devices are chosen for their high quality response and their Omni device for its low-cost 6 DOF tracking and 3 force DOF capabilities. The games market has driven innovation in graphics cards and the GPU is now a powerful computation tool. The introduction of Novint's Falcon device suggests that force feedback may also be adopted by this market. This may then lead to more low cost force feedback interfaces being available for simulation development.

Low simulation cost is usually high on the list of requirements during development of a commercial simulation product. Haptics hardware can add significantly to simulation cost in a visio-haptic simulation. However, cost cutting on hardware in the early stages in an effort to save money may obstruct the production of a simulation with sufficiently high fidelity. Analysis of a high quality simulation can determine if the extra cost is worth the increase in fidelity, and so producing a quality prototype simulation, validating this, and then reducing the fidelity to meet cost requirements, whilst maintaining transfer of training effectiveness may prove to be a better approach. For example, the low cost Falcon device may well be sufficient for many tasks currently using PHANTOM devices (e.g. needle puncture). Evidence of this trend occurring has been seen, although such information is difficult to obtain from companies who do not want to reveal their intellectual property.

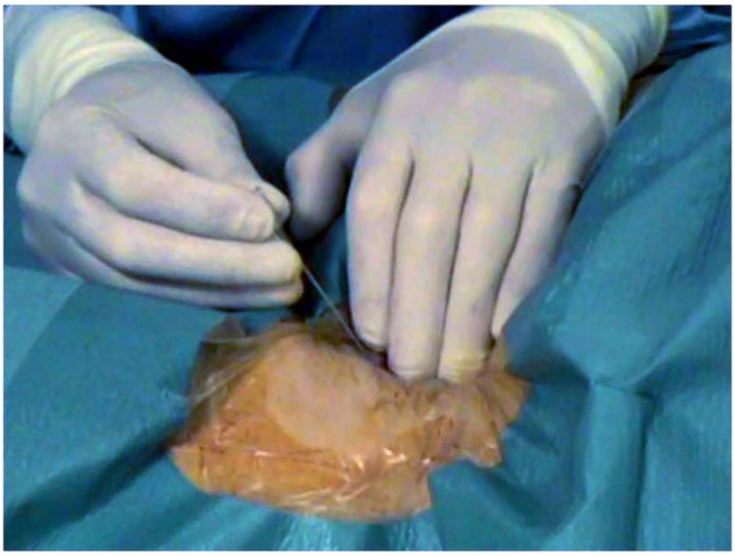
Although many visual displays exist that can be used to effectively display virtual environments in both two and three dimensions, the number of displays capable of combining the displayed visual cues in collocation with the haptic feedback are limited. The most common method for collocated visual and haptic force feedback is a semi transparent mirrored display, even though the view is suboptimal [48]. The use of a video see through augmented reality display for medical simulation has been proposed, but not yet realised as the hardware is commonly bulky, provides only a low resolution visualisation and requires complex tracking. As such, of the reviewed display technologies, the semi transparent mirrored display does appear to provide the optimal trade off between collocation functionality, hassle free usability and cost.

The face validation of simulations typically include a low number of participants due to the complexity of finding a large number of experts in the simulation field. This produces subjective feedback and the questions asked in these studies often vary. As such, comparing simulations is difficult. Comparing device hardware for the use in these simulations is also extremely difficult as each product offers different features. For example, a basic real world needle insertion is a 6 force DOF operation. It could be assumed that nothing less than a device displaying this many DOF could be used to produce a realistic simulation. However, realism comes at a cost and a near realistic affordable simulation may be better than no simulation at all. It is not immediately obvious if the education value of a low cost simulation offering only 3 force DOF is any better or worse than a higher cost solution offering the full 6 force DOF feedback. No study has yet been carried out to show whether this is indeed the case.

Emerging technologies will continue to offer the potential of creating higher fidelity simulations, but should only be used where a clear training benefit can be proven. This survey indicates that haptics technologies have reached this stage and will have a pivotal role to play in the ability to maintain skills competence and reduce the need to train on patients.

Chapter 3 extends the findings of this review to investigate the need for simulation in interventional radiology. This is the clinical area that we have collaborated with, and an exemplar application in which the hypothesis of this thesis can be explored is developed under guidance from interventional radiology experts.

3 Interventional Radiology



3.1 The Procedure

Interventional radiology (IR) is a minimally invasive technique pioneered by Charles Dotter, a radiologist at the University of Oregon in Portland, USA. The first IR procedure was performed in 1964 [180], and is now frequently used to carry out tasks such as unblocking or blocking vessels, biopsies, draining abscesses and infusing drugs to specific sites in the vascular system. An IR procedure often starts with the Seldinger technique [181], an endovascular procedure used to gain access to the vascular system by feeding a wire through a needle which has been inserted into an artery. A full description of the vascular access process is given in section 3.2. After this, a wire and catheter are navigated along the vessel under 2D x-ray guidance (fluoroscopy), used to visualise the position of the high contrast tools. A contrast medium can now be injected through the catheter to aid catheter navigation, diagnostic purposes and interventional planning. Embolic agents can also be injected through the catheter lumen for, e.g., devascularisation of tumours. Re-insertion of the wire allows an exchange for sophisticated therapeutic catheters such as balloons, stents or grafts.

Figure 3-1 An interventional radiology procedure. The patient lies underneath the x-ray machine whilst the practitioner wearing a lead apron performs the procedure. Red: C-Arm x-ray machine. Yellow: 2D visualisation of live X-rays. Green: Patient under x-ray machine. Thanks to Dr Steven Powell of the Royal Liverpool NHS Trust

The numbers of straightforward invasive diagnostic IR procedures, which are the main source of training opportunities, have greatly reduced in frequency as non-invasive imaging techniques have improved. This, combined with a need to cut costs and training times, has made IR a prime candidate for the use of alternative training methods.

Currently, only the wire and catheter manipulation stage of the IR procedure has been fully virtually simulated. Examples of such solutions are Mentice's (Gothenburg, Sweden) VIST (see Figure 3-2) and CAE's (Montreal, Canada) CathLabVR, both reviewed in Chapter 2. However, there is no commercial simulator that includes the Seldinger technique, and the training session begins with a guidewire already inserted into a prepositioned entry site on a patient mannequin. The focus of this work is to develop the techniques and hardware to facilitate a virtual simulation of the initial steps of a vascular access IR procedure in order that a full virtual procedure training simulation can be developed. It is thought this could increase the content and face validity of the current virtual training solutions.

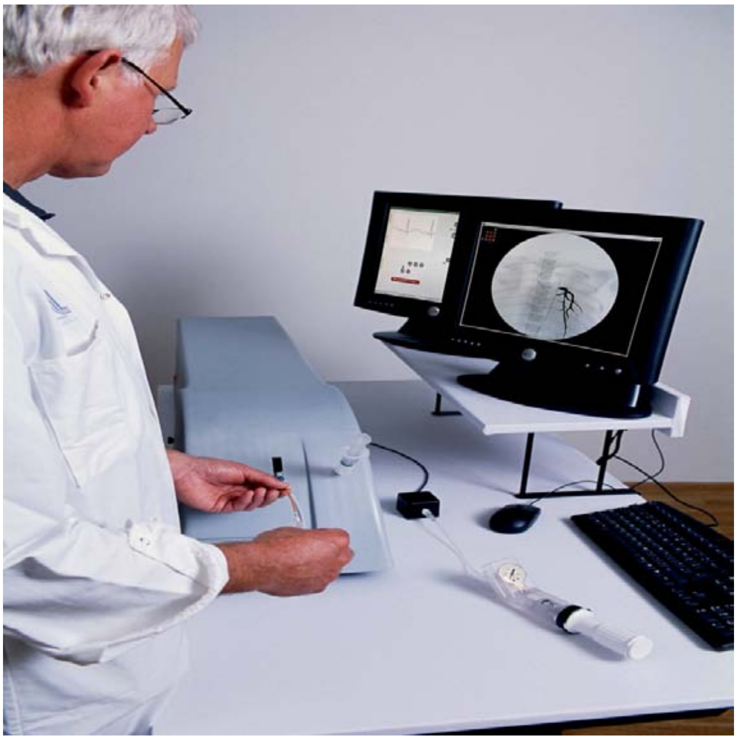


Figure 3-2 Meti, Vascular Intervention Simulator Training (VIST) Medicine Meets Virtual Reality January 25, 2003, Jonas Ohlson, Mentice, AB

3.2 Seldinger Technique

To identify the key challenges to be overcome in a successful virtual simulation of palpation for the femoral pulse with needle insertion for vascular access, a comprehensive task analysis of this IR procedure [182] has been closely evaluated. A step-by-step breakdown of an arterial palpation and puncture is given in Appendix 10.1 and described below.

The femoral artery, a large artery running down to the thigh before branching into smaller arteries that supply blood to the legs, can be palpated and used to access the vascular system as it passes over the femoral head, Figure 3-3. The common femoral artery is palpated over the bony femoral head as an 8 to 10cm long, pulsating, muscular walled, tubular structure. A practitioner will often locate the area of interest visually, however some practitioners will also palpate bony landmarks to localise the position of the pulse. They will then use two or three fingertips (Figure 3-4) to locate the pulse beneath the skins surface by pressing upon the patient's skin. After the artery has been located, local anaesthetic is injected into the surrounding tissue to numb this area. A small nick

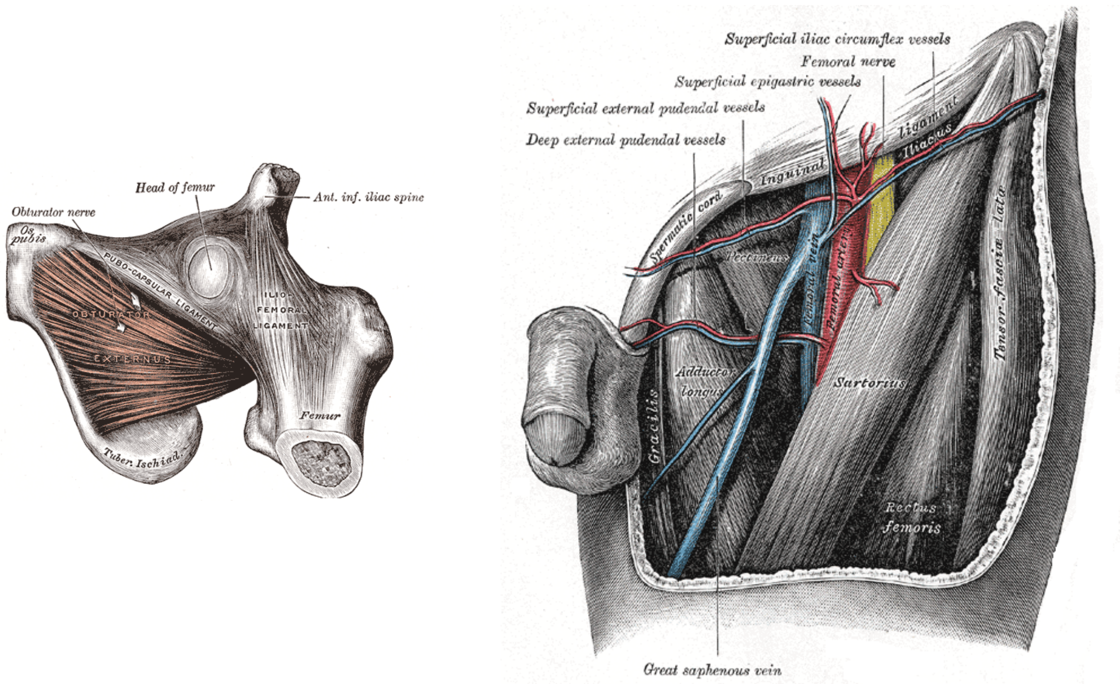


Figure 3-3 The femoral Artery can be felt as it passes up over a patient's femoral head. Images taken from "20th U.S. edition of Gray's Anatomy of the Human Body"

is then made at the prospective insertion point through which the interventional radiology needle is to be inserted. The skin nick reduces uncontrolled motion of the needle tip that would otherwise occur as it breaks through the skin surface. Again using palpation, and guided by the fine tactile cues felt at the fingertips in one hand, the practitioner then picks up the IR needle with the other. With the needle bevel directed up towards the practitioner, the needle is inserted through the nick. The resistance of advancing the needle through the tissue increases slightly with depth until the tip reaches the femoral artery wall, up to 7cm below the skin's surface dependent upon the patient's habitus. At this point, the pulsing of the artery can also be felt at the needle hub. The needle is then advanced again as the pressure builds up until the needle breaks through into the femoral artery. Blood emanates from the needle hub (Figure 3-4) until the wire is inserted through the hub, obstructing the egress of blood and is passed through the needle shaft and into the targeted vessel.

Once the wire has been advanced a sufficient distance into the femoral artery, thereby minimising the risk of dislodgement, the needle is removed over the wire and replaced by the requisite catheter. The catheter is then advanced over the wire and passed through the skin and into the vessel, tracking carefully along the guide wire.

Figure 3-4 Needle puncture *in vivo*. Left: A three finger femoral palpation and the initiation of a needle puncture. Right: As the femoral artery is punctured blood flows up through the need hub. Pictures courtesy of D. Gould.

Procedural methods can vary greatly between medical institutions and even between practitioners within a single institution. An informally surveyed American interventional radiologist described how his institution opted not to perform a skin nick before inserting the IR needle and, although the tissue is anesthetised during insertion, the needle is attached to a syringe of anaesthetic so that the tissue can be further anaesthetised as the needle is advanced. In this situation, only a small amount of blood will wash into the syringe and the fine forces felt during a needle insertion will differ from those felt if the skin had been nicked. This indicates how a global task analysis - outside the scope of this work - would be required to incorporate all inter-operative variability.

The focus of this research is to address the key technological barriers that limit the production of a full virtual Seldinger technique that can accommodate for procedural variability. However, the palpation model for a pulsing femoral artery and the subsequent needle insertion will ignore the procedural complexities of the skin nick. This small procedural variability could be added as future work, and methods that could be used to simulate this are proposed in Chapter 9. As such, during future discussion of needle insertion, the patient's skin will be assumed to be pre-nicked.

As a patient's body habitus varies, so does the necessary tissue displacement to feel a patient's pulse during palpation. In addition to this, the strength of the pulse also varies due to a range of pathologies that might be present in the vessel and also the patient's medical condition. A comprehensive virtual simulation should account for these two varying factors which have a profound effect on the tactile cues available, and thus the 'feel' of the procedure and indeed, at times, its difficulty.

3.3 Simulation Challenges

Currently, only mannequin based femoral palpation and access simulations offer close to acceptable procedure rehearsal due to the high fidelity visio-haptic cues required. An example of such a simulator is FemoraLineMan (Figure 3-5) from SimuLab (Seattle, USA). This physical structure offers a small true to life section of the human anatomy of which a subsection containing a simulated femoral artery allows for simulation of palpation and femoral access. The replaceable soft section of simulated tissue is made of a silicone like material. Of two tested models, one had a hard tissue that was designed to be more resistant to needle puncture and the second had more realistic properties but at a cost of being less durable. The manufacture suggests that between 25 and 50 punctures can be performed before the soft tissue section requires replacing. Damage such as that seen in Figure 3-5 render the model useless as the ideal puncture sight is clearly visible and the haptic feedback is severely compromised. After purchasing the simulation hardware, taking into account the manufacturers durability recommendations and the cost of tissue replacements, each femoral puncture costs between 6.5 and 13 US dollars. The pulsing of the artery is also suboptimal, requiring an expert trainer to manually pulse the artery. Such a requirement

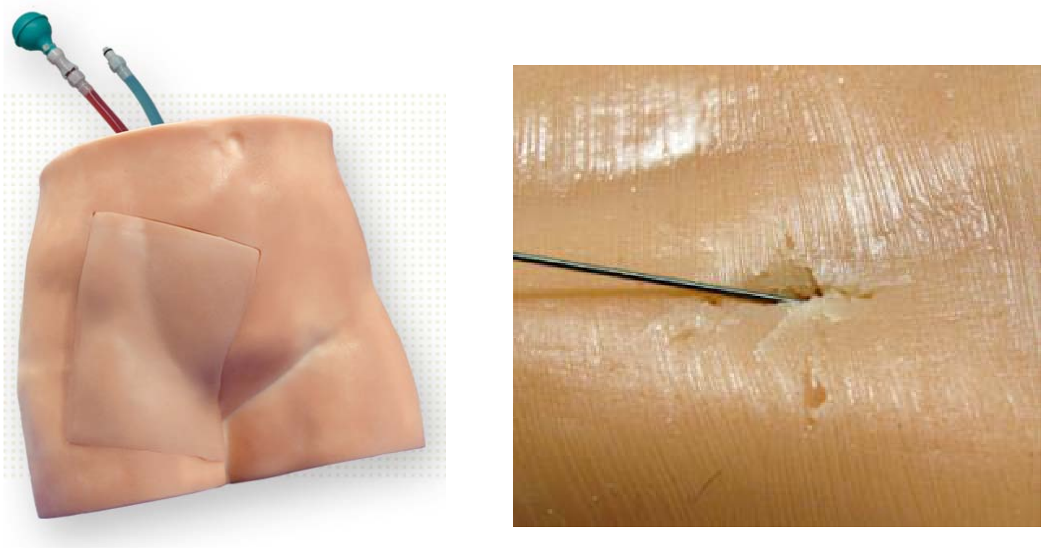


Figure 3-5 Multiple needle and wire punctures can cause rapid degradation of mannequin models. Left: FemoraLineMan from SimuLab (Seattle, USA). Right: puncture of CentralLineMan, SimuLab.

means that trainees cannot train unsupervised, increasing the cost of simulation significantly. Other companies marketing mannequin based femoral access simulations include Blue Phantom (Redmond, WA, USA) and Laerdal Medical (Wappingers Falls NY, USA).

Although femoral palpation and needle insertion has not been fully virtually simulated in a single solution, virtual needle punctures are frequently deployed for training a wide variety of procedures, see Chapter 2.8.2. There is a distinct lack of face validity in both current virtual palpation simulations and needle insertion simulations due to low immersion visio-haptic solutions, inadequate force feedback hardware integration and low fidelity tactile feedback through use of unsuitable haptic hardware interfaces.

As concluded in Chapter 2, there are few displays capable of combining visual and haptic cues in collocation, a necessity for true to life simulation of palpation and needle insertion. The conducted literature review finds the most common method used for collocated visual and haptic force feedback to be a semi-transparent mirrored display [46]. Whilst this method of feedback can be adequate for providing tool interaction with virtual environments, as neither the virtual environment nor the users hand are fully opaque during use, the simulation of direct practitioner/virtual patient touch interaction is hard to achieve in high fidelity. To overcome this problem, Chapter 4 introduces a new method of virtual environment visio-haptic collocation for medical simulation, employing augmented reality techniques.

Of the reviewed medical simulations, commercial off-the-shelf force feedback devices are commonly used alone to convey haptic feedback during palpation, with tactile feedback ignored. Current palpation simulations often require the user to palpate the patient with a tool, but during femoral artery palpation, practitioners use their fingertips to observe fine tactile variations beneath the skin's surface. Virtual simulations that intend to replicate direct hand-skin touch with a simulated patient have opted for a thimble approach into which a single finger is inserted, refer back to Figure 2-1, but these still do not provide the necessary tactile information in combination with the simulated force. Both tactile and force feedback are necessary for femoral artery palpation: force feedback to convey the force felt as the patient's tissue deforms beneath the practitioner's fingertips, and tactile feedback to convey the sensation of the skin deforming around the fingertips and the fine pulsing sensation which propagates through the patient's tissue from the femoral artery below the surface. During a real procedure, a practitioner can observe their hand freely moving above the patient, whilst feeling haptic feedback as they bring their hand into contact with the patient's skin. This introduces an additional requirement for simulation as a thimble based approach restricts the user's movement and can always be felt attached to the finger. No haptic hardware capable of simulating such a task was identified during the comprehensive literature review and, as such, the development of a novel device which will not restrict the user's movement in free space, but provide realistic force and tactile information as the practitioner reaches down to the simulated skin is described in Chapters 5 and 6.

3.4 Procedural Haptic Feedback

The haptic feedback felt must be accurately reproduced if the simulation is to provide meaningful feedback during training. Of the reviewed medical simulations, very few publications provided evidence of using measured force data, with most appearing to tune their simulation using experts to judge if the simulation "feels right". Although expert feedback is useful, a quantifiable feedback is more desirable to draw scientific conclusions about the accuracy of the simulator.

Force measurements of both the palpation and needle insertion forces have been performed *in vivo* such that the results can be used in the simulation development described in the following chapters. The development of the force sensors and the force measurements have been carried out by collaborators at the Royal Liverpool University Hospital [183]. As this work has not yet been comprehensively described through publication, a brief summary is provided below.

3.4.1 Measured Palpation Data

A fingertip shaped cantilever-beam force sensor (Figure 3-6) was developed by the Clinical Engineering Department at the Royal Liverpool University Hospital to measure pulse palpation forces. The sensor's silicone fingertip end effector was positioned over the femoral artery of test subjects using a 10MHz ultrasound probe to ensure correct alignment prior to measurement. The force measurement device used was securely fixed to the ground to ensure steady force readings and accurate displacement information.

The sensor was periodically displaced toward the femoral artery in 5mm intervals. At each known skin displacement, the forces felt at the sensor were recorded for 10 seconds. Each recording encapsulates the resistive force of the patient's tissue at that displacement, the force pushing upwards toward the force sensor in relation to the distance the skin is displaced. Also captured within this data is a low force, high frequency fluctuation that represents the pulsing of the femoral artery below the skin's surface. This becomes more prominent as the depth of displacement increases.

A single patient's data has been used in the development of this simulation as at the time of writing only one patient's data was available for use. However, the simulator framework is designed to support multiple patient data by providing capabilities exceeding those required to simulate this "average" patient, as subjectively judged by IR experts. The simulated patient's tissue above the anterior wall of the artery was 20.7 mm thick and the diameter of the femoral artery was 12.4 mm. This data was used to make informed decisions about the requirements of the produced simulator and was also used to produce a realistic feeling force profile. The average maximum forces for each displacement have been plotted to produce a cubic function used for force approximation during the simulation (Figure 3-6). The full interpretation and use of this 3D dataset, describing displacement toward the femoral artery and the resistive force of the displaced tissue is described in Chapter 6 and the results are validated in Chapters 7 and 8.

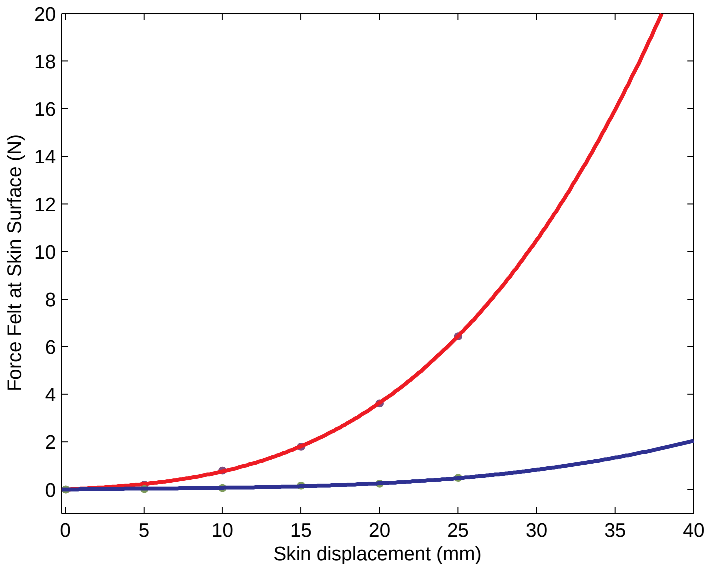


Figure 3-6 A fit of the average forces recorded in-vivo per known skin displacement toward the femoral artery in a thin healthy subject. Red- Palpation force. Blue- tactile force from pulsing femoral artery. Projected from 25 mm to 38mm to avoid excessive force applied to patient. Forces from [183] Right: Finger-tip shaped cantilever-beam force sensor developed by Dr. J. Zhai and Dr. T. How.

3.4.2 Measured Needle Insertion Data

Needle insertion forces were measured *in vivo* using a magnetically tracked needle with an integrated force sensor, again developed at the Royal Liverpool University Hospital. This outline summarises communication with both Dr. J. Zhai and Dr. T. How on unpublished work. The developed sensor uses a cantilever beam configuration mounted with four commercial semiconductor strain gauges from Micron Instruments (California, USA). The sensor can undergo sterilisation to allow for multiple uses. The device is highly sensitive such that the varying resistance to penetration between tissue types can be discriminated. An example trace produced from the *in vivo* force data can be seen in Figure 3-7.

Prior to each recording, the interventional radiology needle is connected to the Luer connector of the sensor. During recording, the operator holds the force sensor in place of the normal needle hub. Forces are recorded as the practitioner manipulates the needle through the skin's surface, then toward and into the femoral artery. Recording stops as blood flows through the hub, indicating a successful arterial puncture.

Detailed specifications of both the needle and palpation force sensors are not included in this thesis as this work was performed by Dr. J. Zhai and Dr. T. How and has yet to be published.

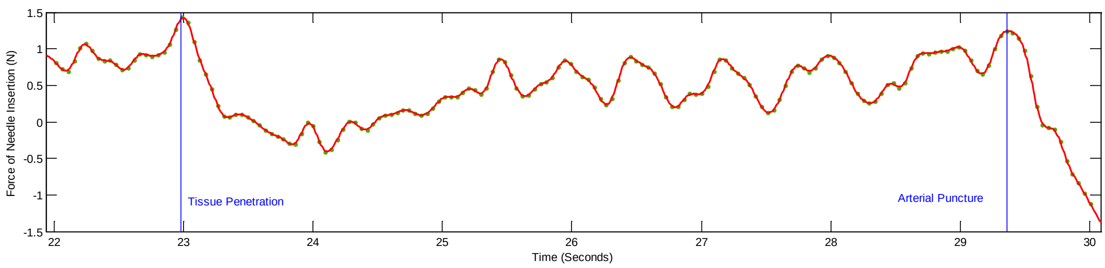


Figure 3-7 *In vivo* force measurements of a femoral needle puncture. Figure produced from unpublished *in vivo* force measurements kindly provided by Dr J Zhai and Dr T How.

3.5 Simulation aims and goals

In the following chapters, an exemplar medical simulation environment, named PalpSim, is developed to simulate a femoral artery palpation and femoral needle insertion, the beginning steps of an IR procedure. This has not been fully virtually simulated before due to the complexities of simulating direct touch between the practitioner's fingertips and a virtually simulated patient's tissue. This virtual simulation, if successful, could provide the initial steps of a virtual interventional radiology procedure for a variety of patient habitus' and medical conditions. It is thought that this simulation could increase the face validity of current IR wire manipulation simulations and that the initial steps of the intervention could be practiced before the practitioner tests their acquired skills *in vivo*.

Four main problems limiting the development of such a solution have been identified to be addressed in this work: Visualisation problems inherent in current medical simulations that permit visio-haptic collocation, none of which cope well with direct touch of a virtual patient; A distinct lack of commercially available tactile feedback devices that can be integrated with a force feedback device for femoral palpation simulation; Cost prohibitive force feedback hardware suitable of conveying the forces felt during a femoral palpation; and low simulation face validity during current needle insertion simulations due to the use of commercial devices which do not reflect the look or feel of a real needle interface. Potential solutions to these issues are described in the following chapters with a face and content validation of PalpSim described in Chapter 8, and a summary of the work conducted and proposals for further work described in Chapter 9. The following chapters are:

- Chapter 4 – Developing a visual solution for collocated visio-haptic interaction.
- Chapter 5 – Development / evaluation of four potential tactile solutions, an optimal solution is selected for use in PalpSim.
- Chapter 6 – Development of force feedback devices for palpation and needle insertion.
- Chapter 7 – The integration of the constituent parts of PalpSim (each of which are described in chapters 4, 5 and 6) to produce a complete simulation solution.

4 Visual Feedback

4.1 Introduction

During many interventional procedures, the practitioner stands over a patient lying on their back upon an operating table. The practitioner will then reach down to palpate the patient's skin with one hand and guided by this, will puncture a needle into the femoral artery with the other hand. Simulation of this procedure has provided the exemplar application for exploring the research ideas presented in this thesis.

The visualisation of a reality-based procedural training simulation should be natural. In a best case scenario, the virtual visualisation will go unnoticed by the user whilst cues can be introduced to modify the user's perception of reality. See-through HMD technology offers such a visualisation opportunity, in theory but in practice, the technology only offers a limited field of view, low visual resolution and is cumbersome to wear. These limitations make it hard for the user to be fully immersed. Additional limitations of a HMD are its high cost, exasperated by the necessity to use high quality head tracking hardware. The collocated visio-haptic alternative to this display, the popular semi-transparent immersive workbench [46], provides a natural interface into which the user can look but also requires the user to wear glasses and suffers from occlusion problems [48]. As such, an alternative to these hardware solutions is investigated here.

An augmented reality (AR) visualisation approach has been developed that replicates a real IR scenario. A standard LCD monitor over which the user stands is used to visualise a patient below them, whilst an image of the user's hands is captured and displayed in full opacity in the virtual scene. This display design encompasses the advantages of the immersive workbench whilst overcoming the undesirable occlusion visualisation problems inherent in these displays, described in Chapter 2. The development of the visual components leading to the realisation of the AR medical training environment is described within this chapter, after the initial visual approaches have been described to provide context. The development of a visio-haptic workbench using this technology is described in Chapter 7.

4.2 Initial Patient Visualisation Approaches

In an initial approach to realistic patient visualisation, real patient data was to be integrated into a finite element deformation model. This work is presented here for completeness and context and is not used in the final visual simulation described in section 4.3. If accurately modelled, this approach could recreate realistic visual and haptic feedback as a patient was palpated. The freely available male dataset from the visible human project [184] was used as a case study.

Initially, an area of skin surface was identified and meshed using ITK-Snap [185], an open source semi-automatic segmentation tool for 3D medical images, see Figure 4-1. The mesh was then refined using RapidForm from INUS Technology Inc. (Seoul, Korea) and 3D Studio Max from Autodesk (San Rafael, USA). Holes in the mesh from where the patient’s hands met their stomach in the scan were filled and the mesh was then re-meshed to reduce its resolution. The patient's femoral artery and other tissues were then to be extracted to produce a realistic deformation model. However, the approach to patient visualisation was changed before this was performed. The new approach is described in section 4.3.

Initially, the extracted surface mesh, seen in Figure 4-1, was used in a rigid body Chai3D simulation [49]. This simulation environment allows meshes in an .obj file format to be imported into an OpenGL visual environment and provides simple rigid body proxy based haptic feedback and collision detection information.

Figure 4-2 depicts the visual setup using this mesh data. In this first simulation prototype, a three dimensional representation of an operating room was created

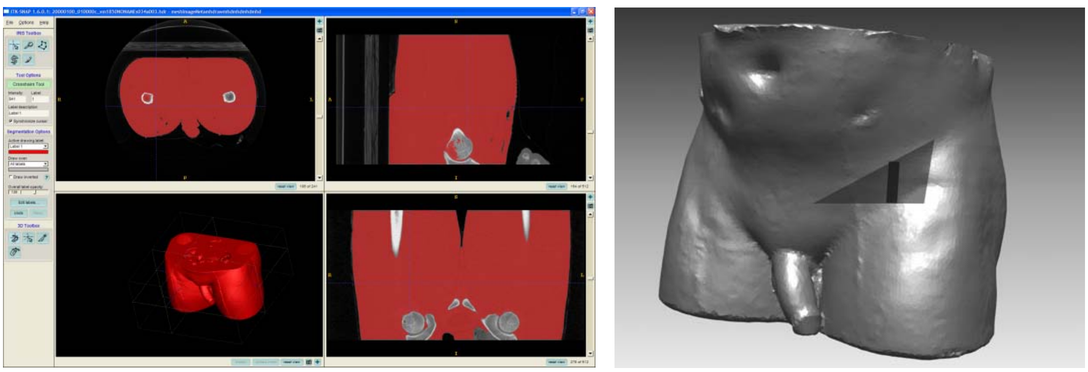


Figure 4-1 Left: Mesh segmentation using ITK-Snap [1] an open source semi automatic segmentation tool for 3D medical images. Right: Location of the palpatable femoral artery marked on a male skin mesh segmented from the visible human project dataset.

to give the simulation context. A virtual draped patient could be seen lying on their back on top of an operating table in the middle of this room, represented by a mesh of a fenestrated surgical drape (a drape with an opening in the centre) produced in 3D Studio Max. Two cut outs in the drape allowed the user to see and touch the skin. An LCD monitor was positioned horizontally and a modified Falcon haptic device, rotated through 90 degrees with a modified end effector (see Chapters 5 and 6) was placed underneath. As the trainee palpated the patient, the view of the simulated operating room was moved so that the practitioner appeared to be standing over the patient. The trainee could not see their hands and during an informal evaluation with colleagues in which verbal feedback was given the simulation was rated as providing only low immersion.

A possible solution to this was to provide a virtual hand avatar that followed the position of the haptic end effector as it was moved. A low resolution 3D hand mesh, which had a static size and pose for all users, was therefore used for these initial tests to gauge the complexity and effectiveness of this approach. It was quickly discovered that for a hand avatar to be effective, the avatar must accurately mimic the pose and configuration of the user's hand below the monitor. Therefore, the user's hand must be tracked in 3D using approaches such as those by Wang *et al.* [186] and Guerin *et al.* [187]. Although this approach was plausible, it was felt a hand avatar may still only offer a low face validity simulation solution as the virtual avatar would not look like the user's real hand.

During this initial development it also became apparent that an accurate finite element model of tissue deformation around the femoral artery was infeasible at either haptic or visual rates. Unlike engineered materials, tissue is highly irregular and the interactions of muscle, fat and vessels are computationally expensive to solve even at a coarse level. In addition, each fingertip contacts a patient's tissue at many contact points limited only by the resolution of the skin surface considered, another complex interaction to solve in real time.

A second approach aimed to simulate visual and haptic feedback of deformation using BulletPhysics' soft body library [188], a position based solver. This model was visually unsatisfactory as the deformation of the coarse model caused visible sharp edges as the mesh deformed. An approach to overcome these artefacts is to map a high resolution visual mesh model to a low resolution mesh used for haptic rendering and collision detection, e.g. Lim *et al.* [189], but a decision to improve the haptic fidelity influenced a change to further decouple the visual and haptic calculations.

The solutions implemented to overcome these problems are described in the following sections.

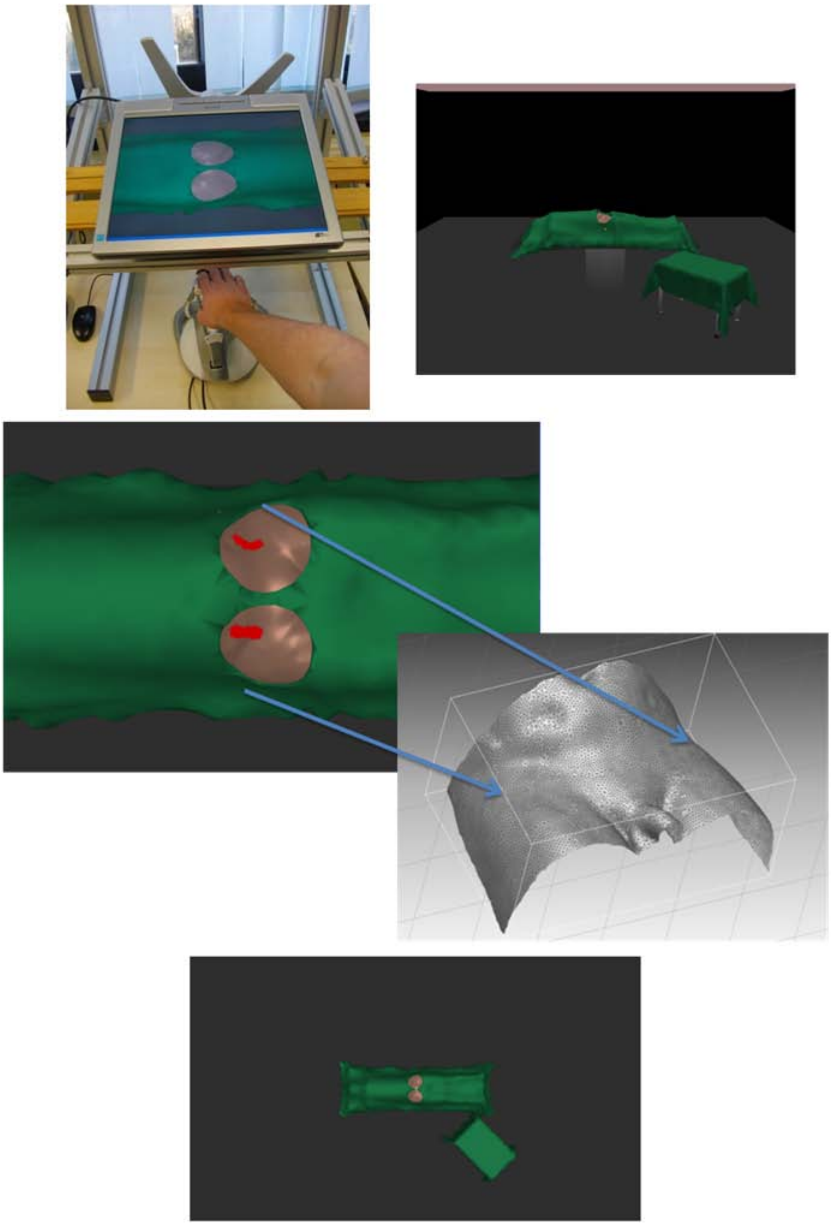


Figure 4-2 Chai3D visualisation used in the first palpation prototype. Top left: Virtual world visualisation during palpation with modified Falcon device underneath. Top right: Side view of virtual operating room. Centre: Highlighted pulsing areas and the trimmed patient mesh to reduce the number of mesh nodes. Bottom: A top view of the operating room environment.

4.3 Visual Components of PalpSim

This chapter describes the final visualisation approaches used in PalpSim. The augmented reality visualisation workstation through which the user views the PalpSim environment is described in section 4.3.1, and the shadowing effect of the trainee's hands, used to increase the user's sense of depth is described in section 2.3.2. The individual OpenGL components; the fenestrated drape, deformable skin, virtual needle and simulated blood flow, are then described. The placement of these components in relation to the haptic and visualisation hardware is detailed in Chapter 7 after the tactile and force feedback hardware has been described. In Chapters 5 and 6 respectively.

4.3.1 Augmented Reality Display

The simulation is visualised on an LCD display mounted at an angle of 30 degrees from horizontal, allowing the LCD screen to be viewed without colour distortion or glare, whilst maximising the size of the workspace behind it, Figure 4-3(1). An

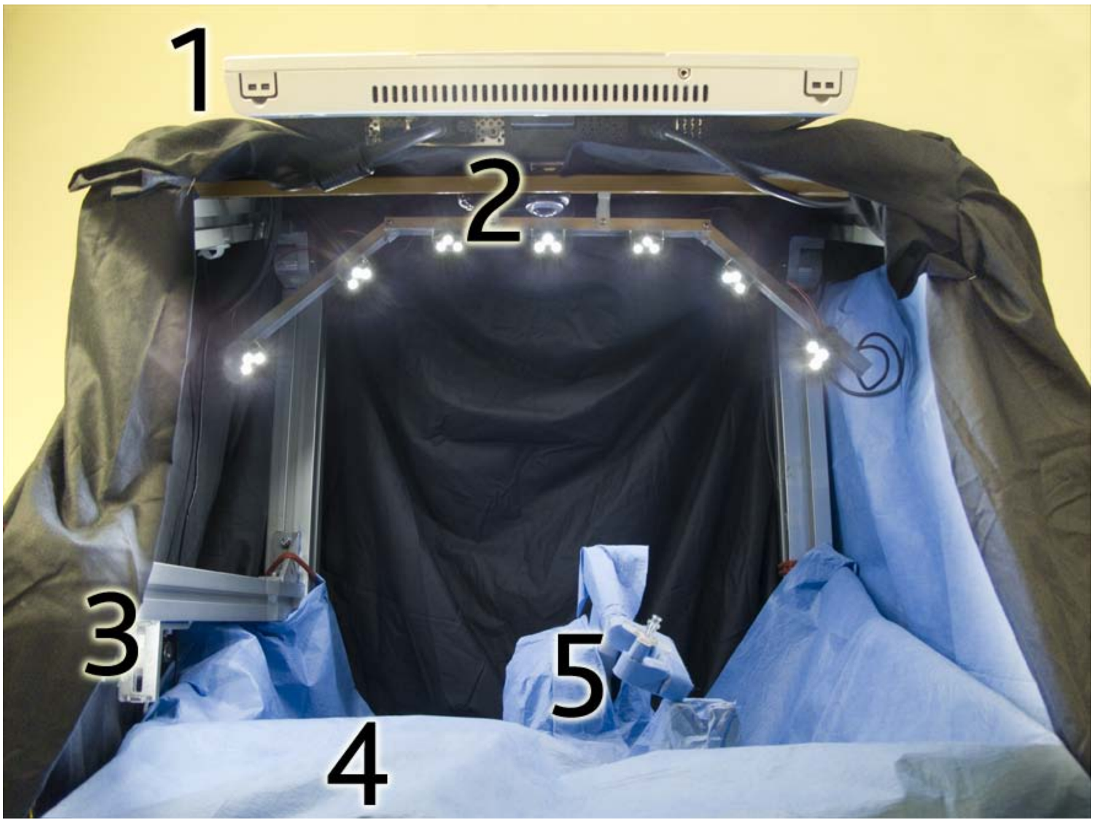


Figure 4-3 A collocated Haptic / Augmented Reality workstation. The LCD display (1) and camera (2) are mounted above the haptics devices and are used to display a live feed of the user's hands. Bright lighting illuminates the workspace to achieve a fast shutter speed. A low resolution side mounted camera (3) is used in a shadowing effect of the user's hands. This detects the height of the user's fingertips above the palpation haptic device (4) hidden below the blue sheet. The real needle hub (5) is attached to a modified Omni haptic device to provide 3 force DOF force feedback.

immersive AR environment is created using a wide-angle camera attached underneath the monitor to film the real time interaction below it, Figure 4-3(2). As in a traditional immersive display, the user can move their hands without resistance in the free space below the LCD display. As the user places their hands underneath the display, the hand image is extracted from this video stream and positioned within the training environment allowing the user to appear to be able to see through the monitor to their hands below. A virtual skin and a fenestrated drape (described in section 4.3.3) are placed in the scene underneath the user's hands to produce the illusion of a virtual patient lying beneath them. Haptics devices aligned with this visual environment provide force and tactile feedback, see Figure 4-3(4,5). These devices are described in the following chapters.

The hand extraction procedure can be broken down into two stages: a pre-processed acquisition stage, which need only be performed once in which the chromatic range of the hand image is captured; and secondly, during simulation, a continuous loop in which the hand image is extracted.

The main image processing functionality uses the OpenCV library [191], an open source library for real time computer vision. An RGB (Red, Green, Blue) image is read from the camera mounted underneath the LCD display. A copy of this image is made and transformed to its HSV (Hue, Saturation and Value) colour representation. This colour representation is used to increase the extraction's robustness to fluctuating light when compared to an extraction using only the RGB colour descriptor. By considering only the *hue* and *saturation* of each pixel and ignoring a pixel's *value* (a description of brightness), a pixel's colour should be independent of the intensity of light reflecting off its surface. As such, the intensity of white ambient light from the surroundings should not affect the extraction quality.

The acquisition stage is performed offline in a separate application to the real time PalpSim visio-haptic software to produces a colour descriptor file that is loaded on initiation of PalpSim and used within its graphics loop. PalpSim and the acquisition program have been developed as standalone applications, as the acquisition phase needs only to be run once unless the user changes the colour of gloves used within the simulation or the ambient lighting changes significantly (i.e. the simulator is moved to a new location). Separating these modules allows the acquisition program to be used and distributed independently, an important design factor as it is thought the AR functionality developed for PalpSim has a wide applicability for many types of medical simulation. The acquisition application is an OpenGL program displayed in a GLUT (OpenGL Utility Toolkit) window. The OpenCV image processing library is also required. The program is primarily controlled via a series of keyboard keys for simple one handed operation. A cube object is placed in the black OpenGL environment for testing purposes and a texture map is used to display the camera image, Figure 4-8.

4.3.1.1 Acquisition Phase

Initially, a user must place their gloved hand into the workspace area below the LCD screen. The camera eye view of the user's hand will appear rendered as a texture map in the OpenGL world. Using the mouse, a square target of 400 pixels (20x20), depicted by 5 red markers can be positioned over sections of the image that represent the users hand and needle within the scene, Figure 4-4. The HSV colours of pixels within this target (data set P) are captured for processing as the "R" key is pressed. Both the *hue* (h) and *saturation* (s) of a pixel will range between 0 and 256 and therefore the complete *hue* and *saturation* colour range can be stored in a single 256 by 256 square data structure, C[256][256], where the value of

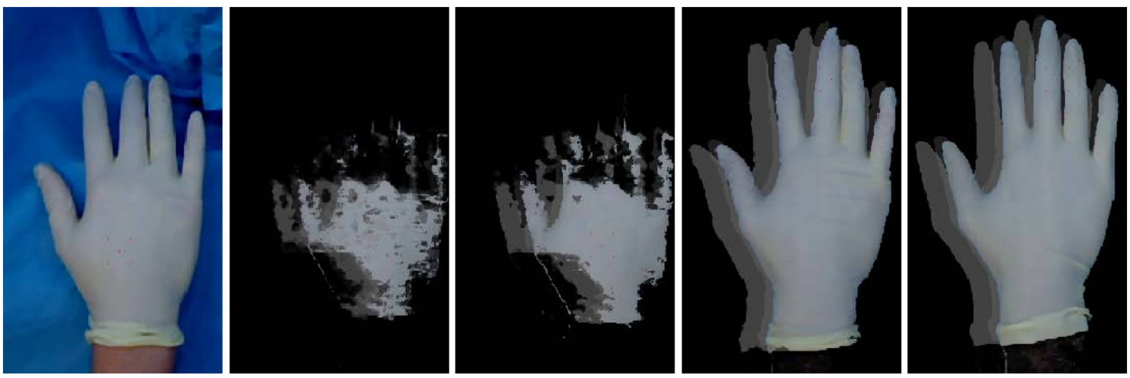


Figure 4-4 The acquisition stage. From Left to Right: A faint red selection target can be seen in the video image of a users hand. 2nd: The hand extracted after a single chrominance acquisition pass. 3rd: Two acquisition passes. 4th: 10 acquisition passes. 5th: 20 acquisition passes.

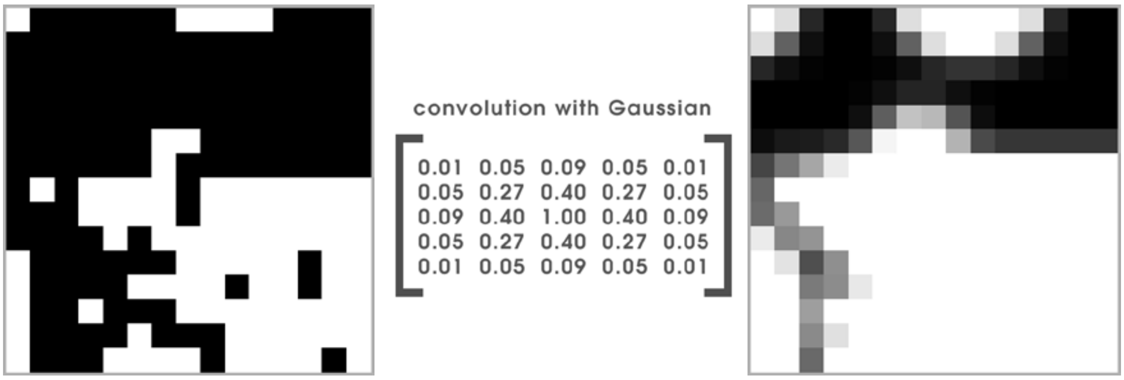


Figure 4-5 Passing a 5x5 Gaussian filter over a 15x15 subset of dataset C. Left image depicts a first incomplete capture of the hands chromatic range. Elements of value zero are rendered black, those of value 255 (hand) white. By applying a Gaussian filter to the data, neighbours that are not discovered to be hand but are surrounded hand coloured pixels will also become hand. Pixels at the boundary, will acquire a value between 0 and 255 and will be rendered semi opaque.

hue denotes the column and *saturation* the row, see Figure 4-4. On program initiation or reset, all elements in the data structure are initialised with zero values (denoting a fully transparent pixel). For each member of P (pixel colours captured as representing the hand), a value of 255 (denoting opaque) is given to C at the position corresponding that member's *hue* and *saturation*. This value will be used in the hand extraction phase to determine the opacity a pixel should have within the live image stream where 255 represents full opacity and 0 represents transparent (explained in next section). After all 400 members of P have been processed, dataset C is then smoothed with a 15x15 Gaussian filter and normalised. This convolution both aids a faster convergence to a satisfactory hand extraction and by expanding the chromatic range of the chrominance model, produces an alpha blending of pixels that fall on the hand/background barrier. This is explained below.

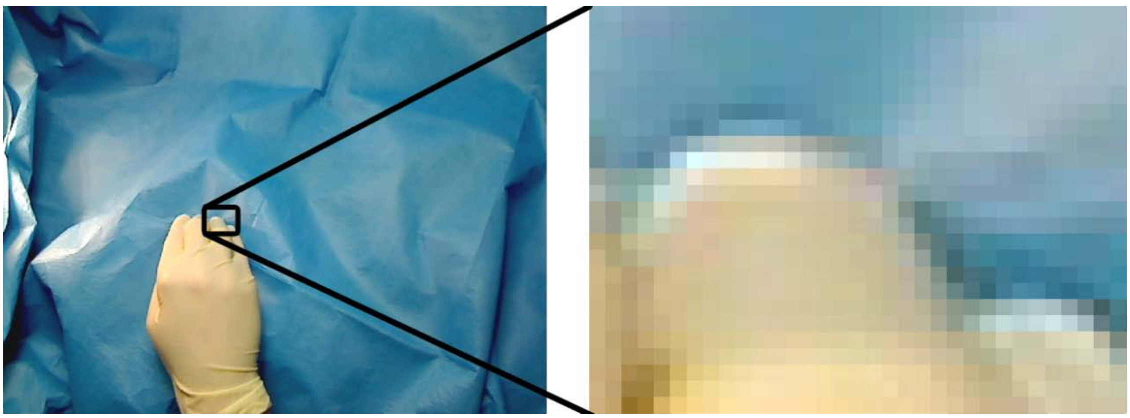


Figure 4-6 The Gaussian smoothes the captured colour range of the hand. Pixels which are at the hand / background boundary are rendered semi transparent so smoothing the hand image into the virtual scene.

Figure 4-5 demonstrates how convolving the captured chrominance model with a Gaussian filter smoothes the model for use in PalpSim. As the hand image is captured, a complete set of all of the colours that represent the hand may not be recorded in a single 400 pixel capture. Multiple captures of the hand's colour will improve this, but it is possible that small fluctuations in colour may not be captured so during the hand extraction phase, a hand image would look incomplete. These un-captured colour values can be seen as black pixels within areas of predominantly white in Figure 4-5. In this situation, the filter's smoothing effect will mark elements in dataset C as hand if a high proportion of its neighbours have been deemed to represent hand, i.e. they have appeared in a set P.

Figure 4-6 highlights the second advantage of using a Gaussian filter. In this figure, a subsection of an image captured from the downward facing camera is expanded to reveal the pixel granularity. It is hard to define at which pixel the users gloved fingertip ends and the green drape starts. If the chrominance model is too strict, the edge of the user's fingertips will be cropped and if it is not strict enough, the drape will be captured adding a border around the extracted hand image. The Gaussian blur of the chrominance model, see Figure 4-5, assigns a value between 0 and 255 to element neighbours or near neighbours of elements that represent a known hand colour (so have a value of 255). This dictates a varying pixel opacity dependent upon the likelihood the pixel is within the hand

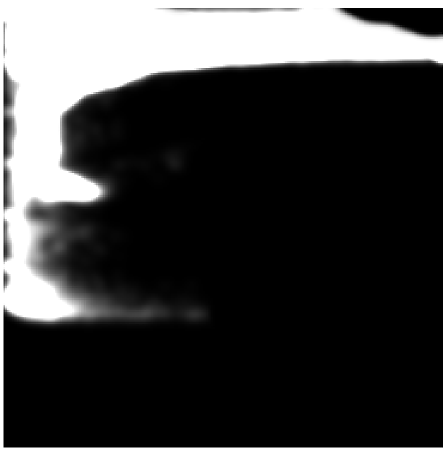


Figure 4-7 A grey scale representation of data set C. This can be conveniently stored as an 8 bit grey scale image for future reference.

(255, full opaque), at the border of the hand (255 – 0, semi opaque) or the background (0, transparent). The real time image processing for hand extraction is further explained in the following subsection.

As the colour capture procedure can be repeated, a chrominance model into which repeated passes are accumulated is stored in addition to a Gaussian filtered chrominance model. This is necessary, as if a Gaussian filter were to be passed over the model twice the number of elements defined as hand would grow without added information.

The quality of the hand extraction can be gauged in the same GLUT window used to select areas of hand image. When a satisfactory extraction has been made, data set C's 256x256 elements will contain a value between 0 and 255. This can be conveniently stored in a similar sized 8 bit grey scale image (stored here as a *.tiff image), encoding 256 colours of grey, see Figure 4-7.

4.3.1.2 Continuous hand extraction

During simulation, a video feed of the interactions below the monitor is manipulated and placed into the virtual environment. On program initiation, the saved *.tiff chrominance model produced in the acquisition phase is loaded into dataset CH[256][256]. In a continuous loop, HSV and RGBA (Red, Green, Blue,

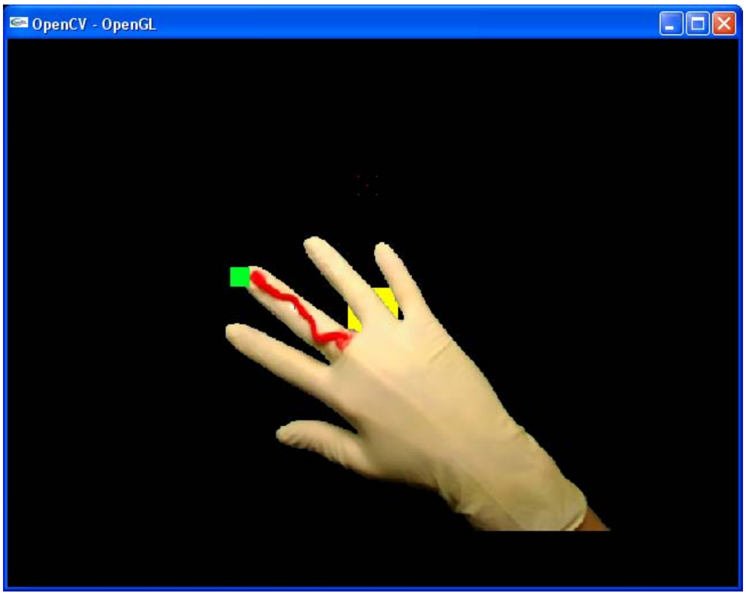


Figure 4-8 Acquisition program. A hand and marker have been extracted. The hand occludes a yellow cube within the scene for testing. The position of a red marker denoted the fingertips position, highlighted here with a green square.

Alpha) images are created from the RGB video stream. Each pixel in the HSV image is traversed and its *hue* and *saturation* used to address elements in dataset CH. At the corresponding pixel, the RGBA image's alpha value is assigned the value at CH[h][s] which will range between 0 to 255. As a complete image is traversed, the RGBA image is placed on the image buffer to be rendered in the virtual world and a new image is captured from the camera to be processed. It can be seen that pixel colours that weren't captured as representing hand will be transparent and so not seen, and conversely pixel colours that were captured as representing hand will be rendered fully opaque.

The Gaussian filtering that was applied will have assigned pixels of a colour close to that of glove colour, a semi opaque alpha value. This allows the glove image to realistically blend into the virtual world behind it. If this alpha blending does not occur the hand's appearance in the virtual world resembles that of a cardboard cut out hand and so does not appear to be realistic. The integration of this image into the complete virtual world is described in Chapter 7 "Realisation".

Figure 4-8 not only shows how an extracted hand can occlude objects behind it, but also how, given a coloured target, particular points on the hand can be effectively tracked. In this case, a red line running down the index finger is tracked and marked by a green square. This additional colour tracking feature was not used in PalpSim. However, the height of a user's hand from the virtual skin (and palpation haptic device) is determined to implement a shadowing cue.

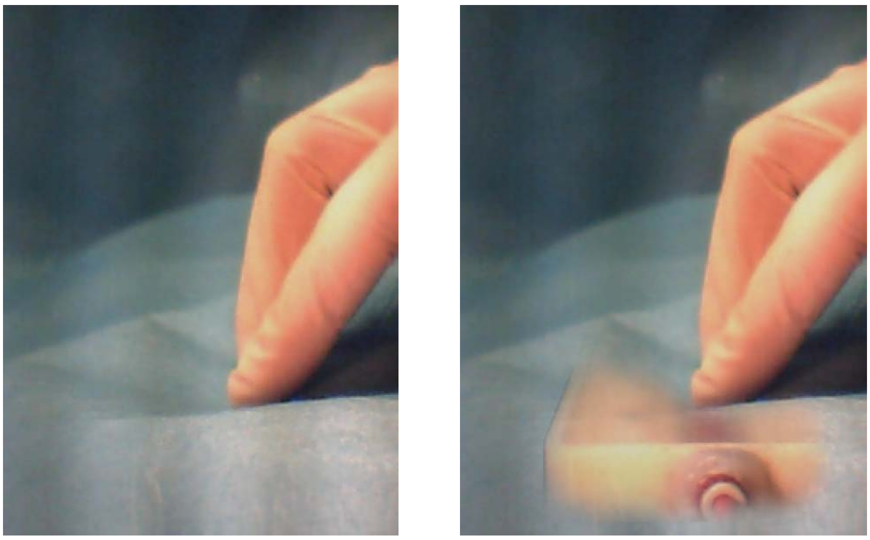


Figure 4-9 Left: the side mounted cameras view of a palpation. Right: the A superposed view of the haptic hardware below the surgical drape.

Figure 4-10 gives an overview of how the chrominance model acquisition stage described in section 4.3.1.1, and the continuous hand extraction stage described in section 4.3.1.2, are used to produce a real time video stream of the extracted user's hands for use in the AR PalpSim environment. The acquisition stage is performed prior to the operation of PalpSim and records a Gaussian blurred chrominance model to file that is loaded as PalpSim is initiated.

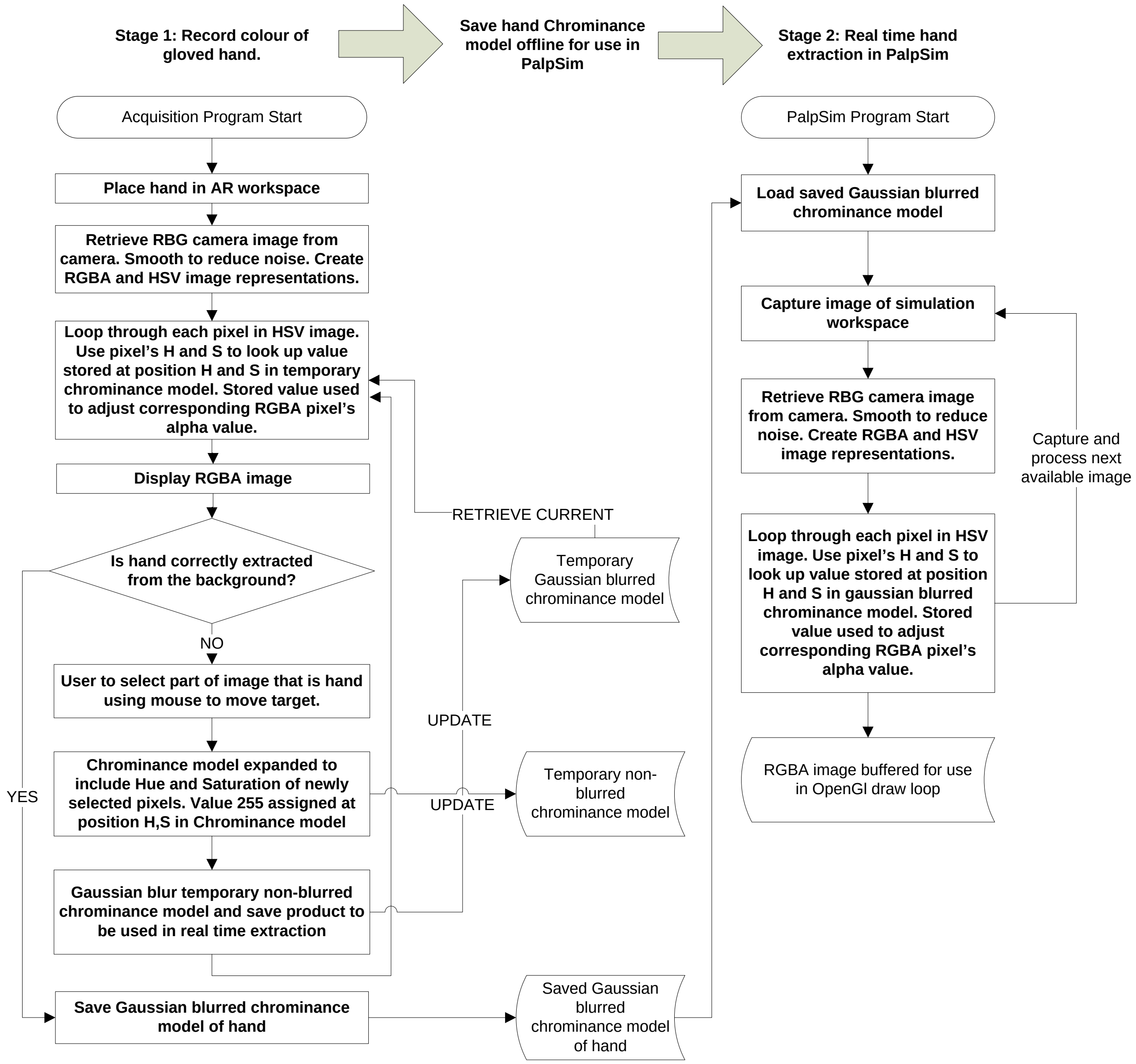


Figure 4-10 A flow diagram describing the initial acquisition of a hand chrominance model and the use of the chrominance model in PalpSim's continuous hand extraction loop. Full explanations of these two stages are given in sections 4.3.1.1 and 4.3.1.2.

4.3.2 Shadowing of the User's Hand

Shadows are used in PalpSim to alleviate inherent depth perception problems whilst using a 2D fixed viewpoint visualisation of a 3D scene. Shadows provide an indication of both the height of the user's hands, and the height of the needle tip from the simulated patient's skin. The shadowing of the palpating hand relies on a second low cost camera providing a lesser quality image to that providing the augmented reality visualisation, see Figure 4-9. Using similar hand extraction methods, a chrominance model for this camera is first captured and, in real time, the hand is extracted from the video stream. The lowest point of visible hand (the fingertips) observed in this image, coupled with the location information of the virtual skin (the height of the palpation haptic device), is used to calculate the height of a user's hand above the virtual skin. During the OpenGL draw procedure, two semi-opaque shadow images, grey scale images of the extracted hand image, are scaled and displaced from the original reference image of the hand depending upon a factor of the hand's height from the skin. As the user's fingertips approach the virtual skin's surface, the displacement of the shadow images from the hand image decrease to zero as the user's fingertips come into contact with the virtual skin/haptic device. This shadowing effect is shown in Figure 4-11. The two shadow objects are rendered at different opacities to provide the illusion that the shadows are created by two different intensity light sources and an ambient light.

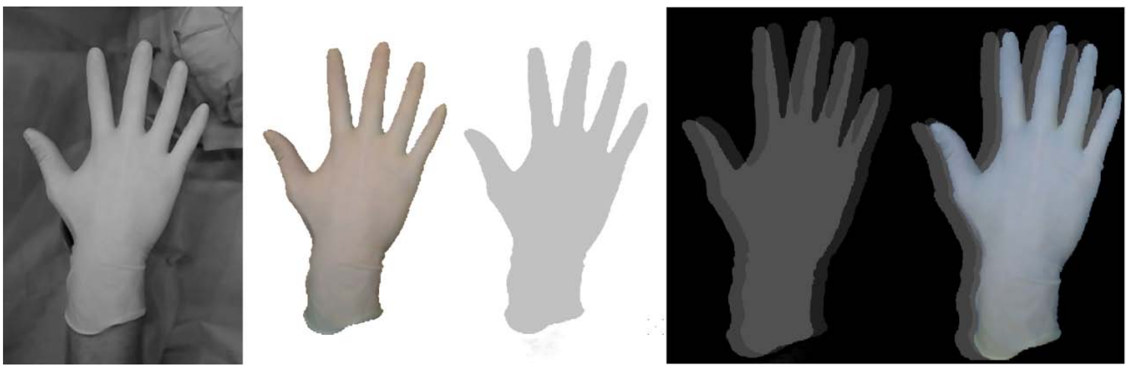


Figure 4-11 From left to right: A greyscale image of the captured video stream. An extracted hand image. A shadow representation of the extracted hand. Two shadows merged. The shadows and hand image indicate the hand is above the skin as when a practitioner touches the skin the shadows cannot be seen.

4.3.3 Virtual Skin and Fenestrated Cloth

In contrast to the initial approach, real patient scans are not used to produce a virtual patient. Instead, a single small visible section of computer generated deformable skin 9.5 cm long and 6 cm wide can be seen by the trainee. This section of skin matches the size of a cut-out in an IR fenestrated drape. The skin is modelled as a circular mesh with its visual behaviour controlled by Bullet Physics' soft body library, a position based solver based on work by Matthias Muller [190]. The circular mesh is pinned at its edges to constrain it to a defined position, whilst the nodes and the triangular faces within the fixed boundaries deform. A mesh of 1001 nodes (50 rings of 20 nodes and a central point) was found to provide a good deformation, whilst allowing for real time performance. Material parameters of the mesh such as damping and drag coefficients and bending constraints have been defined to achieve a visually pleasing deformation as the mesh is deformed by a finger proxy object. This object is modelled as a spherical rigid body and is coupled to the palpation haptic device, as described in Chapter 6. The proxy object deforms the skin mesh in a spherical profile as the user applies pressure to the virtual skin. The size of the deformation depends upon the pressure applied by the user on the force feedback device. The pressure / deformation ratio is calculated in the haptic loop at 1000Hz to provide smooth haptic feedback and the position of the haptic end effector is then relayed to the visual loop at 60 Hz to control the proxy position. The edge of the deformable skin section is covered by a fenestrated surgical drape, a photo texture map of a drape taken from the OpenGL

Figure 4-12 Left: The skin textured bullet physics skin mesh. Right: Fenestrated sheet texture map, created from three merged images.

camera viewpoint. The position of these two components within the virtual environment is discussed in Chapter 7 “Realisation” after the haptic feedback has been described.

4.3.4 Virtual Needle

A virtual needle shaft and hub follows a haptic device with a real needle hub attached to it (see Chapter 6). The virtual needle follows the real needle hub as it is translated and rotated in the real world space. It was identified during the task analysis [182], that the correct orientation of the bevel at the needle’s tip, before the needle is inserted, is considered a critical step in the procedure. As such, a bevel has been added to the needle and its sharp edge highlighted using a grey border as, without this extra cue, it is extremely difficult to see when the needle is at arm’s length, see Figure 4-13. A shadow, rendered as a line primitive, is displayed in the scene to convey depth. The shadow’s position and size within the PalpSim environment is discussed in Chapter 7.

As the position is precisely tracked, the needle is rendered in collocation to the real world needle hub held by the user. When this is coupled with the augmented reality visualisation described in Section 4.3.1, the virtual needle hub is obscured by an image of the real needle hub, so only the virtual shaft is visible.

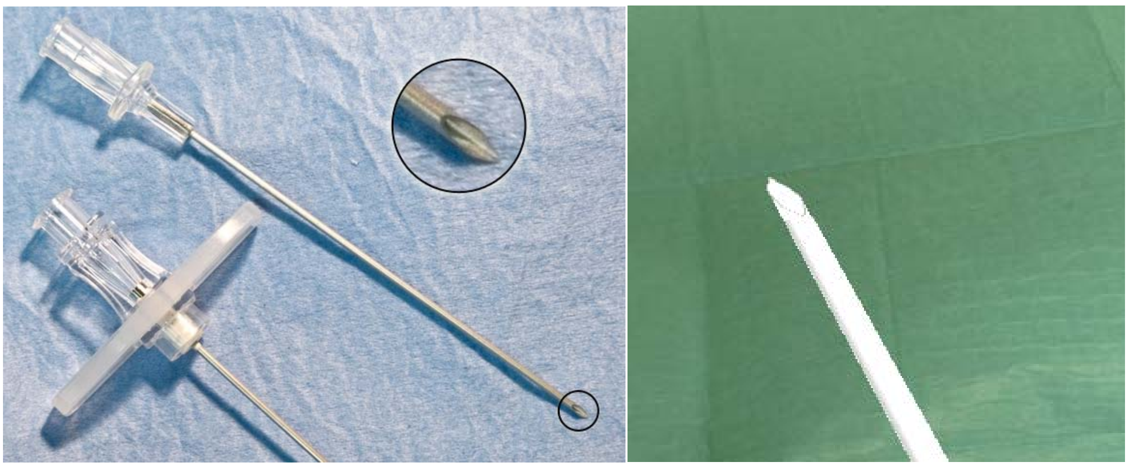


Figure 4-13 Left: Interventional radiology needles. The correct orientation of the bevel at the needle’s tip before the needle is inserted is considered critical in the task analysis. Right: The virtual bevel.

4.3.5 Simulated blood flow

As a patient's femoral artery is punctured, blood will flow out from the needle hub. This is an important visual cue when performing an IR femoral needle puncture. The blood flow indicates that the tip of the needle is within the femoral artery, and that a wire can be inserted through the needle and into the vascular system. During simulation, as the tip enters and stays within the artery, simulated blood emanates from the needle hub in a pulsing motion synchronised with the haptic feedback. The simulated blood flow relies on two visual components, a particle system of up to 2000 particles flowing from the needle hub at any one time, and a blood coloured texture map that is interactively updated as the particles fall onto the cloth.

Individual blood droplets are rendered quadrilaterals that always face upwards with a 16x16 pixel texture mapped to it. The texture has a bell shaped alpha distribution, so it is opaque at the centre and transparent at the corners. A blood droplet has two states, either airborne or inactive. A particle is airborne as it emerges from the needle hub and whilst it is falling towards and onto the surgical sheet. As the particle collides or falls through the sheet, it becomes inactive marking a red droplet onto a blood texture map.

As airborne particles are created, they are given a velocity, position, an elongation factor and an opacity value. While the needle tip is within the simulated femoral

Figure 4-14 Left: Real world blood flow from an IR needle hub. Right: Virtual blood flow.

artery, new blood particles are generated at fixed time steps. The number of particles generated each cycle varies, dependent upon the current femoral pressure as felt at the user's fingertips (tactile device, see Section 5) at that specific time. The higher the simulated blood pressure in the femoral artery (high pressure is felt as a pulse at the fingertips), the greater the number of particles that are created at the given time step. This is determined by the pressure currently felt at the pulsing tactile pad, see chapter 5.2.4. The velocity of a particle is also set relative to this femoral pressure and as such, the higher the femoral pressure, the further the blood will travel. A particle's position of origin is based upon the position of the needle hub at the time the particle is generated. The quadrilateral to which each blood texture is mapped is also elongated dependent upon the velocity of the particle as it exits the needle. On generation, all of the described particle attributes vary slightly by a factor of a small randomly generated number. This makes the trajectory, velocity and appearance of each particle slightly different and adds realism to the behaviour of the particle system which would be complex to recreate accurately.

The velocity of an active particle is updated at each time step. The velocity of the particle is updated to account for drag and gravitational forces and the particle's new position is calculated from its old position, the new velocity and a known time step. Each particle's position is checked to calculate if it has fallen onto or through the surgical drape. These particles are terminated as they have fallen from view but before this occurs, the position at which they fell onto or through the sheet is calculated. At the corresponding position, a single point is placed at that position in a blood texture map. At the end of each update cycle, a Gaussian filtered version of the hit texture is produced. This is superimposed on top of the surgical drape texture allowing a patch of blood to be seen where the blood drops have fallen. The blood texture is semi-opaque and so appears to stain the sheet red without obscuring it. Rendering the fallen blood particles in this manner reduces the computational load, in comparison to keeping a large number of fallen particles active on the sheets surface.

4.4 Visual Feedback Summary

The most popular method of virtual medical training visualisation for collocated visio-haptic simulation has been to use an immersive stereo display. This display offers an affordable collocation method that can be used in combination with haptic hardware, although the displayed image partially occludes the user's hands, therefore lowering the sense of simulation immersion that can be achieved. An alternative to this display is the see-through HMD, which is expensive, only offers a relatively low resolution visualisation, has a narrow field of view and requires precise (expensive) tracking. The foundations of an AR visualisation method have been developed here for integration into an AR immersive workbench. This workbench, integrated with haptic feedback in Chapter 7, overcomes the occlusion problems inherent whilst using the immersive stereo display with haptics, whilst also retaining its usability.

Bullet physics' soft body library has been used to visually deform skin, whilst a separate deformation calculation is required for haptics rendering. The force calculations are described in Chapters 6 and the data exchange between the visual and haptics loop is described in Chapter 7, along with the position information for each visual element. The next chapter describes the development of a tactile feedback medium for palpation that can be obscured from view during simulation using the chroma-key techniques described here.

5 Tactile Feedback

5.1 Introduction

Tactile cues are felt from receptors within or close to the skin's surface, allowing a person to detect if a surface is smooth or rough, hot or cold, as well as conveying pain and information about surface vibrations. Chapter 2 highlighted how tactile technology is little used in medical simulation as the tactile cues felt as during a medical procedure are little understood and the technology to reproduce the tactile sense is in its infancy. During a palpation, both force/torque feedback cues must be used in combination with the fine tactile force felt at the practitioner's fingertips to guide them to perform a femoral needle insertion correctly. During an IR procedure, up to three fingers are typically used to depress a patient's tissue towards their femoral artery, as seen in Figure 5-1. Through a surgical glove, the temperature of the patient and a varied distribution of pressure around the fingertips that relates to the various tissues below the skin can be felt. There is also a time varying tactile cue that can be felt as blood is pumped around the vascular system by the patient's heart. This pressure fluctuation is not uniform between patients as pathologies within the vessel and the patient's medical condition will greatly affect the strength of the pulse felt. The pulse of a larger patient is also more difficult to locate as the fatty tissues must be displaced to reach the femoral artery. Technologies that can produce this tactile variation are investigated in this chapter. The temperature cue of a patient has been ignored as this cue was not judged to be critical during the task analysis [182]. It is therefore assumed its absence will not influence the ability of the simulation to train the critical cues, but will reduce the simulation cost significantly. Future work may investigate the use of temperature to determine if this assumption holds true.

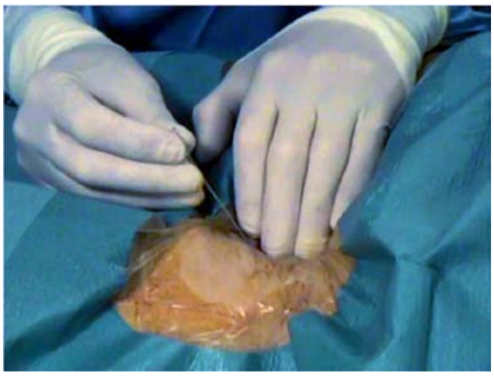


Figure 5-1 A three finger palpation performed by an interventional radiologist to locate the femoral artery before a needle is inserted.

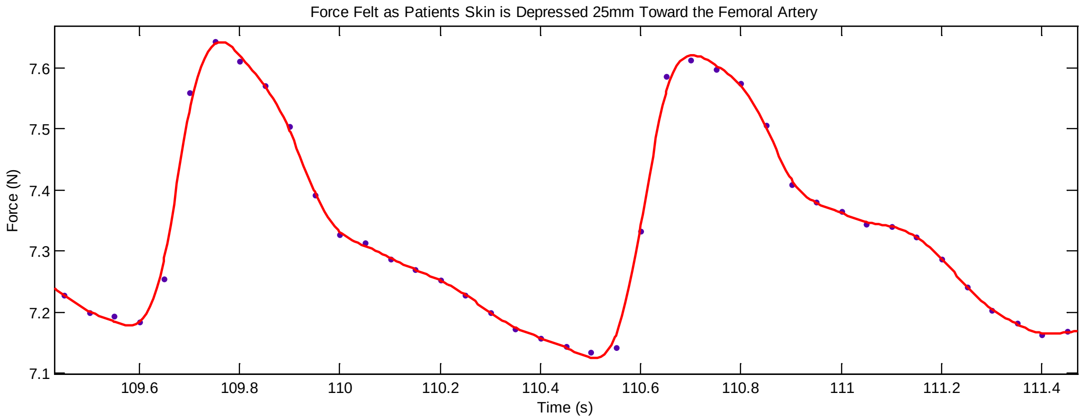


Figure 5-2 A 2 second sample of *in vivo* pulse force measurements at skin displacement toward the femoral artery of 25mm, an average applied force of 7.265N (over a complete measurement) and an average pulse fluctuation of 0.483N

The tactile sensation felt as a palpated femoral artery pulses was recorded *in vivo* by Liverpool University [183], in a set of measurements recording the force felt in relation to various skin surface displacements toward a patient's femoral artery. Two important features of a femoral palpation can be observed in Figure 5-2. Firstly, the magnitude of the highest tactile pressure change recorded *in vivo* for a particular patient. At a skin depression of 25mm, the pulse force felt at the fingertips fluctuated on average by 0.483N per pulse over a 10 second sample. This fluctuation occurred as the practitioner applied, on average, a deformation force of 7.265N. As such, the tactile device must not only be capable of producing this tactile fluctuation, but must do so under a constant pressure of the palpation force. The second noteworthy feature of this tactile waveform is the pulse's profile. A tactile device which aims to recreate the tactile sensation felt as the artery pulses should be able to sufficiently recreate this pulse profile, although a lack of understanding of the tactile sense throws doubt on how accurately the feedback must be reproduced.

The goal of this research is to provide tactile feedback that recreates the small force fluctuations felt at the user's fingertips that do not act to move or resist the finger itself (force feedback). Dependent upon the technology used, this may reproduce realistic skin stretch as a by-product, enhancing the tactile sensation felt, although this is not measured.

5.2 Technology Development

Three novel developed tactile solutions and one commercial tactile device have been evaluated for their ability to provide pulse like tactile feedback. Initially, three tactile solutions: piezoelectric pads, micro speakers and a pin array device underwent an evaluation with an expert interventional radiologist before a fourth tactile solution, hydraulic actuation, was proposed and tested more extensively.

5.2.1 Piezoelectric Pads

When a voltage is applied to a piezoelectric material, it expands and contracts dependent upon the voltage potential applied. A finger pad assembly, which could be held or mounted on a surface, was developed by carefully removing the thin (less than 1mm) circular piezoelectric actuators from audio devices. The pads are driven by an integrated circuit board consisting of a microcontroller and OLED graphics display made by Luminary Micro (Austin, USA) and, signal amplification circuitry, used to control the voltage output and resulting displacement of the pads, see Figure 3-2. When driven at low frequencies (1Hz - 3Hz) [192] [193], not typically used in applications of piezoelectric material, these pads created a pulsing effect that, when pinched between thumb and forefinger, created a distinct feeling closely resembling that of a pulse. The pulse length and pause time can be manipulated to simulate zero to 300 pulses per minute using two of the four buttons on the integrated circuit board. The second set of buttons can be

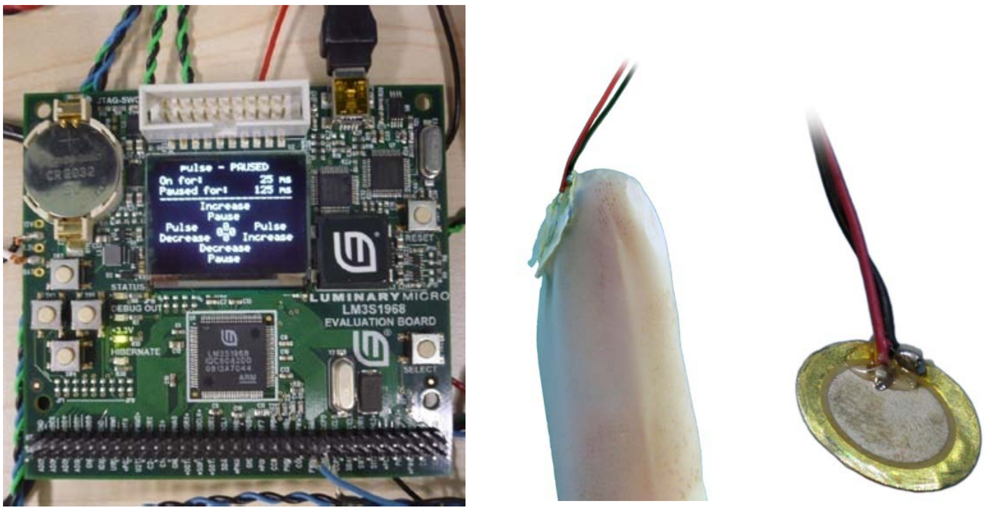


Figure 5-3 Left: An integrated circuit board Luminary Micro used to control the pulse displacements of the piezoelectric and micro speaker hardware. Centre: A piezoelectric pad fitted to a surgical glove. Right: A piezoelectric pad removed from its audio housing.

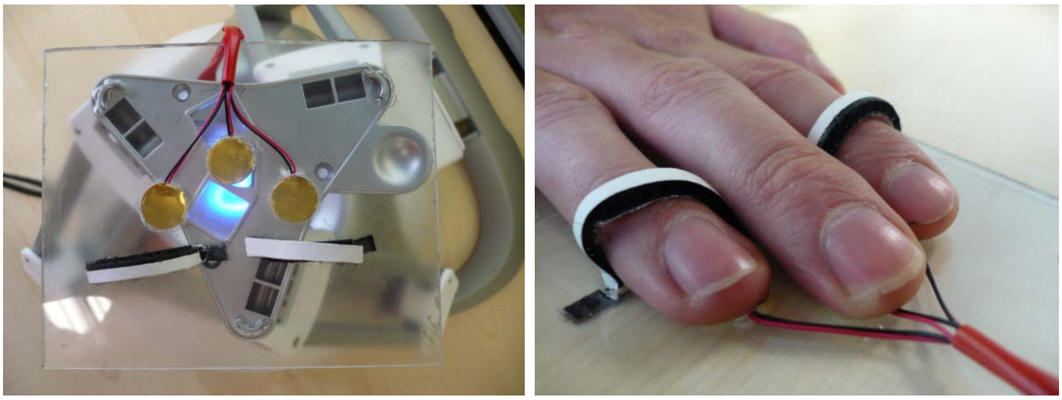


Figure 5-4 Left: Piezoelectric pads mounded onto a customised Falcon end effector. Right: The position of a user's fingertips on top of the modified end effector.

used to vary the pulse profile. Only the square voltage profile (an on/off signal) produced satisfactory tactile feedback with a sinusoidal voltage change imperceptible to the touch. A square voltage change does not produce a square displacement profile as the pad displacement does not change instantaneously, creating a more sinusoidal displacement. As the voltage input was increased, the strength of the pulse felt increased logarithmically, with voltages over 25Volts producing very little increase in pulse strength but increasing the speed of material degradation.

Although pinching the pads between thumb and forefinger created a sensation close to that of a pulse, to be used in combination with a force feedback device, the pads needed to be mounted on an end effector. A customised end effector with three finger pads was developed for the Novint Falcon (see Figure 5-4) but, unfortunately, this arrangement greatly reduced the fidelity of the tactile feedback. Another prototype configuration mounted the tactile pads into a surgical glove interface, but this also reduced the tactile fidelity, see Figure 3-2. Although the low cost of this technology is appealing, the pads could not be successfully combined with a force feedback device. Piezoelectric actuators with a larger displacement could be used to increase the medium's tactile strength when mounted upon a force feedback device, but these actuators are expensive, therefore negating the medium's appeal.

5.2.2 Micro Speakers

Micro audio speakers, similar to those found in laptops, were also tested as potential pulse generation modules. The speakers are driven by small electromagnetic coils and can be activated using similar amplification circuitry and the same integrated circuit board as that used to pulse the piezoelectric pads. Driving the speakers with a square (on/off) signal to displace the voice coil was also found to produce a pulse-like tactile sensation as the cone's membrane was touched. The speakers produce a larger displacement than the piezoelectric pads, but with only a small reaction force at equal voltages that can be easily counteracted when applying a downward force through the device. To effectively feel a pulsing sensation, an alternative device arrangement to that of the piezoelectric pads was investigated.

By strapping the membrane of the speaker to the user's fingertip and allowing the mass of the speaker's base (approximately 7.5 grams) to move as the speaker is driven, the inertia involved in the movement produces a tactile response at the fingertip, which has been likened to a pulse, see Figure 5-5. The base of the speaker has a diameter of 19mm and is comfortable to wear, but as the device does not produce enough force to be effective when mounted onto a force feedback device, a requirement of this simulation, the device has been deemed unsuitable for this simulation application.

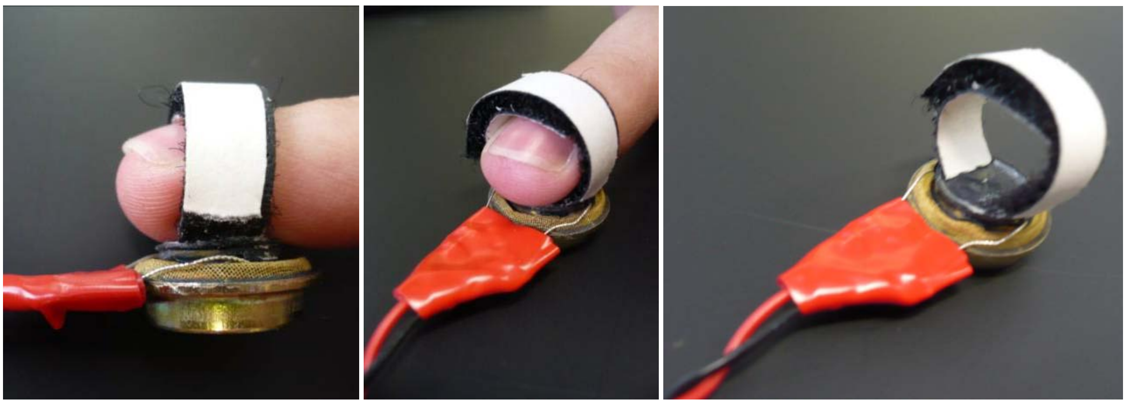


Figure 5-5 Finger mounted micro speaker tactile device.

5.2.3 Pin Array

A pin array device developed at Salford University [194] and further commercialised by Aesthesis (Salford, UK) calling it Aphee-4x, demonstrates an alternative promising concept for pulse simulation, see Figure 5-6. The device has 16 individual, remotely actuated pins in a 4 by 4 configuration, occupying an area of 7.25 x 7.25 mm, see Figure 5-7. The array is small enough to be clipped onto the tip of a user's finger if necessary and each pin can exert a force up to 1.3N with a maximum displacement of up to 2mm, achievable in 170 steps, see Figure 5-7. Although a single pin doesn't provide a force that can overcome that of a palpation, the user's fingertip is supported by the array's base and the pins (1.45mm in diameter) apply pressure to a very small area of the fingertip.

A C++ serial driver and C# control program was written for this USB device, see Figure 5-7. The control program allows the user to control the frequency of pulses, the displacement of pulsing pins, the number and arrangement of active pins, the pulse profile of the pins and the pulse profile's granularity. A pulse profile can be defined by manipulating a cardinal spline curve using a mouse to move the curve's control points. The control points of a cardinal curve lay on the curve itself, as it is composed of Bezier splines joined with C^1 continuity. This makes the curve manipulation interface relatively intuitive to use. The device's high force output and the large pin displacement make this device suitable for pulse simulation. The device and control program were verbally evaluated as offering a realistic tactile feedback by IR experts. Pulse simulation using only the 4 central pins appeared to produce the most realistic feedback. However, the technology is costly, at around

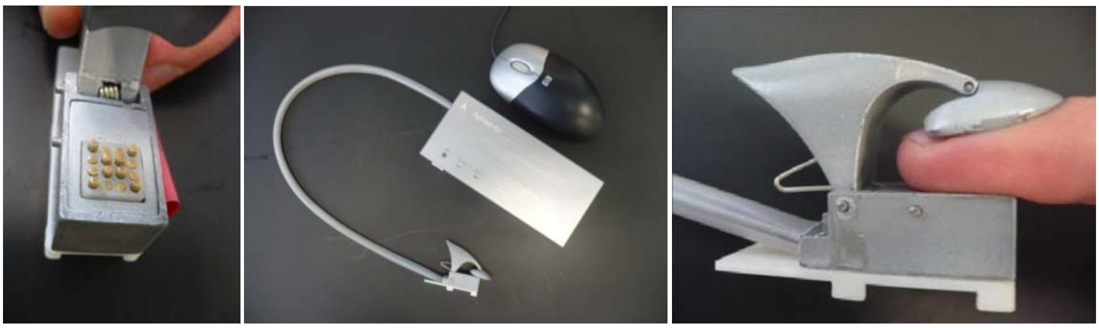


Figure 5-6 Finger mountable pin array device developed at Salford University [194].

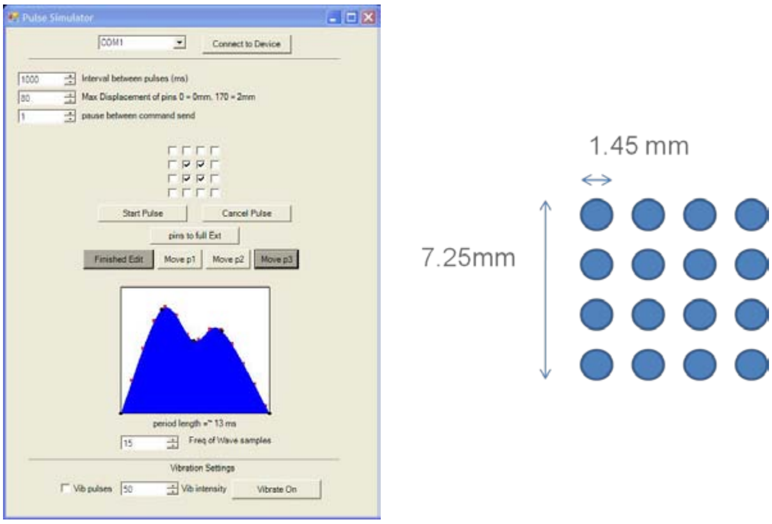


Figure 5-7 Left: Visual C# interface for the control of the Aphee-4X. Right: Pin dimensions and layout of the Aphee-4X pin array.

£10,000 (GBP) per device. As the device offered high fidelity feedback, ideas such as creating a version of the technology which used a single, wider diameter pin were proposed. This was not pursued as a three fingered device would have still been relatively expensive to produce.

5.2.4 Hydraulic actuation

To overcome problems with the previous tactile approaches, a fourth hydraulics-based technology has been investigated. Researchers at the Royal Liverpool University Hospital developed a hydraulics based tactile feedback device (called SimPulse) for use in a mannequin. Taking inspiration from this device, a thermoplastic tray structure containing a water actuated tube that is mounted in silicon has been developed. This rigid tray can be firmly fixed to force feedback hardware, allowing simulation of both force and tactile feedback. The commissioned tray structure was designed in Pro Engineer from PTC (Needham, USA) and manufactured on a Dimension Elite 3D plastic printer from Stratasys (Eden Prairie, USA).

The profile of the tray structure (Figure 5-10) is designed to optimise weight and tactile feedback. A 660 by 860 mm rectangular area that matches the dimensions of the palpable skin is covered in a 5mm thin layer of silicone. This malleable surface provides the feeling of touching soft skin. The user cannot feel the hard

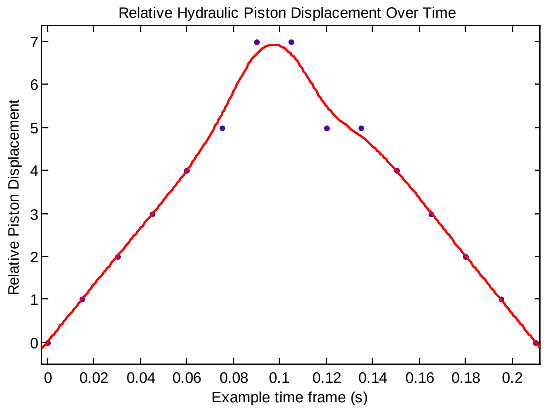


Figure 5-8 An example of the piston displacement used to produce a realistic pulsing sensation at the tactile end effector.

surface below the thin silicone layer, as when a force large enough for the user's fingertips to penetrate down to the hard plastic surface below it is applied, the force feedback device onto which the end effector is placed moves downward instead. The space underneath this thin surface is hollow to reduce the weight of the end effector. The low weight requirement is explained in Chapter 6 with an explanation of the force feedback calculations. The silicone filled area deepens along the tray's centre to a depth of 13mm to accommodate for a hydraulically actuated silicone tube. This tube is sealed at one end and attached via a connecting tube to a piston that is driven by a servo motor.

The actuator driving the hydraulic pulse is a HiTEC HS-7985MG high torque servo motor. Using a USB "Phidget Advanced Servo" controller from Phidgets (Calgary, Canada), the rotation of the servo's armature can be controlled to rotate up to 40 degrees, which in turn drives a piston, see Figure 5-9. Figure 5-8

Figure 5-9 Left: First tactile prototype with rotating base. Right: The sound insulated servo driven hydraulic actuation unit. The device is hung underneath the force feedback hardware to eliminate audible vibrations.

Figure 5-10 Left: The left hand side of the tactile end effectors silicone has been removed to reveal the simulated femoral artery. Right: The profile of the tactile end effector can be seen in the removed silicone section. A simulated femoral artery has been inserted to illustrate its path through the silicone. The ridged profile of the silicone secures it to the plastic base during use.

illustrates the displacement of the piston, which controls the pressure within the hydraulic line. As the displacement increases, the pressure in the simulated femoral artery also increases, which in turn forces the artery to expand. The strength of the pulse can be varied by changing the range of the servo's movement. The pulse profile is not as rounded as that seen in the *in vivo* recordings, Figure 5-2. This is intentional as the *in vivo* waveform was smoothed before it was received and it is assumed that an unavoidable amount of compliance in the system will result in a more rounded tactile output at the end effector in comparison to that produced at the piston. To eliminate the audible output of the hydraulic servo, the servo, pump assembly and controller are contained within a sound insulated box that effectively eliminates audible vibrations when the box is hung below the force feedback device. Using water as the hydraulic fluid provides a clean solution in the unlikely event of a leak.

Three prototype tactile end effector profiles were manufactured in an iterative development cycle. In the first prototype (seen in Figure 5-9), the hydraulically actuated tube ran horizontally from right to left along the length of the silicon filled trough. The tray was mounted on top of a bearing and spring that simulated tissue compliance as a trainee applied torque to the simulated skin surface. This allowed the tray structure to rotate up to 30 degrees, but, although this increased rotation capability provided a desirable feel of tissue compliance,

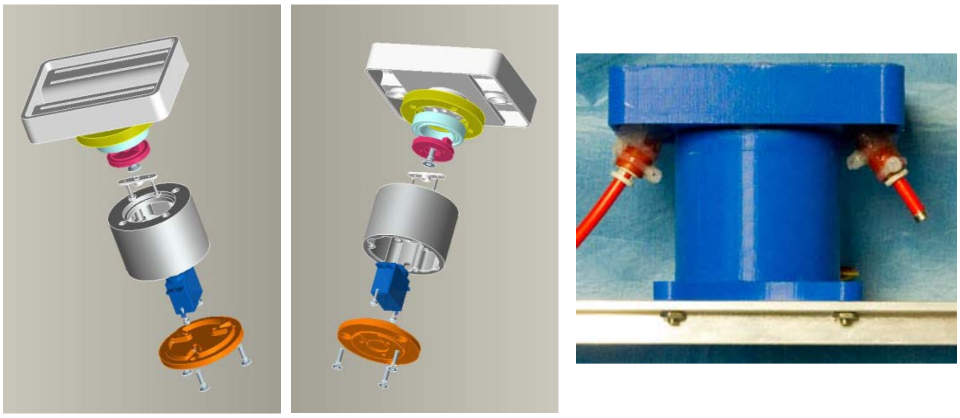


Figure 5-11 Left: Final CAD diagrams of the tactile end effector and rotating base. Right: Final tactile end effector mounted upon a force feedback device.

the rotation was untracked and the torque stiffness could not be modulated.

It is necessary to know the location of the simulated artery in real and virtual world space during simulation so that the tactile feedback can be accurately controlled at all times. As such, the rotation capability of the first end effector was removed in prototype two, lowering its weight from 140 grams to 70 grams. The profile of the simulated artery was also modified. In prototype one, the simulated artery ran the length of the silicon pad producing a long palpable uniform pulse, unlike that felt during a real palpation and encouraging users to palpate up to the edges of the end effector until they felt the hard edge of the device. To more closely represent the biological structure of the femoral artery, the hydraulically actuated tube was repositioned to arch up from the bottom of the silicone trough and back down again, this can be seen in Figure 5-10. A third modification was made to the tray structure by mounting it on top of a HiTEC HS-5055MG micro servo motor encased within a cylindrical housing and bearing structure, see Figure 5-11. Although this increased the weight of the end effector to 88 grams, the end effector can be accurately rotated around its centre point to a known orientation before a simulation commences. This increases the hardware's ability to accommodate for patient variability when the device is coupled with the force feedback hardware described in chapter 6.

The silicone compound was carefully selected from a range of silicone formulae to have a skin-like softness. From the range, Eco-Flex 0030 silicone from Smooth-

On (Easton, Pennsylvania) was chosen, a silicone used to create prosthetics for films. The two part silicone mix has a Shore 00-30 hardness rating and was mixed in a 1:1 ratio using weighing scales. Before filling an end effector, all holes must be carefully sealed. Where the silicone tube penetrated through the end effector's base, the tube was securely glued to the sides of the entry conduit. As the silicone is poured into the end effector, air bubbles are removed using a vacuum, ensuring the silicone will deform uniformly when the mix is dry and that the end effector's surface is smooth. The end effector is both a mould and a final product. The ridges on the bottom of the end effector mould help to secure the silicone in place whilst it is used, see Figure 5-10.

This end effector not only recreates the tactile actuation of a femoral artery, but also the surface deformation of a patient's tissue surface as the user's fingertips slightly penetrate the silicone, providing high tactile fidelity for the simulation of femoral palpation. The current rapid prototype production method is only cost viable for a small run of products, but the hardware could be mass produced at low cost through injection moulding.

5.3 Tactile summary

Although the complex interaction between a practitioner's palpating fingertip and the patient's tissue cannot be computed to a high level of realism in real time, an approximation of this interaction appears to sufficiently recreate the sensation for use in a training simulation. Validation results suggesting this are shown in chapter 8.

Table 2 presents the advantages and disadvantages of the 4 reviewed technologies. Of these, piezoelectric pads were deemed unsuitable for use in a femoral palpation simulation as, although they produced a compelling feeling whilst pinched, when mounted upon a force feedback end effector, their small displacement was hard to feel. Piezoelectric actuators offering larger displacements were deemed to be too expensive for use in PalpSim as one of the requirements is to produce an affordable simulation. Micro speakers did not provide a high enough force output to be usable with force feedback hardware, and the commercially available pin array device, although providing high fidelity feedback, was also deemed too expensive for inclusion into the palpation simulation.

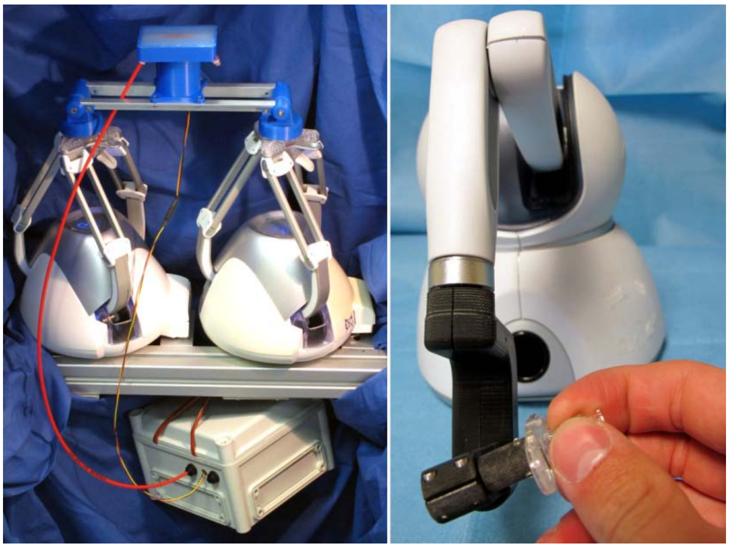
Technology	Advantages	Disadvantages
Piezoelectric Pads	Extremely low cost Thin Medium force	Small displacement Limited life
Micro Speakers	Large displacement Low cost	Limited force
Pin Array	High force Large displacement	High Cost
Hydraulic Actuation	High force Soft to touch Low cost Durable	Potentially Noisy

Table 2 A comparison of tactile technologies reviewed for their effectiveness in a femoral palpation simulation.

Hydraulic actuation, driven by a servo motor, offered high fidelity feedback, which can be effectively mounted onto force feedback hardware and can be produced with low cost components. As such, this feedback has been integrated into the PalpSim simulation (section 7) and validated with positive results (section 8).

In the next chapter, the combination of the hydraulic actuation device with force feedback hardware is described, with the aim of further increasing the fidelity of the palpation simulation.

6 Force Feedback



6.1 Force Feedback

As a practitioner palpates a patient's femoral artery, a resistive force is felt that opposes the motion of the practitioner's fingertips as they displace the patient's tissue beneath. The depth of displacement that is necessary to locate the artery and the strength of the resistive force felt varies with a patient's habitus. If carefully designed, computer controlled force feedback devices can simulate this varying ratio between force and displacement to provide a simulation capable of recreating the forces felt whilst palpating a number of different patient habitus'. As a patient's habitus changes, the forces felt during a femoral needle insertion will also differ and as such, a virtual palpation and needle insertion simulation solution requires both the palpation and needle insertion to be simulated using force feedback hardware if a patient variable procedure is to be correctly recreated for virtual training purposes.

The development of force feedback hardware and algorithms to control the forces felt during palpation and needle insertion simulation are described in subsections 6.2 and 6.3 respectively.

6.2 Virtual palpation

It has been established through video task analysis and discussion with experts that a practitioner typically palpates the femoral artery using three fingers, feeling an individual force at each finger as the tissue is depressed. An ideal haptics device for virtual palpation simulation should not only be inexpensive but capable of accurately recreating a three finger palpation of an arbitrary skin surface. If the fine distribution of forces across a single palpating fingertip is simulated as tactile feedback, the force felt at the tactile end effector can be accurately reproduced with an expensive 6 force DOF device and a tactile interface. However, a far more cost effective approach is used in this simulation, in which the haptic feedback felt at all three fingertips is approximated using a single 5 force DOF force feedback device combined with the hydraulic tactile end effector as described in Chapter 5.

An end effector requirement specified in Chapter 5 was low weight. This is in part due to the limitations of affordable force feedback technology. During a palpation of a patient lying on their back on top of an operating table, the primary direction of resistive force as the skin's surface is deformed is upward. In PalpSim, the hydraulic tactile end effector must be held stationary at the skin's surface until the user palpates the virtual patient, continually exploiting a proportion of the device's force capabilities to do so. Every force feedback device has a limited force output and increased end effector weight will reduce the net force that can be used to resist the motion of the palpating fingertips. Although an ideal force feedback device would offer unlimited force at low cost, in reality, devices offering high forces such as Force Dimension Delta 6 force DOF device, providing up to 20N of force, are expensive (approx £36,000). Chapter 2 discussed how high simulation costs limit the number of institutions willing to pay for computer enhanced training, and consequently there is a requirement to keep simulation costs to a minimum.

In addition to reducing the net force output of the device, a device with increased weight will require a greater force input to counter the device's inertia. This will reduce the device's capability to accurately reproduce force/torque feedback as devices without force/torque sensors (all common commercial devices) rely on a change in end effector position to infer the need for a change in force feedback, unless an end effector is to be guided by a force vector. An end effector with a large mass requires a greater force input to change its motion state and position in comparison to a lighter one.

6.2.1 Haptic Rendering

After the initial finite element simulation (FEM) approach (proposed in Chapter 4) was deemed to be infeasible at haptic rates, requiring FEM force calculations to be made in approximately 1 millisecond, the prototype rigid body simulation developed using the Chai3D (version 1.5.1) haptic development library was redesigned from the bottom up. The cross platform, multi device support provided by Chai3D was utilised to produce a common haptic hardware interface, allowing multiple devices to be used during testing of the simulation environment.

A gravity compensation feature was added to this interface by re-implementing the Libralis' [195] online gravity compensation feature. This library, designed by the SIRSLab in Siena, Italy, was designed to provide gravity compensation for devices using the open source Haptik [50] force feedback library. This gravity compensation procedure first calculates an estimate of the gravity compensation required at a series of positions within the device's workspace using the Libralis Tuner software offline. A 3D matrix of compensation factors for each position is subsequently saved. This matrix is loaded and used online during each haptic feedback calculation. A tri-linear interpolation between the current position of the device and the pre-calculated compensation points is performed during each haptic cycle to produce a gravity compensation estimation that is applied in addition to any force value to be simulated. This compensation procedure has been successfully implemented. However, Novint have subsequently released gravity compensation support for the Falcon through their F-Gen game interface, and, in July 2010, after the implementation of the previously described gravity compensation procedure, Shah *et al.* [196] released a set of open source files, extending the Chai3D 2.0 package, that provide gravity compensation for a dual Falcon setup.

The initial PalpSim prototype simulation used a modified Falcon force feedback device that had been rotated through 90 degrees such that it faced upwards, see

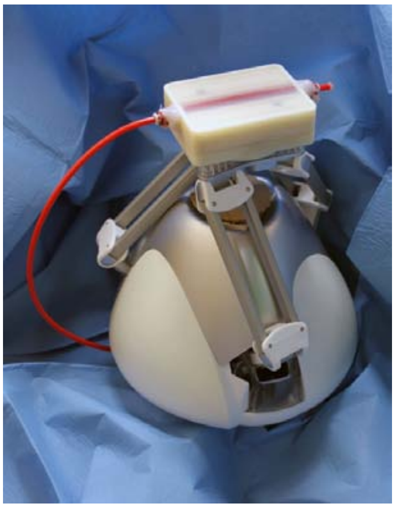


Figure 6-1 Modified Novint Falcon force feedback device. The Falcon has been rotated through 90 degrees and mounted with the first prototype of the hydraulically actuated tactile end effector.

Figure 6-1. As a Falcon will only work if an end effector containing the correct electrical components are attached, the components of an original grip have been wired into the device's base. Instructions how to do this are given in Appendix 10.2.

This force feedback platform and a prototype tactile end effector was then used in a deformable simulation controlled by the Bullet Physics soft body library, a position based solver, for both visual and force feedback. In a constraint-based approach to force feedback like that first proposed by Zilles and Salisbury [198], the haptic end effector is coupled to a virtual proxy, a rigid body object in the physics library, using a stiff spring coupling. However, the virtual proxy's finite size makes this approach similar to that of Ruspini et al [197], as Zilles and Sailsbury's used a point-sized proxy that may fall through any small gaps in the mesh and would not produce such a visually pleasing deformation shape. Figure 6-2 illustrates how the proxy object remains on the object's surface as the physical

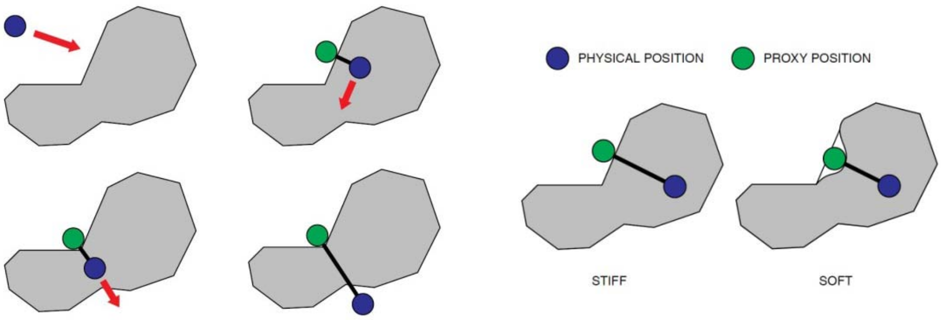


Figure 6-2 The motion of the virtual proxy, as the haptics devices position is altered. As pressure is applied, the surface of a soft body will indent, resulting in the haptics device penetrating the volume of the simulated patient. Images taken from [197]

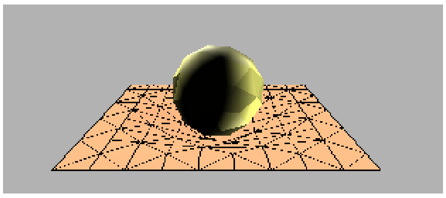


Figure 6-3 Deformable mesh with 10x10 nodes and 162 faces simulated using Bullet Physics' soft body library. Deforming object is a rigid body object controlled by a force feedback device.

haptics device penetrates the object. The greater the penetration, the greater the resultant force produced from the stiff spring constraint. As the haptics device penetrates the deformable object, the proxy object applies pressure to the mesh surface, which stretches around it. The force produced by the spring constraint is conveyed to the user through the force feedback device. Bullet Physics' position based solver, designed to produce visually pleasing deformations of relatively large meshes at graphics update rates of approximately 15ms per scene, was used to deform a low resolution mesh (Figure 6-3) at much faster rates. This approach was optimised to offer feedback at a raw 300Hz and interpolated to provide haptic feedback of 600Hz using rate of change information. In this approach, the force felt was dependent upon the interactions between the proxy object and the mesh nodes. Although the deformable variables of the mesh could be tuned to match that of *in vivo* force measurements, the resultant force is resolved from the interaction of multiple contact points between the mesh and the spherical proxy object. This makes it difficult to interchange *in vivo* measured data to simulate a new patient, where a particular advantage of virtual medical training is that simulation of patient variability should be simple to achieve.

To simplify the integration of a potential atlas of force feedback profiles that could be acquired from *in vivo* force measurements, the haptic and visual feedback calculations were separated. In this new model, visual skin deformations are calculated using Bullet Physics' position based solver to simulate skin deformed by an invisible virtual proxy object controlled as in the approach described above at visual update rates (a description of this was given in Chapter 4). The now separate haptic feedback loop is calculated using a force deformation formula pre-calculated offline from *in vivo* force measurements and used at a haptic feedback rate of 1000Hz to produce smooth realistic feeling feedback.

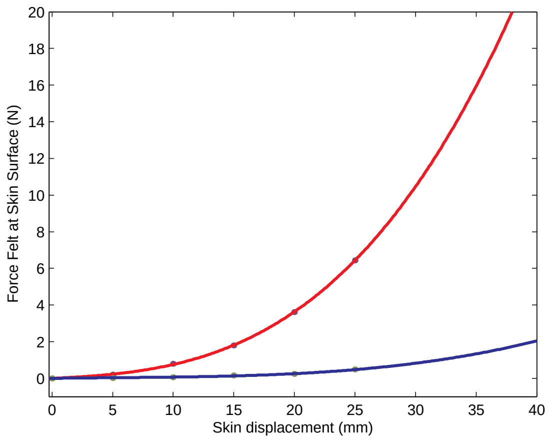
6.2.2 *In Vivo* Palpation Force Calculations

The interpretation of the *in vivo* palpation force measurements, first described in Chapter 3.4, to produce a force profile that can be used during simulation is described here. These measurements encapsulate the palpation force felt at the surface of the skin as it is compressed a known distance towards the femoral artery over a period of 10 seconds per displacement. Three separate variables can be extracted from this data:

- The depth of skin surface compression in mm, varying in depth z at 5 mm intervals between 0 to 25 mm at a single x, y position over the femoral artery.
- The average constant palpation force for this compression in Newtons (N)
- The maxima and minima of the force measurement over time summarised to a magnitude of variance from the average constant palpation force, describing the fluctuation of force due to the femoral artery pulsating.

From each ten second recording at 5, 10, 15, 20 and 25mm skin displacements toward the femoral artery, the average constant force and average variance from the force (pulse pressure felt) were plotted, see Figure 6-4. The best fit of a cubic polynomial curve (5.1) for each data set was then found using the Matlab (MathWorks, Cambridge, UK) software's curve fitting tool (cftool).

$$f(x) = ax^3 + bx^2 + cx + d \tag{5.1}$$



Displacement (mm)	Force Newton	Force Variance (N)
0	0	0
5	0.15	0.027
10	0.75	0.069
15	1.91	0.152
20	4.11	0.237
25	6.44	0.483

Figure 6-4 *In vivo* palpation force recorded on a thin healthy subject. Left: Plot of displacement of skin against average resulting force. Red- Palpation force. Blue- Tactile force variance caused by pulsing femoral artery. Projected from 25 mm to 38mm to avoid excessive force applied to patient. Right: Average forces measured at a given displacement during *in vivo* force measurements. Force measured over a 10 second period at each displacement.

In equation 5.1, the coefficients a, b, c and d of the fitted cubic polynomial curve (constants) are determined using the cftool. The known skin displacement, x, in mm (the position of the palpation end effector) is used during the haptic loop to compute f(x), describing the resistive force felt at the skin's surface resulting from deformation x. This force is then scaled dependent upon a scale factor determined during the force calibration of the haptics device, see Chapter 7.5, and displayed to the user to convey the compliance of a simulated patient's tissue.

6.2.3 Palpation Hardware Solution

The Falcon 3 force DOF device was chosen to be used in this palpation simulation for its mechanical configuration and affordability. When rotated through 90 degrees as seen in Figure 6-1, the triangular 3 DOF mount can securely support a custom palpation end effector and does not provide un-actuated rotations. Despite these advantages, the device only offers 3 force DOF where six is required to accurately reproduce the sensation of tissue deformation. The devices low cost manufacturing leads to undesirable compliance at the end effector, and the device force output is low when compared with various higher cost devices.

The Falcon is marketed as providing 9N of force feedback. In the reviewed set of *in vivo* force measurements (section 6.2.2), the deformation of a thin male patient's skin by 25mm towards the femoral artery asserts 6.44N of upward force at the practitioner's fingertip and the force is predicted to increase as the deformation deepens. Although the devices electric actuators can produce this force, a device rotated through 90 degrees must also support the weight of the tactile end effector and the device's three arms (linkages), in addition to applying a force to simulate tissue resistance. Neglecting the mass of the three device's metallic armatures and friction within the devices joints, it can be observed that, as the tactile end effector alone weighs 88 grams, so the device is required to produce 0.86N of force to provide gravity compensation for the end effector prior to asserting any additional force. This now makes the device only capable of asserting a maximum of 8.14N of simulated force. Furthermore, the Falcon cannot assert 9N throughout its whole workspace. The Falcon's highest force output can be achieved when all three arms push with maximum force, driving the end effector along a straight trajectory directly out from its base. Unfortunately, this only produces one degree of high force output that is not produced if the end effector deviates from this centre line [199]. If the end effector is moved off centre, the device's output capability rapidly reduces as the force is not shared equally between the linkages. The position where the device can only assert a weak force can be observed when only a single armature is producing the majority of the force feedback.

To overcome these two force limitations and provide additional force degrees of freedom, two Falcon devices have been rotated through 90 degrees and joined by a rigid link and two sets of dual revolute joints, replacing the original device's end effectors, see Figure 6-5. The link, that can be seen in more detail in Figure 6-6, is 230 mm in length and comprises of two lightweight aluminium bars, supporting the hydraulic tactile end effector and servo motor (see Chapter 5), mounted exactly half way between the two devices. Each rotating joint uses a high quality deep groove ball bearing that ensures a strong, low-friction rotation and

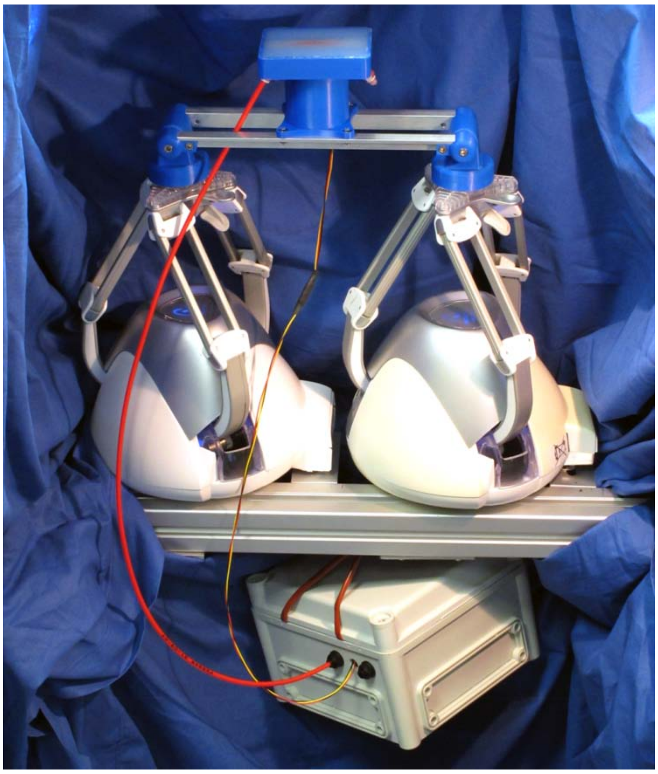


Figure 6-5 Modified Novint Falcon force feedback device for a pulse palpation simulation. The devices have been rotated through 90 degrees, coupled together and mounted with an additional hydraulically actuated tactile end effector, see Chapter 5.

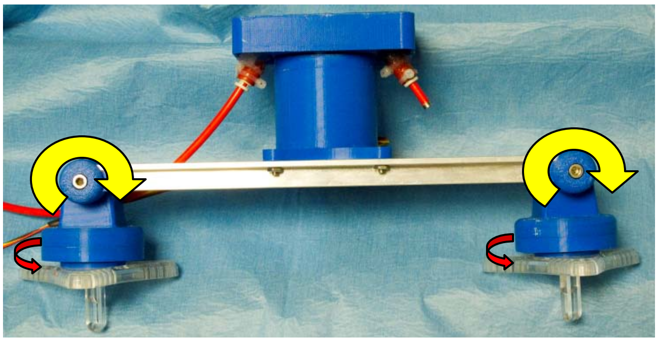
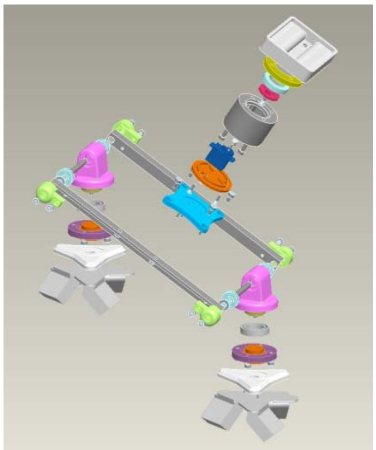



Figure 6-6 A modified end effector connects two Falcon devices to produce a custom 5 force DOF device. Two sets of dual revolute joints are used (one yellow rotation and one red rotation), connected to each Falcon via the clear plastic interfaces. The silicone tray tactile interface is mounted in the centre of the connecting bar.

eliminates undesired perpendicular rotations and lateral movement. This is advantageous as, when fixed to the two Falcon devices, the undesirable compliance that could be felt at the end effector of an individual device is greatly reduced. Researchers had predicted that this compliance could be a significant problem whilst using the Falcon for precise force feedback applications [199]. The linkage component adds 392grams across the two devices, requiring an additional 3.85N of gravity compensation, whilst increasing the force output to 14.15N (considering the added mass) and increasing the overall force DOF to five (Figure 6-7) in addition to stiffening the actuators. An additional, but redundant actuated degree of freedom, is produced by the servo motor mounted on the underside of the tactile pad (Chapter 5). This is used to increase the rotational freedom of the tactile pad so that the orientation of the femoral artery can be set before the simulated palpation occurs, but remains static throughout the simulation, increasing the range of simulation variability that can be introduced.

The position and orientation of the two Falcon devices in relation to one another is important, determining the achievable end effector workspace and the force output of the two combined devices. In this application, both devices are orientated in the same direction to maximise the available workspace of the single end effector. If one of the devices were to be rotated through 180 degrees about its vertical axis so the device still points upward but the leg configuration is mirrored,

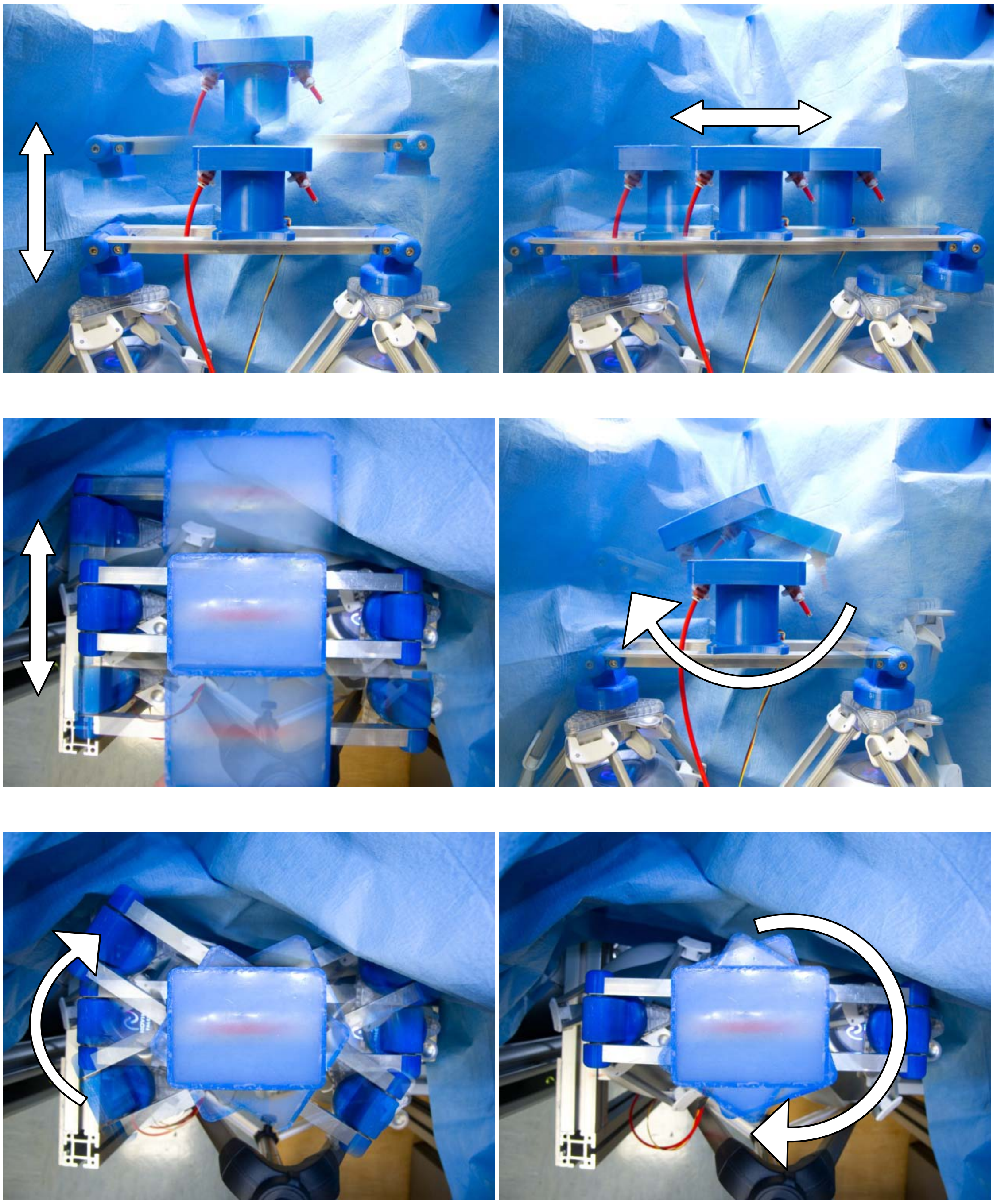


Figure 6-7 Multiple images are overlaid on top of each other to demonstrate the dual Novint Falcon's Five force DOF and the tactile end effector's one degree of freedom. Multiple images of the devices range of movement have been superimposed to demonstrate the possible range of motion. The device offers three force DOF translation and two torque DOF. A single redundant DOF is produced by a servo housed within the tactile end effector (bottom left).

the device's force output, as the palpation end effector is moved off centre, could be increased. However, this would limit the devices workspace and was not considered necessary for this application.

Although the device offers 5 force DOF, currently only 1 DOF force measurements have been recorded *in vivo*. To produce a realistic feeling palpation simulation, the subsequent 4 force DOF deformations felt at the end effector offer a feel of tissue compliance via a spring constant k, determined empirically to feel correct according to subject experts. This allows the end effector to rotate and translate under the spring system, whilst always returning the tactile end effector to its centralised position. This is demonstrated by the pseudo code in Figure 6-8.

```
//A 1DOF in vivo reaction force is calculated for the current end effector
position. a, b, c and d are coefficients of the offline fitted cubic
polynomial curve. x is the displacement of the palpation end effector in
relation to the skin surfaces un-deformed height
inVivoForce = ax^3 + bx^2 + cx + d

//Static X, Y values determining the optimal position of the two device's
end effectors in device space are defined. These are the devices x, y centre
position.
Define static P1_X, P1_Y, P2_X, P2_Y

//The in vivo force and end effector gravity compensation parameters are
summed and the total desired force is divided between the two devices.
forceVector(X_GraityComp/2, Y_GraityComp/2, (Z_GraityComp + inVivoForce)/2);

//The end effectors torsion compliance around the centre point, producing
flat rotation of the end effector on the x, y plane, and the 2DOF compliance
in the x, y plane (pushing and pulling toward the front and back of the
workspace and from left to right) result from the spring constants influence
to return the device to the P1_X, P1_Y, P2_X, P2_Y positions. An adjustment
factor ZCompensationFactor, provides a spring like compensation factor to
balance/correct the additional rotation DOF (a seesaw like motion).
k = Empirically determined stiffness constant
ZCompensationFactor = (Device1PositionZ – Device2PositionZ) * k;

3DOF_Force_Device1=
       forceVector->x - ((PX_1 - Device1PositionX) * k),
       forceVector->y - ((PY_1 - Device1PositionY) * k),
       forceVector->z - ZCompensationFactor;

3DOF_Force_Device2=
       forceVector->x - ((PX_2 - Device2PositionX) * k),
       forceVector->y - ((PY_2 - Device2PositionY) * k),
       forceVector->z + ZCompensationFactor;
```

Figure 6-8 Force feedback calculation within the haptic loop for the 5 Force DOF device.

6.3 Virtual Needle Insertion

Although a needle can be simulated in free space with a 6 DOF sensing device, when inserted into a patient's tissue, 6 degrees of force are also required to accurately recreate the forces felt. The literature review of current needle insertion simulations has shown that although not ideal, a 6 DOF device offering 3 force DOF can sufficiently recreate the sensation of needle insertion. The lowest cost device providing this level of feedback at sufficient fidelity and in a large enough workspace is SensAble's Omni, costing approximately ten times less than the lowest cost 6 force DOF device (suitable for this application), SensAble's Premium 1.5, whilst also capable of providing sufficient force feedback to the user. A sample *in vivo* force recording of a femoral artery puncture, as described in Chapter 3, can be seen in Figure 3-7. The peak force output of this needle insertion does not exceed 1.5N, although it should be noted that this is a single needle insertion performed correctly without excessive force applied.

However, the Omni's standard stylus effector neither looks like a real needle hub, providing visualisation problems when used in an AR environment, nor does its shaft (400% wider in diameter) provide the correct tactile cues as it is grasped between the trainee's fingertips. Therefore, this provides low face validity during a needle insertion simulation. To overcome this limitation, the Omni's end effector has been redesigned to use a real needle as the haptic interface. This modification is described in the following sub section.

During simulation, the force feedback hardware is gravity compensated, again using the re-implemented Libralis driver [195] so that the forces felt during

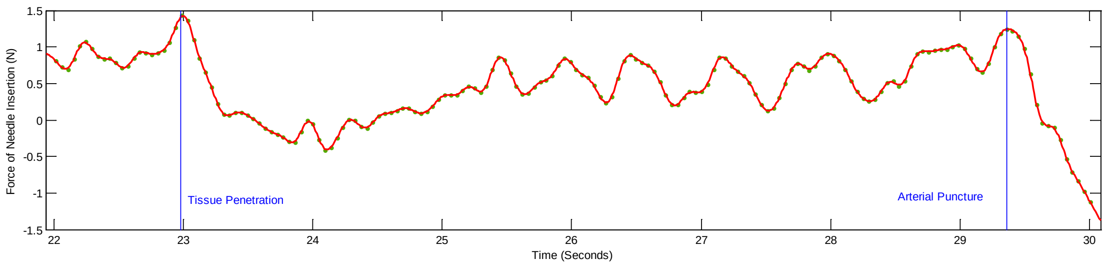


Figure 6-9 *In vivo* force measurements of a femoral needle puncture. Figure produced from unpublished *in vivo* force measurements kindly provided by Dr J Zhai and Dr T How.

simulation are not influenced by the mass of the device's armature. Forces have been developed with consideration of the *in vivo* measured forces seen in Figure 6-9, although they have not been directly calculated from real *in vivo* force data in the same way as the simulated palpation forces. This is seen as an area of future work with the main goal of including needle insertion within the simulation to prove that both direct patient contact and tool interaction can be well simulated in an AR medical simulation environment. An explanation of the implemented force calculations follows and a pseudo code version of these calculations can be found in Figure 6-10. These calculations are based upon prior work described in section 2.8.2.

The force threshold at which a real needle tip will penetrate a specific simulated patient's skin is determined from *in vivo* force measurement. In the particular patient observed in Figure 6-9 this is 1.5 N. The virtual needle tip, whose position is calculated from the known transformation matrix of the modified Omni needle hardware, held in the user's right hand (in this simulation scenario), can be felt contacting the skin as the tip is moved to a height at or slightly below the height of the palpation end effector palpated by the user's left hand. The haptic response of the needle puncture is split into two phases, one in which the needle tip has yet to penetrate the patients skin and a second in which the needle tip is within the tissue.

If the needle is not within the tissue, the forces felt are based on a proxy based force calculation like that seen in Figure 6-2. If however the user increases the pressure exerted on the needle hub above the predetermined skin puncture threshold, the needle will puncture through the surface. The second phase of force calculation is then initiated and the orientation of the virtual needle as it enters the skin is locked so that the tip cannot be unrealistically manipulated within the tissue. As it is not possible to recreate all of the 6 force DOF required to convey the feel of an in vivo needle puncture, see Figure 2-21, the virtual needle's motion is restricted, only allowing the needle to be advanced through the site on the patient's skin at which the virtual needle puncture occurred. The needle insertion force is approximated by simulating a resistance relative to the rate the needle is inserted / retracted through the patient's tissue, multiplied by a factor of the length of the needle shaft within the tissue, empirically determined to mimic the forces felt during a standard rate needle insertion, as seen in Figure 6-2.

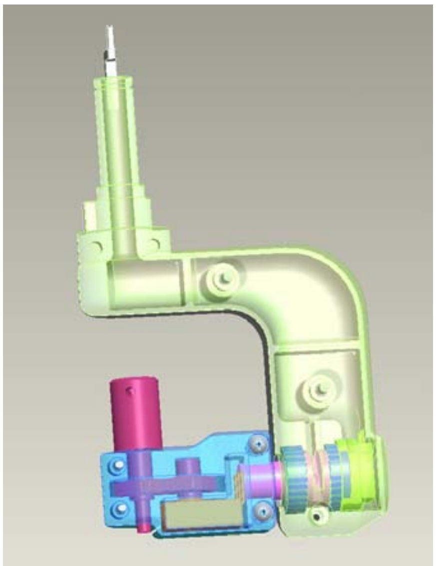
```
//Needle tip calculated by means of trigonometry, using the needle
hardware's position and rotation information, and the known shaft length.
needleTipX, needleTipY, needleTipZ

//The current skin surface height is determined to be at the height of the
palpation device. A deformation of the skin reduces the skins height.
skinHeight

//A force threshold at which the needle tip penetrates the skins surface is
determined from in vivo force data.
needlePunctThesh =  inVivoPuntureThreshold

//The first of two phases; Skin punctured / Not punctured. If the skin
surface has not been punctured, find the force that should be applied to the
needle using proxy based haptic interaction. If puncture threshold is
exceeded set SkinPunctured true. If not punctured apply proxy based force.
if NOT SkinPunctured
       Force = proxy based force feedback;
       if Force > needlePunctThesh
              Save SkinPiercePosition;
              SkinPunctured = True;
       else ForceOutput = Force;

//If the skin has been punctured
if SkinPunctured

       //ResistanceIncWithDepth influences (increases) the resistance felt
       as needle depth into the tissue increases.
       ChngInDisplacementFrmSkinSurf is the change in displacement between
       deformable skin surface and needle tip since last haptic calculation.
       zNeedleForce =
              ResistanceIncWithDepth * ChngInDisplacementFrmSkinSurf;

       //The current point the needle shaft penetrates the skins surface is
       found to calculate the difference between this and the point at which
       the shaft pierced the skin. The spring constant is varied dependent
       upon the depth of the needle penetration into the skin, quickly
       stiffening as depth increases
       xNeedleForce =
              CurrentPointOfShaftThroughSkinX - SkinPiercePositionX *
              LengthofShaftInserted;

       yNeedleForce =
              CurrentPointOfShaftThroughSkinY – SkinPiercePositionY *
              LengthofShaftInserted);

       ForceOutput(xNeedleForce, yNeedleForce, zNeedleForce);

       //A check is made to see if the needle has been retracted from the
       tissue
       if needleTipZ > skinHeight
              SkinPunctured = False;
```

Figure 6-10 Force feedback calculation within the haptic loop for the needle insertion procedure.

6.3.1 Needle Force Feedback Hardware

The Omni has been modified by removing its original stylus assembly and replacing it with a real needle hub and shaft (through which a guidewire can be passed when further steps of the procedure are simulated). It is thought that this will greatly increase the needle puncture simulation's tactile and AR visual fidelity. This reengineered section of the device, called the "wrist" from here on, controls the end effectors non-actuated roll, pitch and yaw with the remaining 3 force DOF actuated from within the device's base, controlling position. A needle insertion is a common medical procedure and it is thought that a force feedback device with a needle end effector fitted to it could be of benefit to needle insertion simulations other than PalpSim. To create a device which can be simply integrated into existing and new applications and to allow inter-changeability between modified and non-modified Omni devices, the kinematics of the new end effector are designed to match those of the existing device. The device can be used with SensAble's existing drivers.

The armature of the wrist section is arranged so that whilst using the augmented reality workbench, first shown in Chapter 4 and explained further in Chapter 7, both the needle hub and users fingers are not obscured from the top mounted



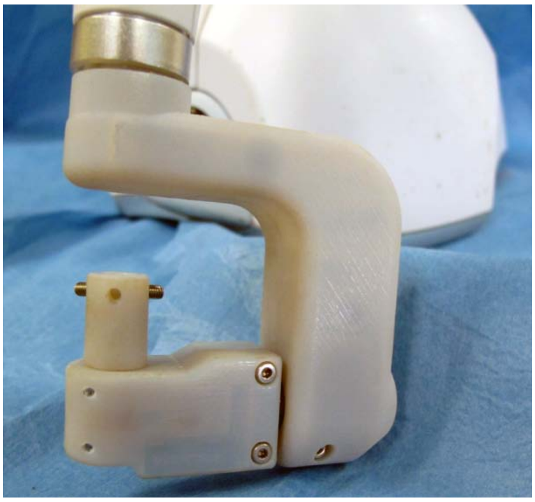


Figure 6-11 Left: A Pro Engineer model of the redesigned Omni wrist. Right: The manufactured wrist structure fitted to an Omni device.

camera lens's line of sight during normal usage.

The replacement wrist has been designed in Pro Engineer, Figure 6-11, and manufactured on two 3D plastic printers. Two printers were used to manufacture the device as the plastic properties of each differ. The lower resolution printer, the Dimension Elite printer from Stratasys (Eden Prairie, USA), is used to manufacture the two part armature of the modified component, as although this plastic does not have such a fine finish, a combination of its coarser, almost honeycombed printing structure, and its specific plastic composite produce a rigid plastic structure capable of conveying forces without deformation of the armature, see Figure 6-12. The higher quality printer, the Eden250 from Objet (Billerica, USA), produces a slightly softer, but finer product and so is used to manufacture the remaining small moving parts which require a fine tolerance to function correctly.

To perform this modification, two sets of covers on the Omni's second arm must be removed to reveal the shaft of the rotating Y shaped wrist, see Figure 6-13. The wires must then be cut to allow the wrist to be removed from the arm by sliding (with force) the arm out through two bearings which hold the rotating Y shape structure. The internals of the new wrist, Figure 6-12, can then be wired and the new structure slid into the bearings through which the original wrist was removed. The wrist connections must then be reconnected within the arm before the structure is closed.

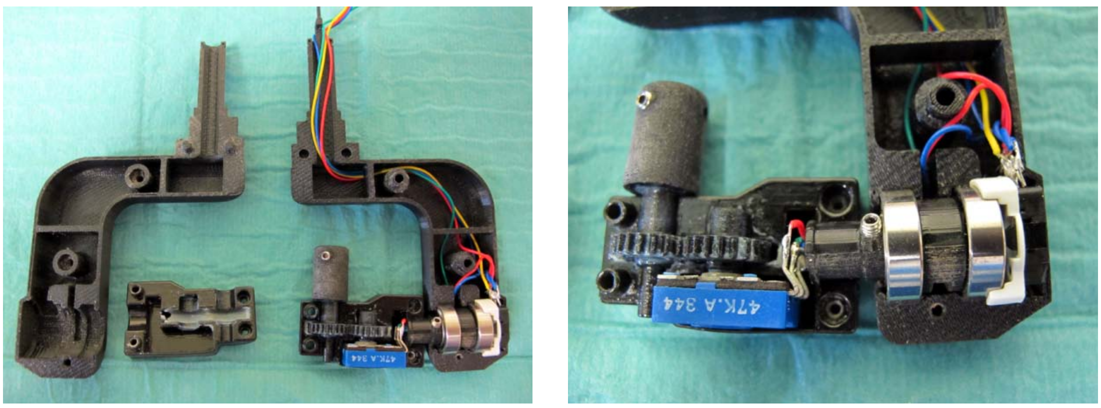


Figure 6-12 Left: Two halves of the modified wrist for the Omni force feedback device. Right: A close up image showing mechanics of the final two degrees of freedom. On the right of the image two sealed deep groove ball bearings can be seen on the end of which the Omni's original potentiometer is mounted. On the left of the image the blue off the shelf Tyco potentiometer can be seen, rotated via the dual cog assembly.

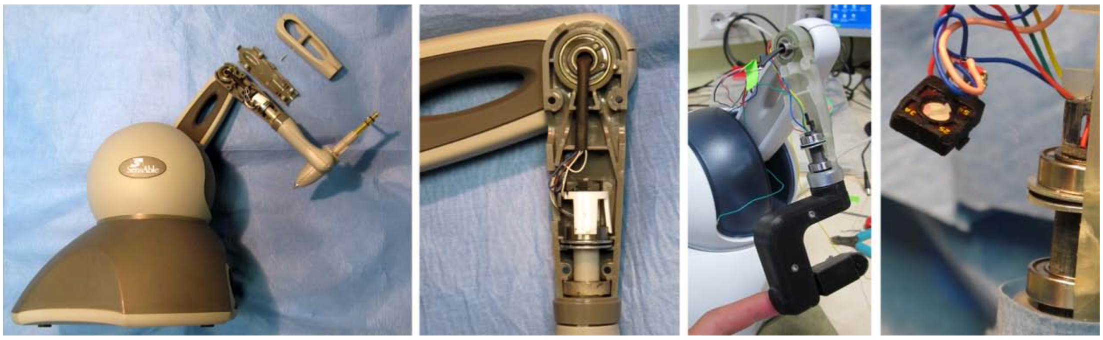


Figure 6-13 Left: The Omni's cover can be detached by removing a single screw on the arms underside. Four screws then need to be removed to uncover the internals of the arm, seen 2^{nd} from left. After sliding the arm out, the new arm can be inserted, 2^{nd} from right. Right: The Omni's potentiometers must be re-connected to the correct wires.

Two of the three original potentiometers within the device can be easily removed without damaging the original device's structure. The third, contained within the section fitted with a microphone style jack was not broken open so as not to permanently damage the device. In place of this, a 47 K-Ohm off-the-shelf potentiometer made by Tyco Electronics (Wilmington, USA) has been used.

To allow the forces of a virtual needle puncture to be performed, whilst not puncturing the tactile pad or causing safety concerns, the shaft of a real interventional radiology needle is shortened to prevent protrusion from the device. The shortened needle is simply secured within a holster by three socket head screws, which pinch a plastic section of the hub below its wings, see Figure Figure 6-14.

Figure 6-14 Left: A visual comparison of the off the shelf Omni with stylus removed and modified end effector. Centre: The needle hub mounted within the Omni end effector, providing high fidelity visual and tactile feedback as grasped. Right: The modified Omni as gripped by a user.

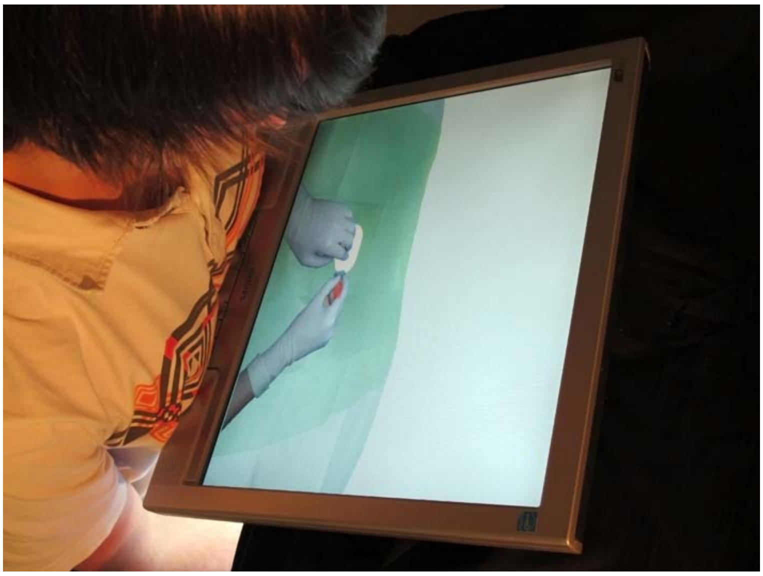
6.4 Force feedback summary

The novel palpation force feedback hardware developed here requires only two £200 (GBP) devices to provide 5 force DOF. The lowest cost device that can match or exceed this device's range of force DOF is the Virtuose 6D Desktop, a 6 force DOF device costing approximately £25,000 at the time of writing, although the device can only produce a continuous force of between 1.4 and 3N, far below that required to simulate a palpation. Devices meeting both the desired number of force DOF and the necessary force capabilities, such as Force Dimension's Delta 6, cost £36,000 or more. A concise force calculation, encapsulating *in vivo* measured force data has been used in PalpSim so that an atlas of patient specific force profiles can be simply integrated when data becomes available. The hydraulic tactile end effector, described in Chapter 5, can be securely fastened onto the hardware thus allowing both force and tactile feedback to be simulated. Actuation from below the tactile end effector allows the AR camera to capture a full hand image like that seen during an *in vivo* palpation.

There are currently no virtually simulated needle simulations that use a real needle hub and provide the 6 DOF required for both natural needle movement in real world space, allowing the selection of an arbitrary puncture site on a virtual patient. The hardware produced here provides these 6 DOF and 3 force DOF via a real needle hub interface to closely, although not completely, recreate the forces felt during a needle insertion. The device's wrist arrangement allows a needle to be used in natural positions and orientations without obscuring the AR camera's view of the user's hand. The real needle hub provides meaningful visual feedback together with the correct tactile cues as the needle is grasped, as well as allowing a real guidewire to be passed through it, facilitating the device's future integration into a full interventional radiology simulation solution.

The integration of these two hardware solutions within the PalpSim environment is described in the following chapter.

7 Simulation Realisation



7.1 Introduction

In this chapter, the individual components of the simulation environment described in chapters 4, 5 and 6 are brought together to produce an exemplar AR visio-haptic training environment called PalpSim. PalpSim provides an AR femoral artery palpation and needle insertion solution that overcomes problems inherent in both virtual and mannequin based simulations approaches.

Figure 7-1, illustrates the hardware setup as seen by the trainee as they approach the simulator, whilst Figure 7-2 depicts the side view of the rigid frame to which the hardware is securely mounted. To view the simulation correctly, the user stands close to the display with their chest touching the edge of the monitor. The fixed position of the hardware ensures that once the components have been correctly aligned, the simulation will always perform accurately, allowing trainees to spend time improving their skills rather than calibrating the hardware prior to

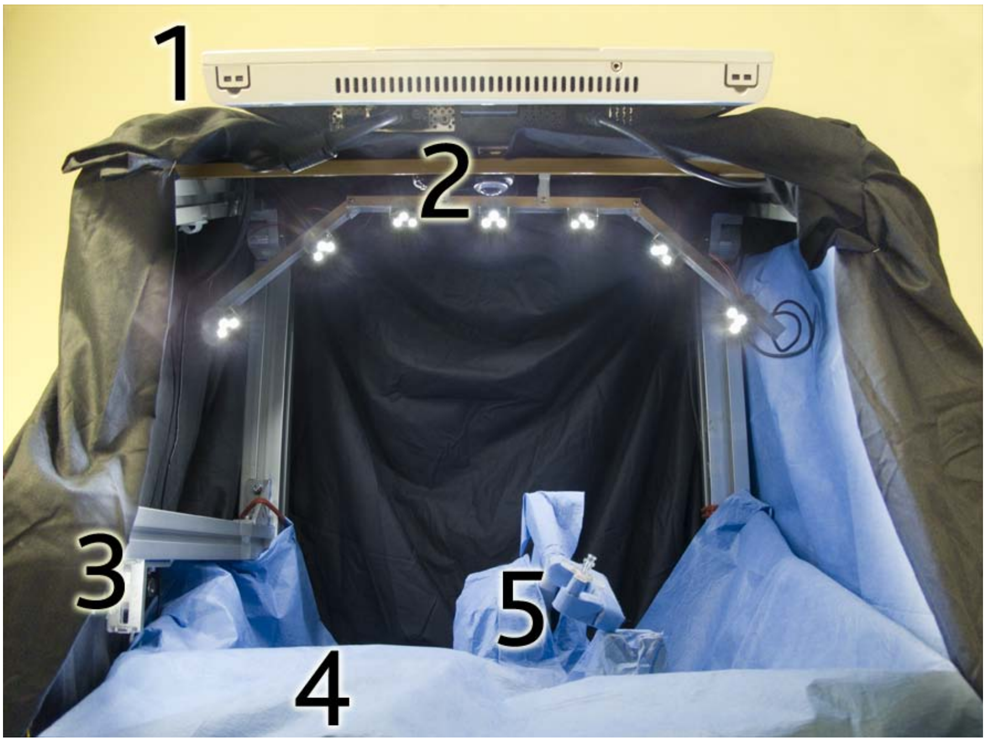


Figure 7-1. A collocated Haptic / Augmented Reality workstation. The LCD display (1) and camera (2) are mounted above the haptics devices and are used to display a live feed of the user's hands. Bright lighting illuminates the workspace to achieve a fast shutter speed. A low resolution side mounted camera (3) is used in a shadowing effect of the user's hands. This detects the height of the user's fingertips above the palpation haptic device (4) hidden below the blue sheet. The real needle hub (5) is attached to a modified Omni haptic device to provide 3 force DOF force feedback.

training.

7.2 Component Alignment

There are multiple components within the PalpSim environment. Static objects are placed at known distances from the camera and the positions of objects influenced by haptic interfaces are calculated by summing the known position of the device's base with the translation and rotation of the end effector. A perspective view of the virtual world is created at a fixed position and orientation, determined by a pre-calculated optimum eye position above the screen, see Figure 7-2. Legs which can be simply adjusted in length are to be added to the display in subsequent revisions, catering for all heights of trainee. From this eye position, the projection's field of view (FOV) can be calculated from physical measurements of the eye's position above the screen and of the screen's height (OpenGL's y coordinate). The aspect ratio of the screen's width to its height is then measured, to be used in an OpenGL perspective projection. Within the scene, objects described in chapters 4, 5 and 6 are placed at the positions in which they would exist in a real world palpation and needle insertion. A description of the

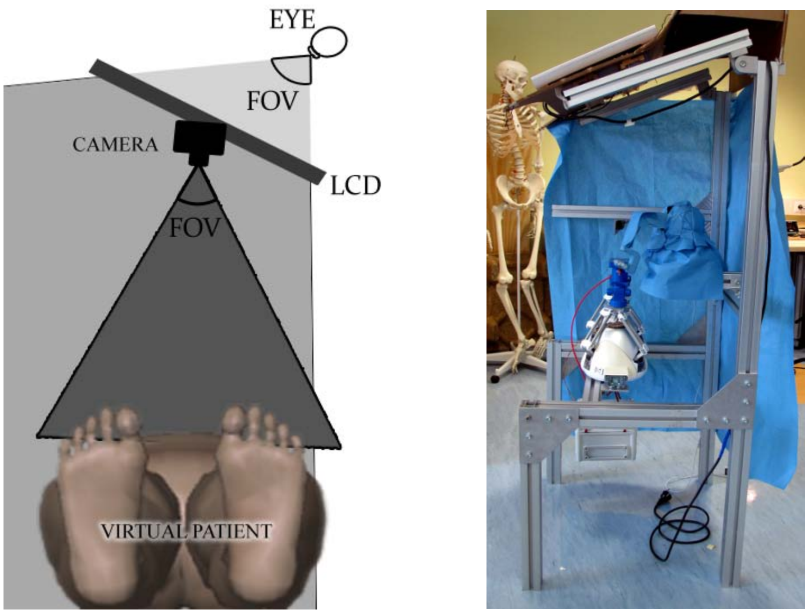


Figure 7-2 Left: The eye's and camera's view point of the AR visualisation. The virtual projections field of view (FOV) is calculated through physical measurement of the eye position above an LCD monitor of known screen height and width. The camera films the interactions below the monitor and this image is super imposed into the virtual scene at a fixed height above the virtual patient. Right: A side view of the uncovered simulation hardware reveals the positions of the devices securely fixed to a rigid frame.

individual components and their position within the PalpSim follows.

7.2.1 Visual components

The visually rendered simulation components and their position within the scene are described below. The full simulation can be seen in use in Figure 7-8.

- **Fenestrated surgical drape** – A static positioned texture map of a patient covered by a surgical drape. This represents a draped patient lying on an operation table at waist height and is placed fractionally above the deformable skin mesh. See Chapter 4.3.3, Figure 4-12.
- **Deformable skin mesh** – A deformable mesh modelled using Bullet Physics' soft body library and deformed using a proxy object, which mimics the position of the palpation end effector. See Chapter 4.3.1, Figure 4.3. The position of the circular mesh's border is fixed just outside of and below the transparent opening in the fenestrated drape texture. The nodes within the fixed border are free to move under the control of the position based solver. A skin texture is mapped to individual faces of the deformable mesh as it is generated so that as the skin mesh deforms, the skin texture visibly stretches.
- **AR hand visualisation** – A static positioned texture map used to display the modified video stream from a camera mounted below the LCD monitor, see Figure 7-1. The modified video stream, displaying only the user's hands and the needle hub (see Chapter 4.3.1.1, Figure 4-4(5)), is positioned 50mm above the fixed height of the skin's border pixels. This height has been determined to be the height in space at which the hands are most frequently placed during the palpation and needle insertion procedure. At this fixed height in real world space, the x and y span of the camera image is physically measured and these measurements are used to determine the x and y dimensions of the texture map in the virtual space with a 1:1 mapping between real and virtual measurements. The user can move their hands and the needle hub approximately 200mm higher or lower than this real and virtual world fixed texture height without a noticeable image distortion in the AR visualisation.
- **Shadow image** – Two grey shadow textures of varying opacity are placed within the scene at the same height as the hand texture but at varying x, y offsets depending upon a factor of the hand height, see Chapter 4.3.2, Figure 4-11.

- **Needle** – An IR needle is modelled using OpenGL shape primitives. The needle position and orientation follow the modified Omni force feedback device at it is moved, see Chapter 4.3.4, Figure 4-14 and 4-13. The needle position is homed whilst the haptic end effector is placed in a fixed position calibration shoe. The end effector's translations and rotations from this position are then accurately applied to the virtual needle. The virtual hub is obscured from view by the real hub seen in the AR visualisation and the virtual shaft follows this real world hub image.
- **Needle shaft shadow** – The needle shadow is drawn as a simple semi-opaque line primitive. The head and tail of the line are rendered a constant one millimetre lower than needle tip. The x, y head and tail of the needle shadow are positioned at the x, y position of the needle tip and hub respectively, adjusted by a factor of the needle tip's height from the skin surface.
- **Blood** – As the needle's tip penetrates the femoral artery, particles of blood are produced at the needle hub to simulate bleeding. The position of each particle is determined by the position at which it exited the needle hub, the pressure at which it was produced and the length of time it has been airborne. Gravity acts realistically to pull the blood particles down onto the surgical drape. As the particles hit the drape, they are destroyed and a semi-opaque drop is rendered at the hit position on a previously transparent texture map, statically placed 0.1 mm above the surgical drape texture. This texture appears to stain the drape below it. See Chapter 4.3.5, Figure 4-14.

7.2.2 Haptic Components

The position and orientation of the haptic devices dictate the positions of visually rendered components with which they are collocated and also the haptic feedback felt. The device's alignment with visual components and a brief explanation of how the positional information of each device is used to produce meaningful haptic feedback is outlined below.

- **Palpation haptic device** – The centre of the palpation hardware's tactile pad is positioned at exactly the same height as that of the circular skin mesh's border and the x, y translation matches the un-deformed skin's centre. The end effector remains at this position by compensating for the gravitational forces acting on it. As the user palpates the virtual patient, the position of the device changes and the tactile device begins to pulse as the skin is compressed towards the femoral artery. The resistive force felt is relative to the displacement of the haptic end effector. The device's position is relayed to the graphics loop so that the skin can be visually deformed in synchronisation with the haptic deformation. For a full explanation of the force calculation see Chapter 6.2.2.
- **Needle force feedback device** – The needle position is homed using the calibration shoe, as described in the aforementioned visual representation of the needle device. The transformation matrix of the needle hardware is used to calculate the position of the needles tip. As the needle touches the skin, the needle rotates around the tip position allowing the user to feel the skin's surface before it is punctured. As the needle enters the patient's tissue, the angle of entry is fixed such that the haptic representation of the needle's orientation does not change, even if the needle haptic device's pitch and yaw are manipulated during a needle puncture. This highlights a limitation in the needle simulation hardware as a realistic limit to the adjustment of the needle's pitch and yaw cannot be conveyed to the trainee. The force felt during needle insertion is calculated as a factor of the rate of change in needle tip displacement in relation to the needle insertion site (the point of puncture) on the skin's surface. As the skin's surface is compressed (moving the palpation haptic device downward), the depth of the needle tip within the tissue decreases proportionately and a retracting force is felt at the needle hub unless the needle is also advanced downward. For full details see Chapter 6.3.

7.3 Multithreaded program design

Figure 7-3, describes an overview of PalpSim's multithreaded operation and data exchange. The multithreaded program design optimises the use of available processor time on a multi-core processor architecture and has been tested on an Intel Core 2 Quad Q9650 processor.

Program Initiation

On initiation, the program creates the individual threads which run without interruption simultaneously until the program is terminated. Information is passed between threads using a double buffering technique for image data and mutual exclusion to prevent information being accessed simultaneously.

Side Hand Extraction Loop

On initiation, this thread loads a hand chrominance model for the side-mounted camera. A continuous hand extraction cycle is then initiated in which the lowest position of the hand within the image is discovered (normally the fingertips). This position is relayed to the state buffer and is used in the draw procedure to adjust the position of the hand shadows accordingly. A high refresh rate of 60Hz or higher can be achieved as the image is processed from the bottom up. Therefore, the speed of image processing increases as the hand approaches the skin's surface, as less of the image needs to be processed. This in turn increases the shadow displacement accuracy, because as the hand comes close to the skin, the height information is updated more frequently.

AR Hand Extraction Loop

On initiation, this thread loads a hand chrominance model for the top-mounted camera. A continuous hand extraction cycle is then initiated. The hand and needle hub are extracted from the background to create a "hand" image. A grey scale "shadow" representation of the latest hand image is also created. The hand and shadow image buffers are swapped as a complete camera image is processed and a new image is captured. The completed images are used in the draw procedure.

Graphics Loop

This thread regulates the draw procedure for the visual environment. Positional information pertaining to the palpation and needle end effectors is read from the position buffer. From this information, the needle position is modified and the skin deformation performed. The virtual drape, skin, needle and needle shadow are then rendered before the latest hand and shadow images are read from the image buffer along with the hand's position above the skin. The two shadow positions are then calculated and the shadow images are rendered before the hand image is then rendered in the scene. The blood particle system operates in synchronisation with the pressure felt at the palpation hardware's tactile interface. To ensure this synchronisation, momentarily before rendering the blood, the current pulse state is read and used to update the system. This cycle is continuously repeated with the latest available information.

Force Feedback Loop

On initiation, the individual palpation and needle insertion devices are started, each of which run in an individual thread. The primary force feedback loop uses the position and orientation information to produce meaningful feedback at 1000Hz. The position information is passed to the state buffer for use during visual rendering.

Tactile Feedback Loop

The fixed time step pulse loop controls the rhythmic generation of a pulse profile that is translated into a series of piston displacement/servo rotation commands. This is transmitted to the USB servo controller producing a pulsing tactile cue. The current state of the pulse pressure is written to the state buffer at each time step to visually synchronise the pressure of the blood flowing from the needle hub as the artery is punctured. The current position of the palpation end effector is read from the buffer to determine if the device should pulse or not. If a palpation is not occurring, the device does not pulse.

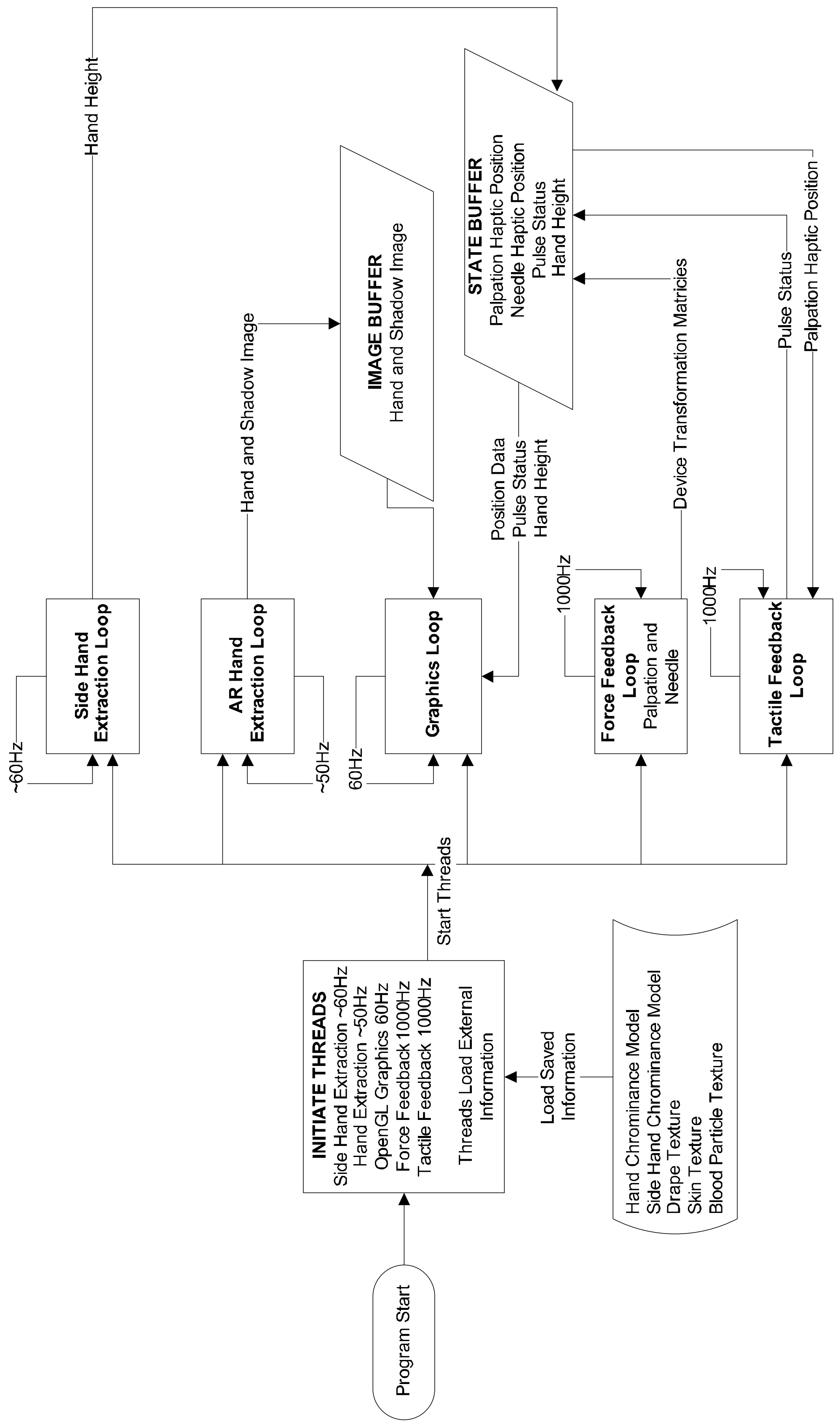


Figure 7-3 A flow diagram description of the multi-threaded interaction within PalpSim

7.4 Device Workspace Calibration

Each component in the simulation is calibrated with the other components to use exactly the same virtual and real world measurements of space. A 1:1 scale of millimetre measurements is used between the virtual and real world measurements to ensure consistency. Figure 7-2 illustrates the fixed positions of the hardware components, securely fixed to a rigid frame at known positions.

An example of the calibration procedure can be seen in Figure 7-4. Here, 15 measurements at 1mm intervals have been made in the device's z coordinates. Full device calibration is achieved when the perceived virtual world end effector position matches its real world position. If a device is correctly calibrated, a comparison of the real world measurements (from the device's movements) with the virtual world measurements (the position at which the device thinks it is) would create a plot that tracks along the red identity line in Figure 7-4. In this example, the device's z scale must be adjusted as translation of 15mm in the real world is overestimated in the virtual world by a factor of approximately 1:1.49 (where a translation of 15mm in the real world gives a device reading of 22.35mm). This is repeated for each axis (x, y and z) multiple times until the calibration is deemed accurate. This is calibration is performed for both of the Falcons and for the Omni so that if the palpation and needle devices are translated together in the real world the device's virtual world representations also move together.

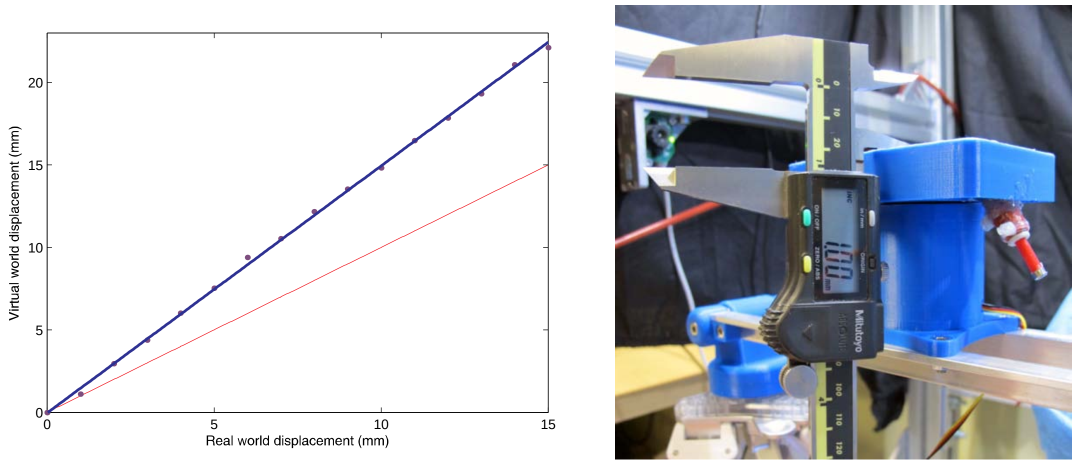


Figure 7-4 Left: A plot for calibration. Comparing the expected device position in red (device reading) with the actual (real world measurement) position in blue. Right: Precise measurement of the end effectors real world position.

The 1:1 AR mapping between virtual and real world space was verified by visually comparing the alignment of real world objects in the AR video stream to virtual representations of the objects of equal dimensions. Both the virtual and real world objects were placed at the same position and orientation and the alignment of their edges was visually compared. Figure 7-5 shows this virtual and real world calibration check. Here a black block of known dimensions is placed into the augmented reality workspace at a known position within the real world. A virtual red line with exactly the same length as the edge of the real world block is then drawn in virtual space where the blocks edge is expected to lay. If the AR cameras image has been correctly scaled and positioned the block's edge will align with red line perfectly. The calibration of the force feedback hardware's workspace can then be checked by touching the two ends of the black block and comparing the recorded virtual world position with the known real world position of the calibration blocks corners. This error should be zero mm to ensure correct alignment although an error of 0.5 mm was accepted over a 150 mm calibration distance (end to end of the block) as this was considered the accuracy to which the device could be effectively moved and aligned by hand.

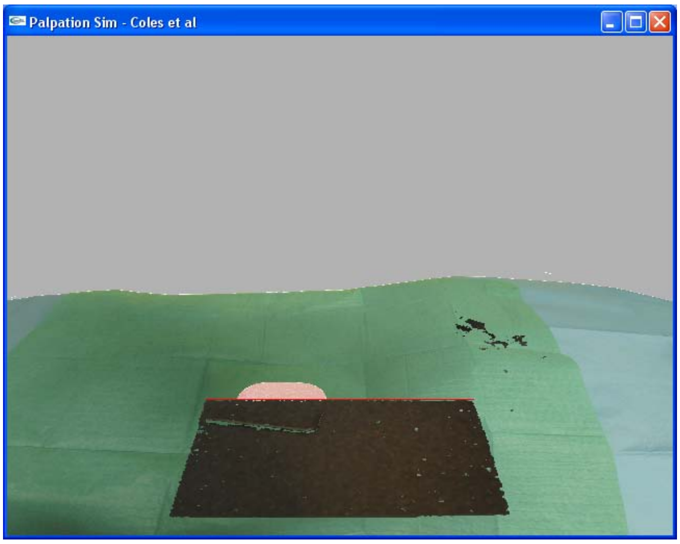


Figure 7-5 Visual calibration to check alignment of AR camera. A real world black calibration block of known dimensions is placed at a known real world position within the AR workspace. The AR image of this block is then compared against a virtual world representation of the blocks edge, drawn where the blocks edge is expected to lie.

7.5 Force Calibration

It is important in a haptic simulation to calibrate the force output as an un-calibrated force feedback device instructed to provide X Newton(s) of force may not be providing this force at the end effector. In the case of force felt during palpation, a device that provides less force than required will appear to simulate a larger patient with more body fat and a device that exceeds the desired force, a thinner patient. Where a training simulation is used to train the processes and procedures of a task, high force accuracy may not be critical but, if pre-procedure rehearsal is a simulation goal, a miscalibration between simulated and real world force could be critical.

Little is known about the accuracy to which a practitioner can recall the force profile of a medical procedure they have performed and as such the necessary accuracy to which the forces must be simulated is unknown. During simulation it may only be necessary to reproduce coherent cues, i.e. a palpation of fatty tissue is followed by a needle insertion of fatty tissue, for effective training however, it is reasonable to assume a higher accuracy will be required. Using an immersive visio-haptic environment such as that developed here, psychological studies can be performed in the future to further understand the perceivable force accuracy. These studies can be used to develop guidelines regarding the simulation fidelity necessary for effective training.

The simulation calibration phase used the Royal Liverpool University Hospital's

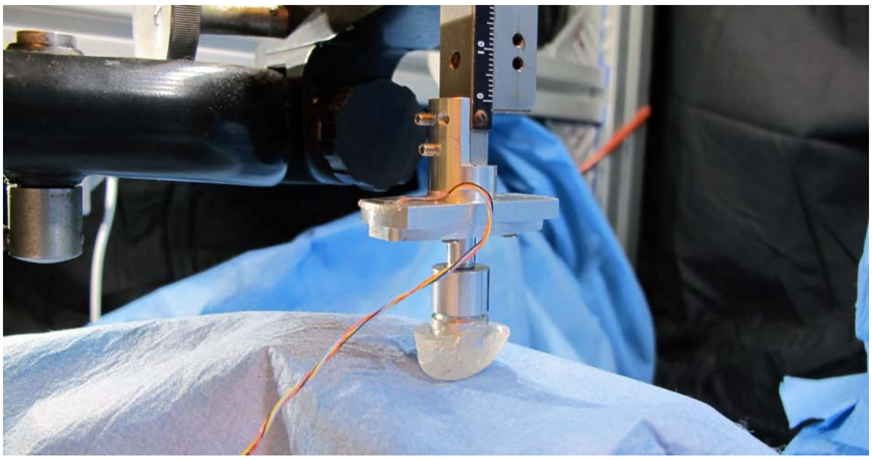


Figure 7-6 The Royal Liverpool University Hospital's palpation force sensor [183] used to record the forces felt during an *in vivo* palpation have been used to calibrate the force felt at the haptic end effector.

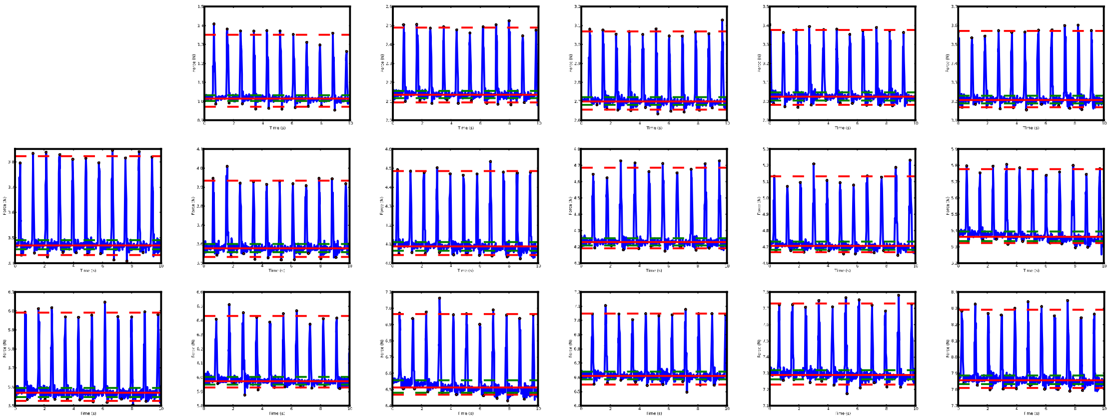


Figure 7-7 The calibration phase analysed 17 ten second force measurements of the force felt at the palpation haptic devices tactile end effector to find the mean force output for each. This mean force output was then compared against the desired mean force output, see Figure 7-8.

palpation force sensor [183], the same force sensor used to record the palpation forces *in vivo*, to measure the forces felt at the palpation hardware's end effector, see Figure 7-6. A desired force output (the simulated force) was then compared against the recorded force output (the force felt at the tactile end effector by the force sensor).

The calibration compared 17 simulated forces with 17 recorded forces as felt at the pulsing palpation end effector in a similar measurement approach to the *in vivo* force measurements, although only five different force measurements were made at the *in vivo* stage. The range of measurements made (17) was determined by the force limitations of the sensor used (9 Newton). Each calibration

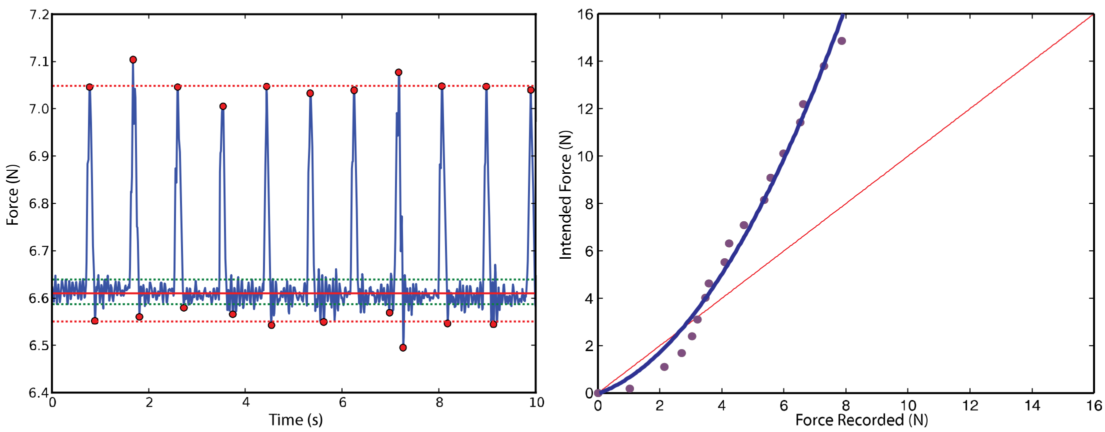


Figure 7-8 Left: An example 10 second force reading taken from the palpation end effector using the same force sensor used to obtain the *in vivo* palpation force readings. Dashed red lines identify the mean maxima and minima of the pulse peaks. Green dashed line indicates the 80^{th} and 10^{th} percentiles of the complete data set. Solid red line indicates the mean of the date which falls between the 80^{th} and 10^{th} percentiles. Right: The desired / intended force output (device instruction) compared against the achieved force output (mean force recorded at the end effector).

recording encapsulates a constant palpation force, a time varying pulse and sensor noise, see Figure 7-7. Figure 7-8 (left) shows a single example. For each ten second recording, the average constant force output was approximated (highlighted by a solid red line) by finding the mean of the recorded data within the 10^{th} and 80^{th} percentiles (highlighted by the dashed green line). Recorded data points outside this percentile range were considered a force peak generated by the pulsing femoral artery. Each of the 17 individual force recordings were processed, see Figure 7-7, and the expected force outputs compared with the recorded outputs in Figure 7-8 (right). If correct calibration is achieved, each data point should lie on the red identity line. However, this does not occur and, on average, the expected force output exceeds the achieved output. This miscalibration could have occurred for a number of reasons. If the miscalibration had been linear, the force adjustment could have been quite simple to adjust by applying a scale factor to the intended device output. However, the observed non-linear behaviour could be due to a number of factors, including a non-linear force output of the Falcon device [199], inaccuracies in the gravity compensation procedure and the non-modelled device friction. Initial attempts to correct this have taken a two-part approach. First, a slight reduction in the force output of the gravity compensation procedure in an effort to straighten the initial curvature of the plot, and, secondly, a scaling of the device's force output. Additional time with the palpation force sensor would allow the force output to be accurately measured and calibrated, but unfortunately the sensor was not available for additional use. As such the accuracy of the force output in comparison the *in vivo* pulse profile is unknown until further recordings can be made. A commercial force sensor could be used to enable a faster feedback cycle as processing time could be reduced.

7.6 Summary

Figure 7-9 shows PalpSim in use as it was presented for face and content validation. In this figure the user's palpating fingertips slightly penetrate the silicone tray and feel the sensation of realistic skin stretch as the skin is touched. The fine, high frequency changes of the pulsing femoral artery can be felt in addition to tissue compliance as pressure is applied. The user's right hand can be seen holding the real needle hub and the virtual needle shaft can be seen penetrating the deforming skin. Pulsing blood flow appears to emanate from the real needle hub.

The multithreaded simulation runs smoothly on a single quad core processor (see section 7.3) with a alignment error of less than 0.5 mm per 150 mm. The results of the evaluation study carried out at the Royal Liverpool Hospital are described in the following chapter.

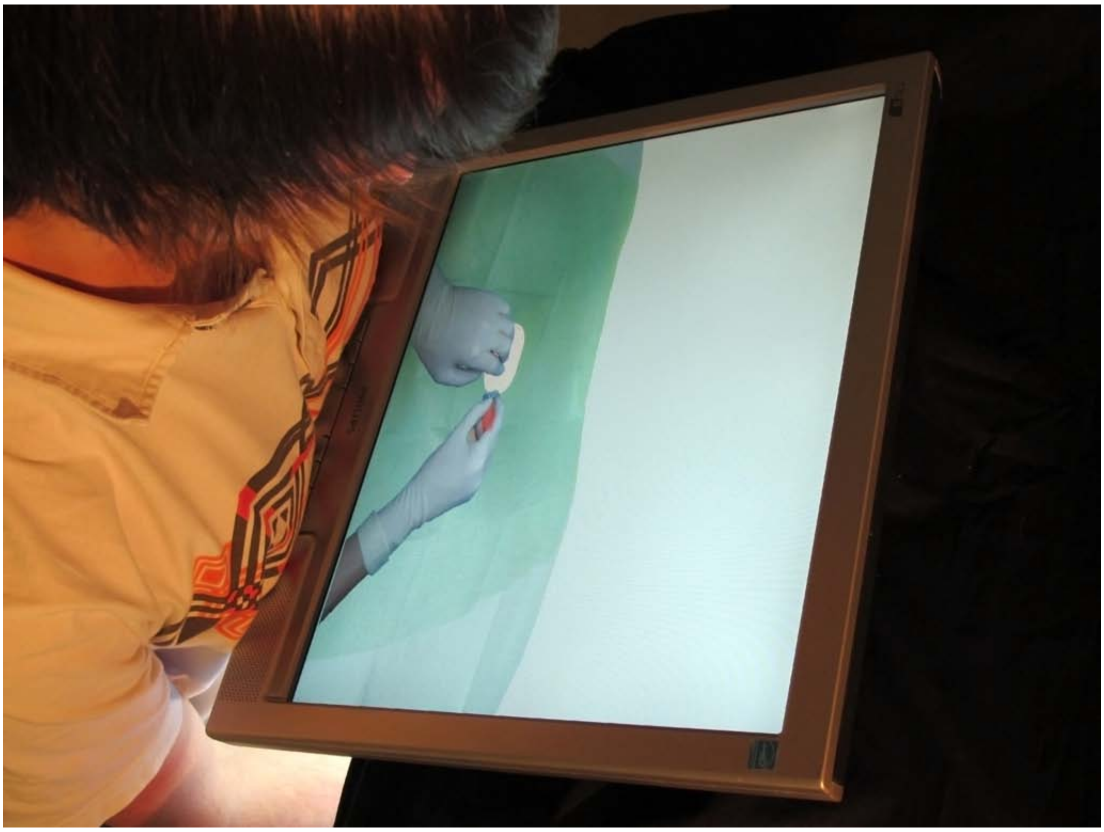
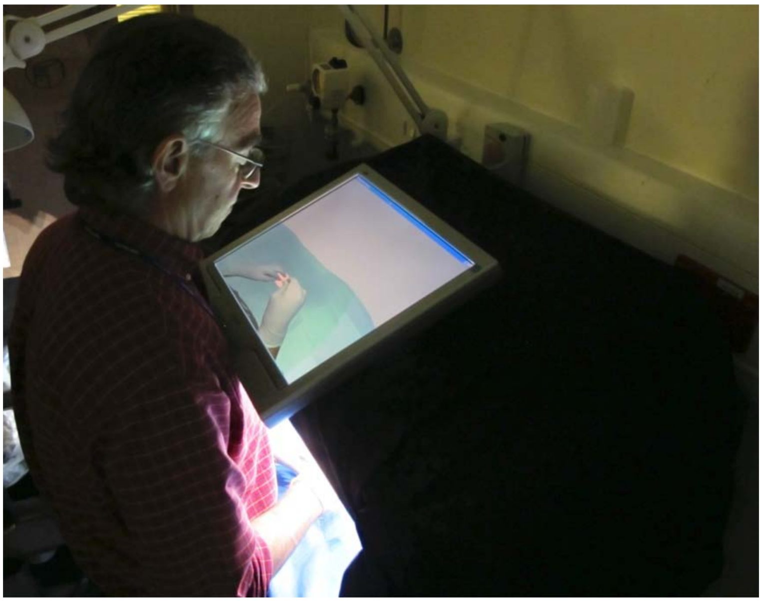


Figure 7-9 PalpSim in use. The user can be seen palpating the virtual patient with their left hand and inserting the needle with their right. Blood emanates from the needle hub as the femoral artery is punctured.

8 Validation



8.1 Validation Overview

The PalpSim simulation environment is validated here to draw conclusions on the effectiveness of the developed visio-haptic environment. A face and content validation study, the first of many validation steps, has been conducted so that experts can subjectively judge the simulation. The tactile feedback felt at the end effector has then been evaluated in section 8.3 and future validation steps are discussed in section 8.4.

8.2 Face and Content Validation

A face and content validation study was conducted at the Royal Liverpool University Hospital in which seven Interventional Radiology experts with five or more years experience as a consultant were asked to test PalpSim and provide objective feedback in a 29 point questionnaire, see Appendix 10.3. The format of the questionnaire was loosely based on previous work from the CRaIVE consortium in the UK [122]. The 29 questions asked were phrased by an expert interventional radiologist to ensure the question followed tasks described in the task analysis where appropriate.

Figure 8-1 Left: A full length view on the simulator situated in the radiology department of the Royal Liverpool University Hospital during the face and content validation study. An additional monitor allows the trainer to watch the progress of the trainee. Right: Consultant Vascular Interventional Radiologist Prof Derek Gould testing the simulation.

The intention of this study was to support the hypothesis that an AR approach to medical simulation offers a powerful alternative to existing simulation methods, facilitating collocated visio-haptic simulation of both direct touch between practitioner and patient (touching the patient's skin), and touch between a tool and a patient (touching with the needle). If this is demonstrated, the simulation can then be extended to simulate varied, case specific patient data.

The 29 questions were devised to gauge the practitioners' expertise in the field and to obtain objective feedback about the simulation's validity in reproducing the key points identified during a comprehensive task analysis of the interventional radiology procedure [26]. A five point Likert scale was used to evaluate the skills and experiences of the practitioners and a seven point Likert scale was used to record validity feedback. Question 29 asked for additional feedback, the responses given can be read in Appendix 10.4. Questions 4 and 8 were posed without space for feedback as the tasks of locating anatomical landmarks and the pre nicking of the skins surface were not simulated. These questions were included to inform the practitioner the cues were missing but not forgotten. PalpSim was set up in a working radiology department; allowing the consultants to take time to use the simulation between cases, whilst the cues of a real procedure were fresh in their minds, see Figure 8-1. The results of the study are shown in Figure 8-2 and discussed below.

8.2.1 Practitioner Experience

Of the seven experts, three had used a simulator in a training program before and most played computer games of some sort either monthly or at least every six months (Q18). All of the reviewers rated their computer skills between average and very good (Q19), and their interventional radiology skills as good or very good (Q1) whilst feeling that they either performed average or good whilst using the simulator (Q2).

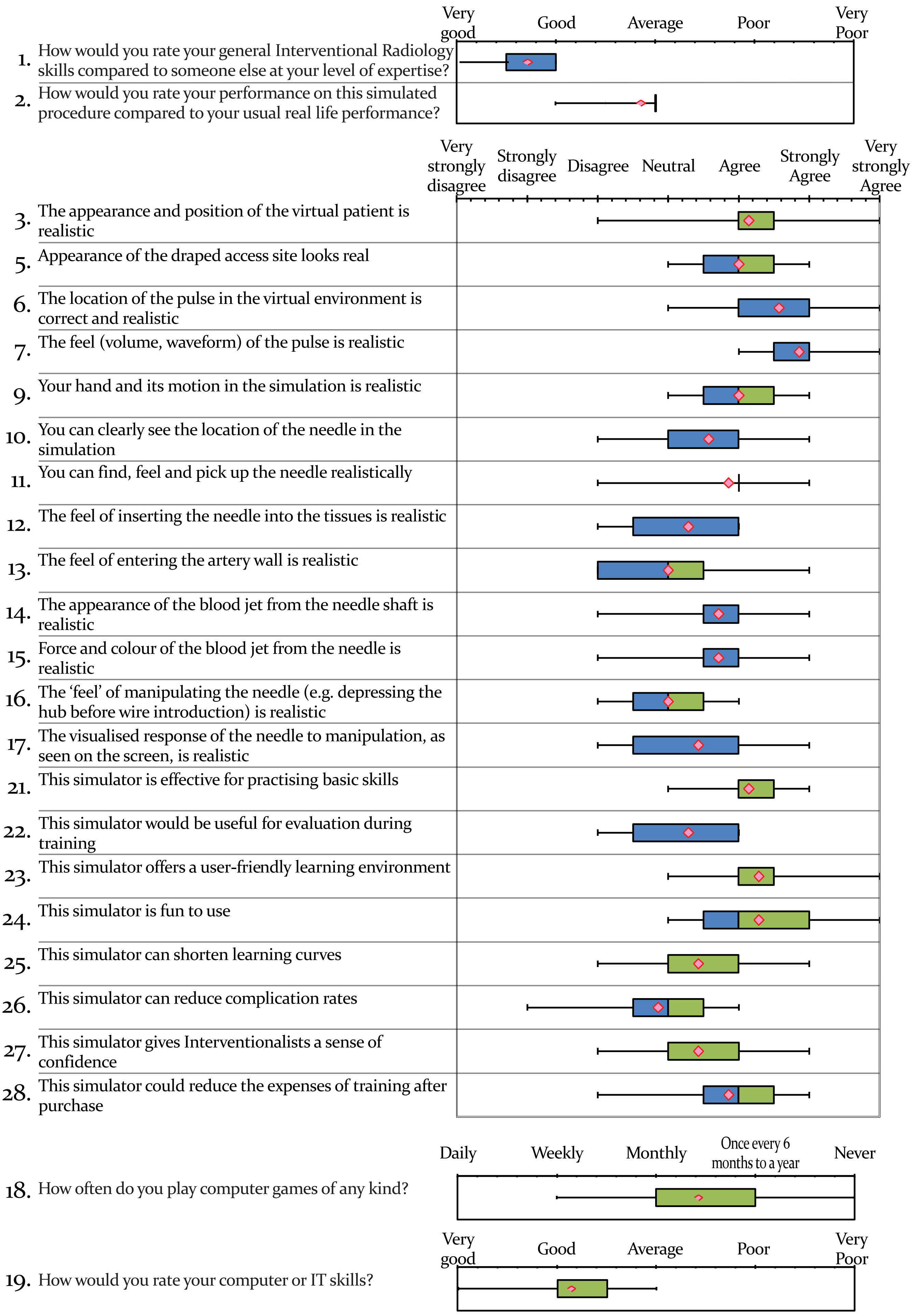


Figure 8-2 PalpSim validation. Questionnaire results from 5 and 7 point Likert ratings. Whiskers indicate data range. Green, range between median to the upper quartile and blue the median to lower quartile. Red diamond indicates the mean feedback.

8.2.2 Visual Appearance

The experts mostly agreed that the appearance and position of the virtual patient (Q3) and the draped access site looked realistic (Q5). The movement of the users hands within the augmented reality workspace was rated as realistic (Q9), a critical element of the AR simulation. The mean feedback to the statement "You can clearly see the location of the needle in the simulation" was slightly towards agreement, but it is thought the use of a higher resolution monitor would achieve a far better level of feedback (Q10). The relatively low resolution of the monitor used during evaluation meant that, as the virtual needle was inserted into the patient, the shaft was only a couple of pixels wide due to the LCD's coarse pixel granularity. This is simple to improve through a relatively inexpensive hardware upgrade. Despite this, it was agreed that the needle could be found and picked up realistically (Q11). A range of responses from "agree" to "disagree" with a positive mean were provided regarding the visual response of the needle as it was manipulated in space (Q17). Negative feedback could have been influenced by a small visual lag sometimes observed between the virtual needle shaft and the AR rendered hand and needle hub. This occurs if a user translates or rotates the Omni needle device very quickly, as the OpenGL needle shaft will be precisely updated to follow the needle hardware each OpenGL draw cycle. However, the AR hand image intermittently takes two OpenGL draw cycles to display. When this occurs a short one-frame scene lag can be observed in which the virtual needle shaft moves in advance of the hand image. This was hard to notice at standard procedure speeds, but much easier to see if the needle is moved rapidly.

8.2.3 Virtual Palpation – Feel

The simulation of femoral palpation, both the haptic hardware and visualisation, was received extremely enthusiastically by the practitioners (Q29). Both the location of the pulsing femoral artery and the tactile feel of the volume and waveform scored highly with the mean of the results tending towards strong agreement for both statements (Q6, Q7).

8.2.4 Virtual Needle Insertion – Feel

A difference in opinion with regards to how the force conveyed during a needle insertion was interpreted lead to only slightly positive feedback for the feel of inserting the needle through tissue (Q12). As a control to ensure that participants were not exhibiting a positive bias in their response, a further question asked them to score the 'feel' of entering the artery wall (Q13). This functionality has not been implemented in the current version of the simulator and so should have received a low score, which indeed it did. However, in hindsight, it is thought this question confused participants as they tried to match the forces they had felt during the trial to the forces they were asked to rate in the questionnaire. Previously, in questions 4 and 8, the practitioners had been explicitly notified that cues were missing. After a discussion and a repeated (unrecorded) trial with the practitioner who thought he felt the needle "pop through into" the artery, it emerged that the pop he recalled was a slight pop through the skins surface. This indicates the force profile of the needle insertion may need further work. This is also supported by the neutral feedback given for the feel of manipulating the inserted needle within the tissue prior to the insertion of a wire through the needle hub, although wire insertion is not simulated (Q16). Here the non-actuated rotation DOF would have also lowered the obtainable score that such hardware could obtain, and low feedback was expected.

8.2.5 Virtual Blood

The consultants mostly agreed that the appearance of the blood jet from the needle shaft was realistic (Q14) and also its colour, and apparent force that varied as the pulse pressure changed (Q15), was realistic. This illustrates one of the advantages of such an AR visio-haptic environment, as this simple to implement cue is difficult to realistically incorporate using other simulation mediums. Physical simulation mediums require liquid to be introduced into the simulation and non AR virtual simulations of blood flow suffer from the previously mentioned occlusion problems, where the user's hand holding the needle would be incorrectly obscured by an image of the patient and the blood that had fallen onto the drape.

8.2.6 General Opinions

Questions 21 to 28 gave a broad overview of the perceived usefulness of the simulation as a training tool and received positive feedback. Although the simulation is designed to train a very limited set of skills, practitioners agreed that the simulation is effective for basic skills training (Q21), offers a user-friendly learning environment (Q23) and is fun to use (Q24). Practitioners did not feel this proof of concept simulator would reduce complication rates (Q26) and gave mostly neutral feedback on the simulation's ability to be useful for evaluation during training (Q22). It is thought that introducing an evaluation screen to the operators console would improve this rating, although this was not the goal of this technology validation. Slightly positive feedback was given to the simulator's ability to shorten learning curves (Q25) and for giving the interventional radiologist a sense of confidence (Q27). Practitioners also felt (a mean toward agree) the simulator could reduce the expenses of training after purchase.

8.2.7 Conclusions

During the validation, multiple practitioners verbally commented (Q29) that they were worried about pricking themselves with the virtual interventional radiology needle even though no physical needle shaft was exposed in the simulation. One practitioner said "I keep thinking I am going to prick myself, I had to check again there was no needle" as he looked under the monitor. This indicates a high sense of immersion was being achieved.

The palpation stage of the simulation scored highly with a positive attitude towards the novel augmented reality visio-haptic environment. A practitioner commented "The screen view is great, left hand feel is brilliant" (Q29). The needle insertion stage of the experience scored well, although not as highly as the palpation. This stage was primarily included within PalpSim to demonstrate the AR technologies ability to integrate simultaneous simulation of both direct touch between practitioner and patient and touch between a tool and a patient. This goal appears to have been achieved. It was acknowledged prior to testing that future simulation development should focus on a more in-depth analysis of the needle insertion force profile. The positive feedback obtained from the simulation of blood flow from the half-real, half-virtual needle hub is further evidence of the AR technology's powerful ability to create illusions not possible using other technologies. There were no comments that the hydraulic pulse hardware was audible, demonstrating the sound proofing of the pulse actuator was successful.

Although all of the available consultant interventional radiologists at the Royal Liverpool University Hospital participated in the validation study, the total number of participants is still too low to draw any conclusions with statistical significance. A typical radiology department has a limited number of experienced practitioners, one reason supporting the need for virtual training simulations, as practitioners have very little time to dedicate to training.

The results obtained during this study pave the way for repeated face and content validation studies to increase the result's significance and for more in-depth skills transfer studies now the technology has been shown to offer an intuitive training environment. However, further validation should only be performed after this simulation has been integrated with the simulation of additional tasks such as guidewire manipulation. This recommendation takes into consideration that additional validation will require a significant investment of psychologist's and trainee practitioner's time. Future validation steps are described in section 8.4 and the further development of the simulation before additional validation should be undertaken is described in section 9.2.

8.3 Validation of Tactile Feedback

To evaluate the accuracy of the pulse simulation, the tactile waveform felt at the end effector as an average palpation force of approximately 7.3N is applied has been measured using the fingertip force sensor designed by the Royal Liverpool University Hospital [183]. These measurements have been visually compared with

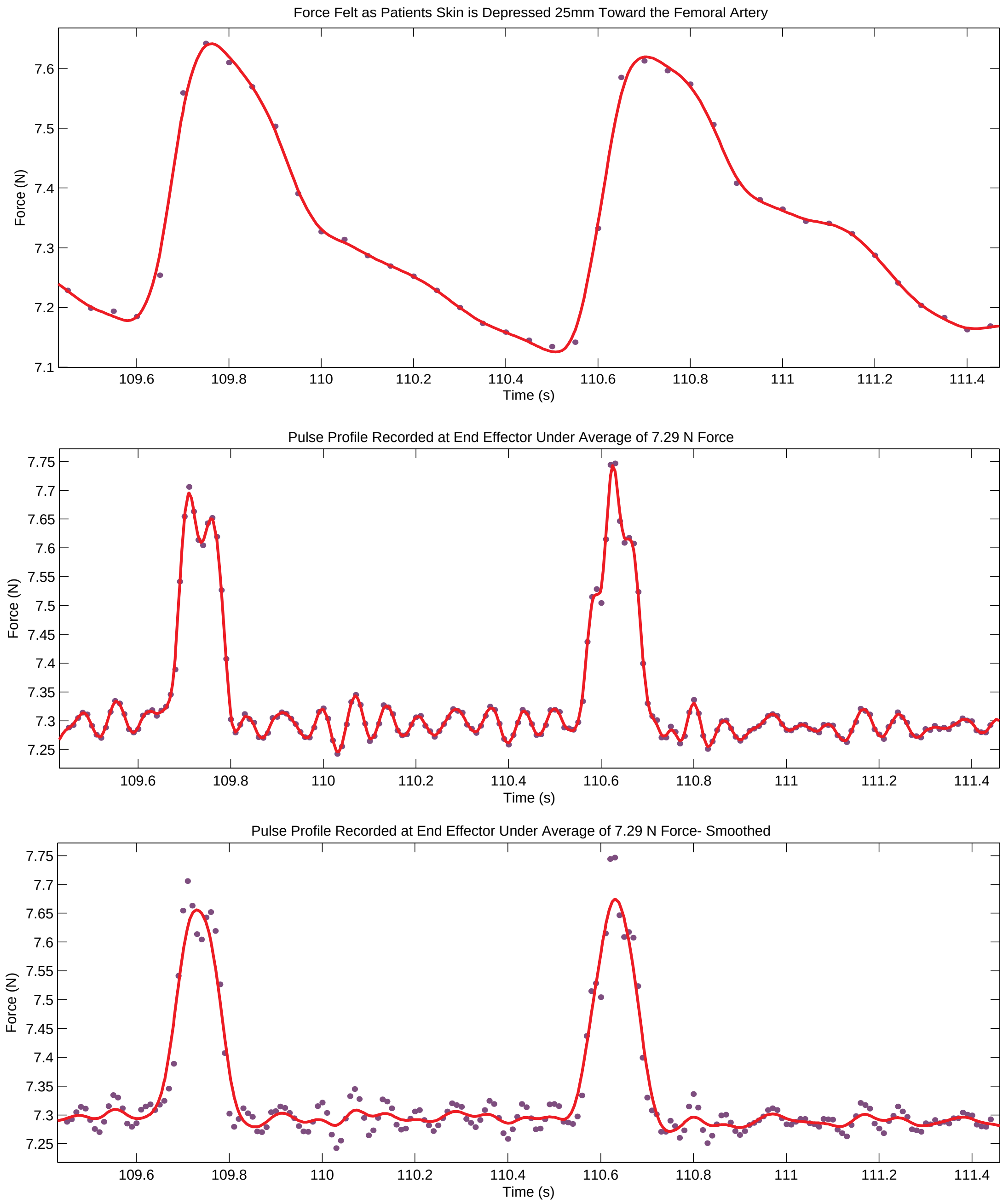


Figure 8-3 Top: A 2 second sample of *in vivo* pulse force measurements at a skin displacement toward the femoral artery of 25mm, an average applied force of 7.265N (over a complete measurement) and an average pulse fluctuation of 0.483N. Center: A 2 second sample of pulse force measurements from the tactile end effector with an average applied force of 7.2895N (over a complete measurement) and resulting average pulse fluctuation of 0.500N. Bottom: A smoothed fit of the measured force from the palpation haptic device to eliminate noise purportedly from the force sensor.

that of an *in vivo* palpation applying approximately the same palpation force. This comparison can be can be seen in Figure 8-3

The tactile devices pulse waveform appears to closely resemble that of the displacement profile of the hydraulic piston, seen in Figure 8-4. This profile was expected to become smoother and longer as the hydraulic tube running to the tactile end effector moved slightly during each pulse and the suspended casing swayed a little. The designed piston displacement profile pre-empted this smoothing effect, which doesn't seem to have occurred.

Through visual comparison of the *in vivo* measured pulse waveform with the simulated haptic waveform, both under approximately the same palpation compression force of 7.3N, it can be said that the tactile response achieved appears to be comparable as the artery expands. The fluctuating pulse pressure also appears to provide the correct variation in force for this constant palpation pressure. However, the *in vivo* measured force profile tapers off at a much slower rate that the simulated one, casting doubt on how realistic the "realistic pulse" that received very positive feedback in the face and content validation stages, actually is.

Although the tails of the simulated individual pulse waveforms do not accurately match those of the *in vivo* waveforms, it is thought that the fast increase in pressure that creates the correct fluctuation in tactile force could stimulate the correct tactile mechanoreceptors, whilst the relatively slow reduction in pressure

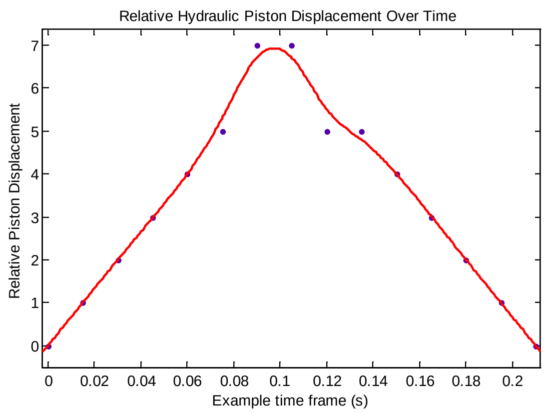


Figure 8-4 An example of the piston displacement used to produce a realistic pulsing sensation at the tactile end effector.

at the pulse's tail may not create such a significant tactile cue. It should also be noted that, during a femoral palpation and needle insertion, the pulsing femoral artery is used to locate and puncture the artery with a needle tip, rather than for diagnostic purposes where the profile of the pulse may be of higher importance. A full study could be conducted to better understand how a mismatch in pulse profile achieved a high practitioner rating, but this is deemed unnecessary as the virtual simulation's pulse profile can be simply modified to closely match the *in vivo* pulse in future simulation revisions, therefore producing a more valid simulation solution without consideration of feedback at the mechanoreceptor level.

8.4 Summary

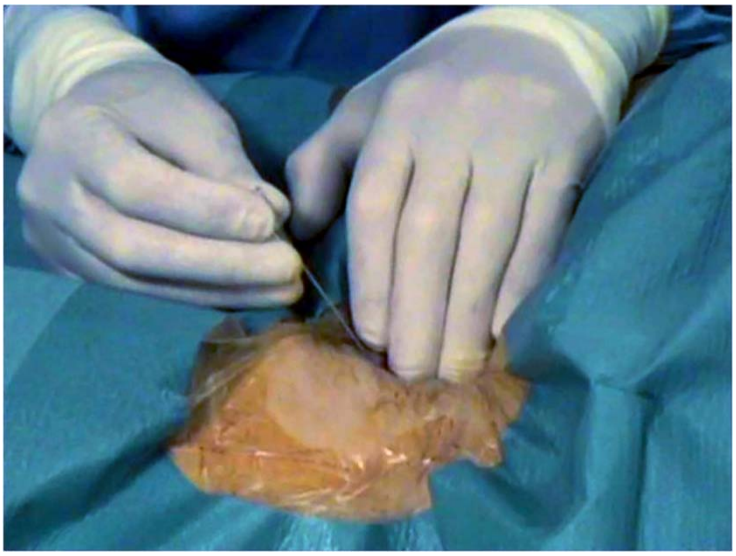
The successful first validation stage highlighted minor changes that could be made and these should be addressed before further cycles of validation are performed. Further force calibration of the palpation and needle haptic hardware should also be carried out.

Current methods of medical simulation validation are subjective. By increasing the size of the validation group, the statistical significance of results can be increased and therefore stronger inferences can be made. Until such a time that the multipoint contact forces and interactions between practitioner and patient can be comprehensively recorded and accurately modelled in real time, subjective reviews will remain the most accurate evaluation tool.

The performed simulation validation process required a practitioner to pass judgment on the simulation, whilst no definition of the simulated patient's habitus or medical condition was provided. Although this method of evaluation can be used to gauge if the technology has the potential to provide a high fidelity simulation environment, a more in-depth comparison of real and simulated feedback should be made. As the simulation is refined to accurately reproduce patient specific palpation and needle force data and the force measurement procedure becomes a fast seamless process, an accurate comparison of a real patient with a simulated patient could be made. This is the next proposed validation step for PalpSim. For this to be conducted a patient would first be palpated with a force sensor and the force results would be automatically fed into the simulation environment that would be situated in the same room as the real patient. A practitioner could then palpate the virtual patient's femoral artery and immediately after, compare the forces felt by palpating the real patient PalpSim aimed to simulate. If this comparison proved to be successful, a further validation step could be performed in which, during multiple trials, inaccuracies would be introduced into the simulated forces. The practitioner's feedback would then hopefully reflect and identify these intentionally introduced inaccuracies when comparing the simulation with the real patient.

Subsequent validation steps will aim to prove that the developed palpation and needle insertion simulation, as part of a full procedure training platform including guidewire and catheter manipulation, is beneficial to the training curricula through transfer of skills studies. In these studies, metrics such as those discussed in section 2.11 can be used to track as user's progression from novice to expert during training. However, an independent evaluation and comparison of *in vivo* procedural performance must also be made between the first procedures performed by trainee practitioners who have received virtual training and those who have not, to demonstrate (or otherwise) an advantage of virtual training. These studies must be designed in close collaboration with clinicians and psychologists to ensure the evaluation criteria are fair and meaningful and talks to this accord are currently being undertaken by members of Bangor University. Transfer of skill studies must be performed with a large subject group over an extended period of time.

9 Conclusions and Future Work



9.1 Conclusions

During this research, an investigation has been carried out into the application of Augmented Reality visio-haptic techniques for medical training. An exemplar simulation named PalpSim [200] has been developed to defend the thesis hypothesis. The hypothesis was:

> Femoral artery palpation and needle insertion can be virtually simulated, effectively substituting existing mannequin-based training methods. Off the shelf visualisation and haptics technologies can be modified to produce a low-cost visio-haptic simulation platform that provides a high fidelity femoral pulse palpation and needle insertion simulation, whilst overcoming the patient variability and simulation durability problems inherent in mannequin-based simulation approaches.

Four questions were posed in the introductory chapter to address the hypothesis. The first of these was: What are the problems with existing virtual simulation technology?

To answer this, a comprehensive review of current medical training simulations using haptics to enhance the training experience for a variety of medical procedures has been performed. The key issues identified are presented in Chapter 2 and can be summarised as:

- A simulation's cost affects it's uptake in training centres, so simulation cost must be weighed up against training benefit.
  - Customised force feedback devices are often required by surgical simulators, although the cost of multi-purpose hardware is commonly much lower. Modification of commercial hardware can be used to increase simulation face validity at a relatively low cost.
  - It may sometimes be possible to reduce the force DOF used in a simulation without a large impact upon training effectiveness.
- Tactile feedback is in its infancy and the number of devices that can be used in conjunction with force feedback hardware is very limited.

- Patient/tool interaction is commonly simulated, but simulation of open surgery and procedures where practitioners must touch soft tissue directly with their hands, remains a research challenge.
- Patient variability can be accommodated through virtual simulation if correctly designed. A full procedure simulation should also allow procedural variability between practitioners and institutions.
- A task analysis is required to define a simulation's requirements correctly and to perform validation. This can also be used to design performance metrics.
- Current virtual visio-haptic collocation methods are flawed. The immersive stereo display unrealistically occludes the user's hands as they interact below the projection plane. HMDs provide a narrow field of view, are high in cost and require expensive high precision tracking.

The second question asked was: Does haptics technology that can be used for effective virtual simulation of a femoral palpation and needle insertion currently exist? If not, can the technology be developed?

If either a palpation or needle insertion is to accommodate patient variability, then both the palpation and needle insertion simulations must be virtually simulated to make use of the variability provided by the other. A review of both procedures in Chapter 2 found the current haptic hardware to be insufficient for palpation simulation, whereas the current force feedback hardware typically used for needle insertion simulation provides low face validity. Hardware issues to be addressed were identified as:

- Palpation requires tactile as well as force feedback. A device capable of producing tactile feedback that could be mounted on to a force feedback device, whilst low in cost to develop, was not identified during the technology review.
- Six force DOF should be exerted at each contact during a palpation in combination with tactile feedback. Devices capable of producing sufficient force are expensive and a low cost device capable of achieving this functionality was not identified.
- Reviewed literature showed that a SensAble Omni force feedback device, the lowest cost stylus based device offering 6 position sensed DOF and 3 force DOF, could be

used to produce sufficient force feedback during a virtual needle insertion. However, the stylus interface does not provide meaningful tactile information as it is grasped, in comparison to a real needle hub, and if the stylus is not obscured from view, it does not provide the correct visual cues either.

These issues were addressed through the development of new hardware interfaces, a contribution of this work. Where possible, low-cost commercial devices were modified rather than manufacturing new devices to reduce production costs and to increase the efficiency at which the simulation could be distributed if required. These developments are:

- Four tactile technologies have been tested and one, hydraulic actuation, has been deemed most suitable to provide tactile feedback in a femoral palpation simulation. The developed end effector, a rigid plastic tray containing a pulsing femoral artery embedded in silicone, is hydraulically actuated using an off the shelf servo motor. The device can be securely mounted onto a force feedback device developed in Chapter 6.
- Two 3 force DOF Novint Falcons were modified by first rotating them through 90 degrees, and then the two device's end effectors were replaced with a single end effector that joins the two Falcons to produce a single 5 force DOF device. The underside of the hydraulic tactile end effector developed in Chapter 5 can be attached to it.
- A SensAble Omni force feedback device has been modified, replacing its standard stylus with a real interventional radiology needle. The 3 DOF wrist component of the stylus has been redesigned for use in an augmented reality environment.

The palpation haptic hardware that combines force and tactile feedback received positive feedback from interventional radiology experts during a face and content validation study. Its force capabilities are sufficient to reproduce a palpation procedure and can be used to provide patient variable deformation force profiles when new *in vivo* data becomes available. The mechanical arrangement of the hardware allows force to be provided from below the palpating fingertips and so functions well in an augmented reality simulation environment.

The modified Omni force feedback device provides meaningful visual and tactile feedback as a real needle hub is now grasped [201]. The modified hardware has already been commissioned for use in the award winning Imagine-S simulator (Image Guided Interventional Needle Simulation) [96], a simulator developed by the CRaIVE consortium and funded by the UK Department of Health.

The third question asked was: Is there an ideal visualisation method for a medical visio-haptic training simulation?

Chapter 2 identified that, although immersive mirror displays are the most commonly used for medical visio-haptic training simulations, the display suffers from occlusion limitations that lower the sense of immersion a user can experience. Equally, limitations of current HMD and tracking technology also make this alternative visio-haptic collocation method sub-optimal for use in a low cost medical training simulation. To overcome these problems, chroma-key technology has been investigated to produce a novel medical AR immersive workbench that replaces the immersive mirror display, whilst eliminating the inherent occlusion issues. This display can be used in conjunction with haptic hardware to provide collocated visio-haptic feedback for medical simulation.

The last of the four questions asked to address the thesis hypothesis was: Can a virtual femoral palpation and needle puncture simulation offer increased functionality over traditional training techniques?

The PalpSim visio-haptic augmented reality training environment has been designed to offer full virtual feedback. The environment is reconfigurable and can reproduce the visual and haptic cues of multiple patients, although the content and face validation has focused on a single patient simulation. The simulation design allows an atlas of *in vivo* measured patient data to be simply integrated into the simulation as new measurements are made. In contrast to a mannequin based simulation approach, needle insertions do not puncture the palpation medium as the forces felt during a needle puncture and the visible needle shaft are virtually generated, allowing multiple tissue punctures to be made without leaving markings on the model or degrading materials. A high fidelity mannequin based approach to blood flow simulation as the femoral artery is punctured requires liquid blood to be used during simulation, so time must be spent cleaning after the procedure has been simulated, although lower fidelity mannequin based blood flow simulations using LEDs have been seen. An AR simulation of this approach can offer high fidelity visual feedback, whilst allowing the simulated blood flow to be reset at the click of a button, leaving more time for training. The simulation can also provide metric information that could not otherwise be captured, although this is not currently displayed and will be the subject of future work.

PalpSim is the first fully virtual medical simulator to combine augmented reality and active haptics, allowing both direct touch and tool interaction to be simulated without the undesirable occlusion found in other collocated visio-haptic display methods.

The hardware and visualisation approaches developed here can be used to produce a training simulation that addresses patient variability. At each stage in development, the use of low-cost hardware has been a priority, whilst care has been taken not to sacrifice fidelity where possible. The overall cost of the hardware components used in PalpSim (not including computer) is approximately £2500. It is suggested that a large increase in hardware cost from that currently used, may only provide a very small gain in force feedback fidelity. The face and content validation study has shown that the developed PalpSim simulation shows promise and this supports the hypothesis made at the instigation of this project. Future work will now be described to summarise the directions the author feels that this work may take.

9.2 Future work

The initial validation steps conducted here show promise for the developed technology; however, this work represents only a first step toward producing a full procedure medical simulation for the interventional radiology procedure. As the developed simulation approaches are integrated into a full procedural simulation, further transfer of skills studies must be performed. Future validation steps of the PalpSim environment prior to its integration into a full procedure simulation are described in Chapter 8.4. The Omni needle modification is already undergoing integration into the Imagine-S [96] simulator and will be part of a task training validation study that will be performed at the Royal Liverpool University Hospital in 2011. Results from this validation can be used to further support part of the simulation hardware developed here.

Five simple to implement modifications are recommended to initially increase PalpSim's usability before integration into a full procedure simulation. These are:

- The use of a higher resolution LCD monitor. This would allow the needle shaft to be seen more clearly as the needle enters a patient's tissue. To implement this change, the simulation's field of view may need to be adjusted, depending upon the dimensions of the new screen.
- The simulation should be mounted upon adjustable display legs. This will allow users of different heights to view the display comfortably. Legs with a screw mechanism would allow for this.
- Further force calibration. Although small calibration changes have been made, it is necessary for additional force measurements to be performed in order to achieve a precise 1:1 calibration with the simulated *in vivo* forces. This would require additional time with an accurate force sensing device to measure the palpation and needle insertion forces felt at the haptic end effectors.
- Integrate the anaesthesia and skin nick stages to the procedure. This would allow the skin to be realistically nicked, producing a fuller procedural simulation with little extra work. The Omni force feedback device could be used to simulate both of these stages. A syringe could be clipped onto the needle hub to be used in the

anaesthesia simulation and a scalpel handle could be attached to realistically simulate the nicking of the skin. Dependent upon whether the skin was nicked or not, the force profile of the needle insertion could be modified, accommodating for procedural variability.

- Introduce a simple metrics screen that can be viewed in the trainer's console. This would allow the trainer to track the progress of the palpation by displaying the force and deformation applied and the progress of the needle insertion by displaying the number of punctures attempted, tissue stress and time taken. This could be analysed to spot trends between users. Further investigation into metrics may provide other evaluation criteria.

To broaden the procedural scope of the PalpSim system, wider simulation goals are considered. These are:

- Tools such as ultrasound transducers could be integrated into the simulation to simulate ultrasound guided needle puncture. This technique is sometimes used to locate the femoral artery in patients that have arteries which are difficult to locate. To simulate this effectively, an additional Omni force feedback device with a transducer shaped end effector could be introduced into the environment. Examples of simulations of ultrasound guided needle puncture using two Omni devices are [40] [95], one of which was developed at Bangor University.

- PalpSim should be integrated with a guidewire manipulation simulation. A parallel project conducted by Bangor University and other partners of the CRaIVE consortium is perusing simulation of guidewire manipulation. Integration of this with PalpSim could provide a full virtual interventional radiology training environment, allowing IR procedures to be practiced from start to finish for a variety of patients.

Although PalpSim has been shown to offer a compelling visio-haptic simulation environment, methods to increase the fidelity of both the look and feel of the palpation and needle insertion have been considered. These are:

- Incorporating an atlas of *in vivo* measured patient data. Recording the forces involved during palpation and needle insertion in a wide variety of patients would allow multiple patients to be simulated using the single hardware interface. It may be possible to identify trends in force profiles so that patient data can be inferred from the atlas, creating simulation data for virtual patients not directly represented by a set of *in vivo* measurements. In addition to this, extending force measurements from 2D force/displacement (z) data to a 4D dataset encompassing the force felt during three dimensional displacements (x, y and z), would allow more accurate force feedback to be simulated if the user moves the palpation end effector to palpate a section of tissue not faithfully represented by the 1 force DOF measurements. Machine learning techniques could be used to produce a simple representation of the 4D force/position dataset that can be integrated in a similar approach to the 2D dataset.

- Adding a 3D augmented reality visualisation. A user can currently view the 3D simulation environment in 2D, with additional depth cues used to provide an indication of the depth of the user's hands and the height of the virtual needle within the 3D scene. However, additional cues such as motion parallax, observed as the user moves their head and stereopsis, observed as a binocular disparity (see Chapter 2.5 for a review of stereo display technology), could also provide valuable depth information. To simulate motion parallax, a user's eyes could be tracked without the need for headwear using an additional camera, although it is thought that simple manipulation of the OpenGL viewing angle to correspond to the user's view through the monitor would produce an unrealistic illusion, as only a 2D view of the top of the user's hands is currently captured. Fortunately, the recent release of the Microsoft (Redmond, USA) Kinect hardware offers a potential solution to this problem at a relatively low cost (approximately £130 GBP at the time of writing). O. Kreylos [202] has used this device to combine the 2D fixed viewpoint camera image with a 3D depth map of the scene. An image can then be mapped on to the shaped objects from which they came, so parts of objects (the side of a user's hand) that would be occluded in the real world as the user moves their head to the left or right will be occluded in the virtual visualisation of the world, as the virtual camera is manipulated to follow the user's eyes. The accuracy of this approach will be critical for use in PlapSim as the curvature of the user's hands must be identified. Combining two Kinect devices, one mounted on the top right of the display and another on the top left may allow a better visualisation to be reconstructed. If this approach was deemed accurate, a stereo visualisation could also be provided using the shaped hand image.

- Integration of alternative actuation methods. The developed AR display technology offers a potential visualisation method for the simulation of open surgery procedures. Future technologies that may provide haptic feedback for open procedures such as the noncontact ultrasound based tactile display [20] will be carefully observed for future work.

The intuitive and highly immersive AR environment presented in this thesis has great potential as a platform for training clinical procedures. It is particularly suited to scenarios where a variable patient habitus is required, and where the position in which a surgical tool or the surgeon's hands are applied is arbitrarily chosen.

The novel approach of concealing suitably designed haptic hardware below the projection plane, whilst still displaying a realistic view of the trainees hands and tools, not only provides the ability to enhance the achievable immersion of existing simulations but in the future, enables the simulation of more complex tasks such as open surgery procedures. Effective open procedure simulation will require sophisticated tactile, force and visual feedback and this research has identified and started to solve some of the challenges that will be involved.

The development of novel force and tactile feedback hardware will be pivotal in the progression of medical simulation. Prior to this work, only tool or single point thimble interaction had been simulated for the palpation and direct touch of a fully virtual patient's skin. Simulation of open procedures, recreating the feel as a practitioner reaches into a patient, will require an innovative rather than incremental progression in haptic feedback technology, although new technology such as the ultrasonic non-contact tactile display [20] shows promise and could be simply integrated into an AR simulation environment such as that used in PalpSim. An ideal haptic technology will provide both high force to resist the user's motion in combination with fine tactile stimulation and importantly, will be affordable, facilitating its use in low cost training environments.

In conclusion, this thesis demonstrates that an augmented reality approach to simulation can be used to merge the virtual and real worlds into one seamless visio-haptic experience, and it is confidently expected that this technology will be used in the next generation of medical simulations, becoming indispensible as increasingly complex procedures are simulated. PalpSim is the first such example of a fully functional AR visio-haptic medical training platform.

10Appendix

10.1 Task analysis of an arterial puncture

Task analysis [182] courtesy of the CRaIVE consortium. Arterial Puncture Task Analysis performed by Sheena Johnson, University of Liverpool.

Step 10: Locating femoral artery

10.1. Pick up 5ml local anaesthetic syringe and place on sheet covering patient (to ensure it is easily accessible when required)

10.2. Palpate patient to locate anterior superior iliac spine and pubic tubercle (to find inguinal ligament to mark the mid point where the external iliac artery becomes the common femoral artery. The common femoral artery must be located initially by palpation to do the puncture).

10.3. Feel carefully for pulsations in artery. Is it located ok?

Yes (go to step 10.9)
No (go to step 10.4)

10.4. Place metal marker in region of proposed incision site

10.5. Ask radiographer to move the image intensifier into position over patient

10.6. Screen using foot pedal (see exposing fluoroscopy algorithm)

10.7. Use picture on screen to relate marker to anterior superior iliac spine, pubic tubercle, femoral head and any vascular calcification.

10.8. Feel in correct anatomical location for pulsation of the femoral artery. Located ok?

Yes (go to step 10.9)
No – possible reasons are; (a) calcification (b) obese patient (c)scarring of groin (d) upstream obstruction. (repeat step 10.4, other possibilities

include ultrasound, blind puncture, feel the artery as a tube, use x-ray image to identify calcium deposits in artery – cross over methods with contrast and roadmap). If unable to locate artery this is a complex case

10.9. Feel more proximal and distal to the strongest palpable arterial pulsation to ascertain the direction of the artery

10.10. Locate the centre line of the artery by feeling both the medial and lateral sides of the artery

10.11. Position fingers to fix artery with fingers on either side of vessel (the aim is to puncture into mid line of artery). Complications of not puncturing the mid line of the artery; (a) guide wire entering into wall of artery rather than travelling along the artery (b) occlusion of artery (c) damage of artery wall

10.12. Press down with fingers (especially in large / obese patients in order to displace fat and get closer to the artery, pressing too hard may result in occluding (blocking) the artery and loss of the pulse)

10.13. Feel for pulse. Does pulse feel ok?

Yes (go to step 11.1)
No (complex case)

Step 11: Puncturing artery

11.1. Position vascular access needle on incision site – between 2 fingers pressing down (see step 1.11), Bevel uppermost (see holding a vascular needle algorithm)

11.2. Insert needle through the nick in the skin at a 45 degree angle towards artery (with the orifice on the bevel of the needle pointing upwards and forwards so the wire can exit easily)

11.3. Feel the artery pulsation using non needle holding hand and align the needle trajectory with the artery

11.4. Advance the needle towards the artery.

11.5. Is there any indication from patient that more local anaesthetic is needed?

Yes (insert more local through arterial puncture needle and go to step 11.6)

No (continue to step 11.6)

11.6. Feel for the artery pulsating through the needle. Can you feel pulsation?

Yes (indicates near artery, go to step 11.7)

No (reposition needle and repeat step 11.6)

11.7 Puncture artery with either:

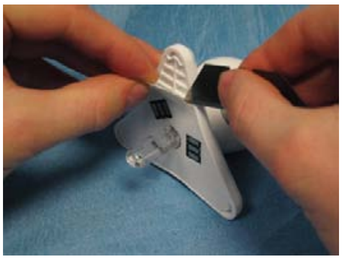
- A sharp stab
- Gently increase pressure

11.8 Immediately but gently decrease angle between needle and patient

10.2 Re-wiring a Falcon Grip

A Falcons grip is wired into the devices base so that any end effector can be used.

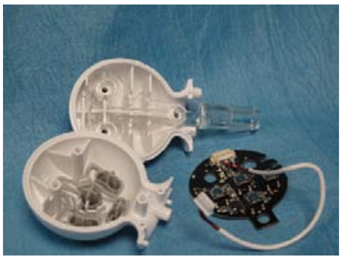
1.The necessary components can be removed from a Falcon ball grip



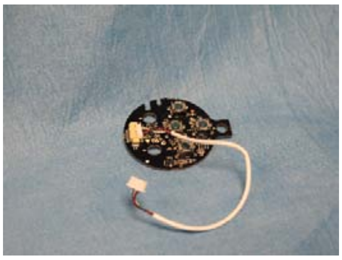
2.Peel back the sticky plasctic cover concealing three retaining screws

3.Remove the screws and detatch the five pin connector labeled P1

4.Remove the triangular base and another three screws



5.Open and remove the circuit board leaving the wire connected



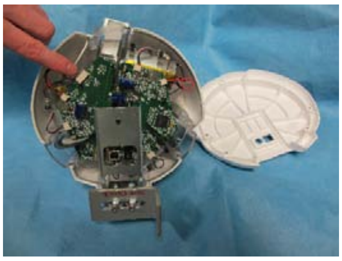
6.Discard all but the circuit board, remove the unatached connector.

7.A Novint Falcon with a 90 degree rotated mount is shown here

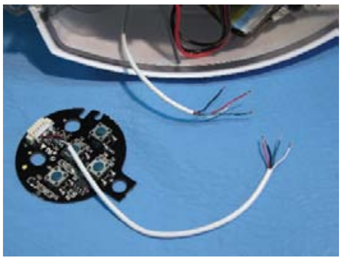
8.The silver dust protectors can be unclipped with a screwdriver

9.Six screws must be removed so the the back cover can be detatched

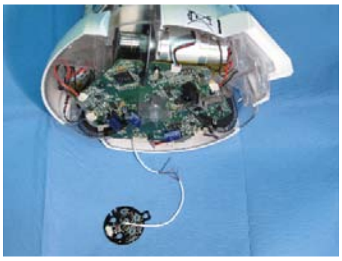
10.The eight pin connector leading from the end effector is labeled P2



11.Four wires connector the disconnect for main circuit board

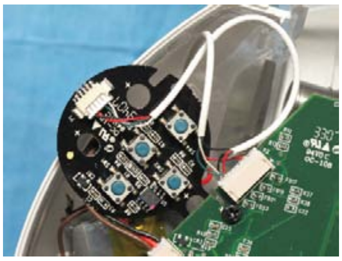


12.The connector must be removed and the wires spliced together



13.Wire connections from “P1” away

Device	End Effector
Black	White
Red	Green
Green	Red
White	Black



14.The newly attached end effector can be placed inside the device

15.The device can now be closed and will function without a standard Novint end effector connected

10.3Face and Content Validation Questionnaire

Many thanks are extended to S. Johnson, H. Woolnough and C. Hunt, psychologists from the University of Manchester for providing prior questionnaires and a detailed task analysis upon which these questions are based.

Femoral Artery Palpation Simulator

a. Initials or leave blank: ______________________________

b. Gender: Male ☐ Female ☐ **Training:** Consultant ☐ Trainee ☐

If trainee which year: 1☐ 2☐ 3☐ 4☐ 5☐

c. Years of experience in Interventional Radiology: 0 / 1 / 2 / 3 / 4 / 5 / 6-10 / 11-20 / 20+

d. On average how many procedures involving the Seldinger technique would you complete in a month?
0 / <1 / 1-2 / 3 -5 / 6-10 / 11-20 / 21-50 / 50+

e. On average how many <u>other</u> interventional procedures (nephrostomy, biopsy, abscess drain etc) would you complete in a month? 0 / <1 / 1-2 / 3 -5 / 6-10 / 11-20 / 21-50 / 50+

		Very good	Good	Average	Poor	Very poor
1	**How would you rate your general Interventional Radiology skills compared to someone else at your level of expertise?**	1	2	3	4	5
2.	**How would you rate your performance on this simulated procedure compared to your usual real life performance?**	1	2	3	4	5

The simulator is realistic, compared to a real life procedure, in the following areas:

		Very strongly disagree	Strongly disagree	Disagree	Neutral	Agree	Strongly agree	Very strongly agree
3.	**The appearance and position of the virtual patient is realistic**	1	2	3	4	5	6	7
4.	**The feel and location of anatomical landmarks**	**THESE ARE NOT REPLICATED IN THIS SIMULATION**						
5.	**Appearance of the draped access site looks real**	1	2	3	4	5	6	7
6.	**The location of the pulse in the virtual environment is correct and realistic**	1	2	3	4	5	6	7
7.	**The feel (volume, waveform) of the pulse is realistic**	1	2	3	4	5	6	7
8.	**Using local anaesthetic to make a small skin nick**	**THIS IS NOT REPLICATED IN THIS SIMULATION**						
9.	**Your hand and its motion in the simulation is realistic**	1	2	3	4	5	6	7
10.	**You can clearly see the location of the needle in the simulation**	1	2	3	4	5	6	7
11.	**You can find, feel and pick up the needle realistically.**	1	2	3	4	5	6	7
12.	**The feel of inserting the needle into the tissues is realistic.**	1	2	3	4	5	6	7
13.	**The 'feel' of entering the artery wall is realistic.**	1	2	3	4	5	6	7
14.	**The appearance of the blood jet from the needle shaft is realistic**	1	2	3	4	5	6	7
15.	**Force and colour of the blood jet from the needle is realistic.**	1	2	3	4	5	6	7

Femoral Artery Palpation Simulator

16.	**The 'feel' of manipulating the needle (e.g. depressing the hub before wire introduction) is realistic. NOTE: wire introduction is not simulated**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
17.	**The visualised response of the needle to manipulation, as seen on the screen, is realistic.**	**1**	**2**	**3**	**4**	**5**	**6**	**7**

18. How often do you play computer games of any kind?

Daily	Weekly	Monthly	Once every 6 months to a year	Never
☐	☐	☐	☐	☐

19. How would you rate your computer or IT skills?

Very Good	Good	Average	Poor	Very poor
☐	☐	☐	☐	☐

20. Have you used a medical simulator in any type of training program?

Yes ☐
No ☐

20.a If yes, please provide details here e.g. the name of the simulator: ______________________

____

After practise on the Femoral Artery Simulator, what are your views on the following?

		Very strongly disagree	Strongly disagree	Disagree	Neutral	Agree	Strongly agree	Very strongly agree
21.	**This simulator is effective for practising basic skills**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
22.	**This simulator would be useful for evaluation during training**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
23.	**This simulator offers a user-friendly learning environment**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
24.	**This simulator is fun to use**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
25.	**This simulator can shorten learning curves**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
26.	**This simulator can reduce complication rates**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
27.	**This simulator gives Interventionalists a sense of confidence**	**1**	**2**	**3**	**4**	**5**	**6**	**7**
28.	**This simulator could reduce the expenses of training after purchase**	**1**	**2**	**3**	**4**	**5**	**6**	**7**

29. If you have any further comment to make on the simulator please feel free to enter it below. We are interested in all feedback, both positive and negative.

Thank you for completing this questionnaire.

10.4 Open Ended Questionnaire Feedback

Question 29 of the PalpSim content and face validation questionnaire allowed practitioners to give open ended feedback. This feedback both oral and written is given below in the category to which it reefers.

Comments about the palpation simulation

"Left hand feel is brilliant"

"I like the palpation, it does feel like a real palpation"

"The palpation feels very realistic"

"Palpation feel really good"

Comments about the needle insertion simulation

"I keep thinking I am going to prick myself, I had to check again there was no needle"

"Needle hub feel is good"

"Some movements of the needle are not totally realistic, some resistance in "air", some excess freedom in the tissues"

Comments about the shadows

"The shadows were very helpful"

"Shadow of needle and hand are great"

Comments about the AR environment

"The screen view is great"

"I like the setup"

"I really like the environment"

General comments

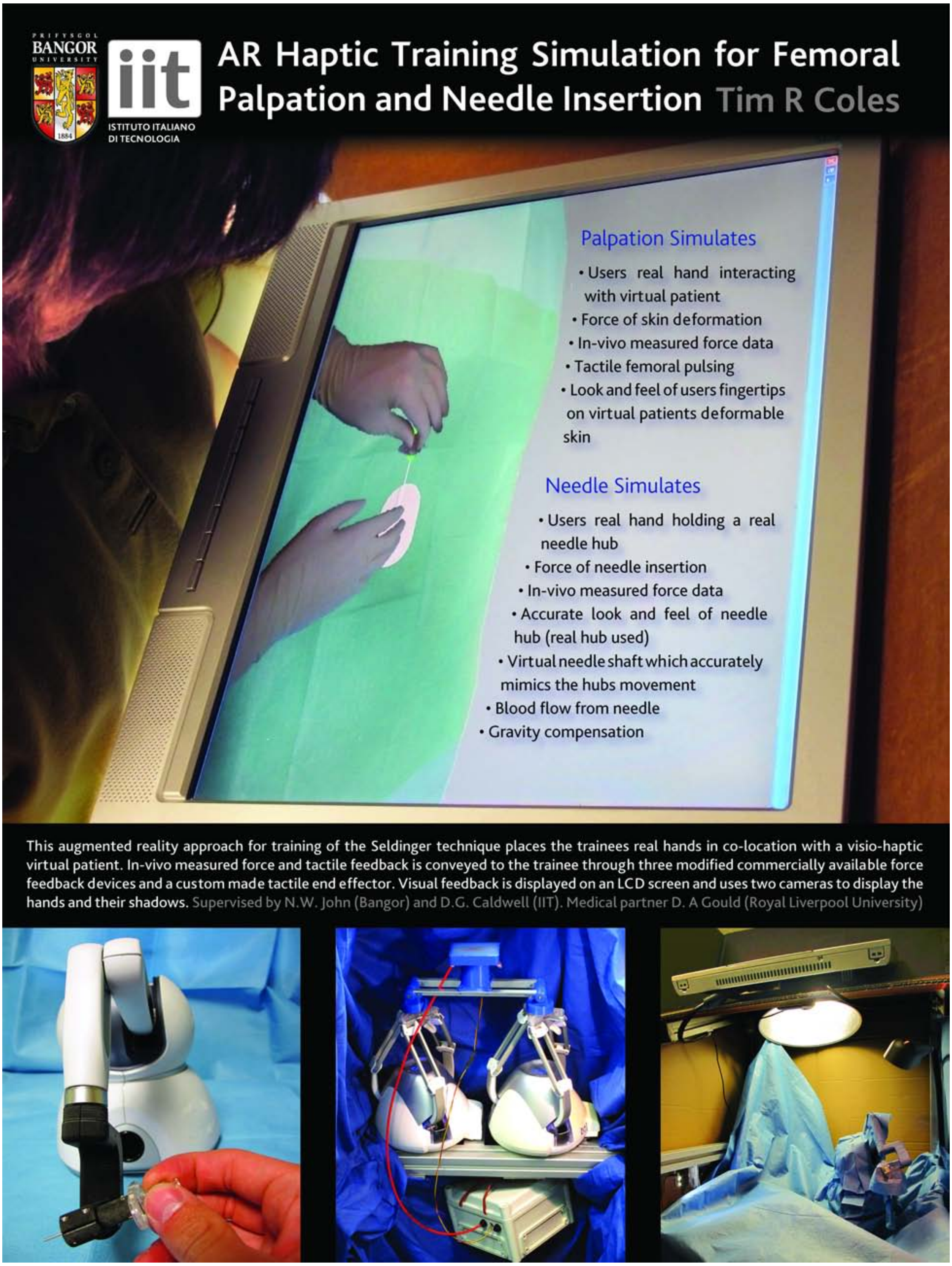
"It's very nice work, well done"

"Great progress, well done". A comment comparing previous mannequin based simulation attempts from the CRAiVE group. The practitioner had not previously tried the PalpSim simulation.

"Really nice, good stuff"

10.5North American Summer School Poster

A poster presented at the 2010 North American Summer School on Surgical robotics and Simulation.



11 Glossary

A

Acupuncture -	A traditional Chinese theraputic practice involving the use of needles that is used for many treatments.
Artery -	Blood vessels that carry blood away from the heart.
Arthroscopy -	A minimally invasive procedure for examination and treatment within a joint.

B

Brachial artery -	An artery located in the upper arm.
Brachytherapy -	A practice in which a radiation source is placed inside or next to an area requiring treatment, often used to treat cancer of the prostate, breast, cervix and skin.

C

Cadaver -	A dead human body.
Cannula -	A hollow flexible tube inserted into the body. Typically a trocar is used to insert the cannula.
Cardiology -	A medical field specialising in disorders of the heart.
Cholecystectomy -	A typically minimally invasive procedure to remove a patient's gallbladder.
CRaIVE-	Collaborators in Radiological Interventional Virtual Environments, a consortium comprising of clinicians, physicists, computer scientists, clinical engineers and psychologists. Visit www.craive.org.uk.

D

DOF -	Degrees Of Freedom. A solid real world object in free space will have six degrees of freedom. Three translations (x, y and z) and three rotations (roll, pitch and yaw). See also "force DOF".

E

Endoscope - A rigid or flexible tube with a lens and light system that allows a practitioner to see the interior of hollow organs.

Endoscopy - Refers to the practice of looking inside a patient's body with an endoscope.

Endovaginal ultrasound – Also known as transvaginal ultrasound. A procedure commonly performed for assessment of early pregnancy.

Epidural - A form of anesthesia where drugs are injected into epidural space, the outermost part of the spinal canal. Commonly used during labour and for operations such as knee and hip replacements.

F

Fluoroscopy - An imaging technique used to obtain real-time images of the internal structures of a patient.

Force DOF - Actuated Degrees Of Freedom. See DOF.

FOV - Field Of View

G

GPU - Graphics Processing Unit.

Gynecology - A medical field specialising in the female reproductive system.

H

Habitus - A patient's physique or body build, relating to their muscle and fat mass.

HMD - Head Mounted Display.

Hysteroscopy - An endoscopic inspection of the uterus.

I

IR - Interventional Radiology. See Chapter 3.

K

Keyhole surgery - An alternative name for Laparoscopic surgery.

L

Laparoscopic Surgery - Sometimes called minimally invasive surgery (MIS). See Chapter 2.8.3 for a full explanation.

Lumbar puncture - A procedure used to sample cerebrospinal fluid from the spinal canal.

M

MIS - Minimally Invasive Surgery. See Laparoscopic Surgery.

Ms - Abbreviation for milliseconds.

N

N - Abbreviation for Newton(s)

Neurology - A medical field specialising in the nervous system.

P

Palpation - A method of examination in which a practitioner presses upon a patient's skin to determine the shape, size and stiffness or location of a patient's organs. In the case of femoral palpation a clinician presses lightly on the skin of a patient's groin area to locate the femoral artery.

R

RapidForm - A software product developed by INUS Technology Inc. (Seoul, Korea) designed for the manipulation of 3D point clouds captured from 3D scanners.

T

Trocar - A sharp-pointed surgical instrument, used with a cannula to puncture a body cavity or in laparoscopic surgery, used to introduce tools into a patient's body.

3D - Three-dimensional.

3D Studio Max - A software product developed by Autodesk (San Rafael, USA) for 3D modelling, animation and rendering.

V

Venipuncture - A needle puncture to access a patients vein. Commonly used to obtain a sample of a patient's blood.

Vertebroplasty - A minimally invasive procedure performed to bind spinal fracture components.

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