

MRI-compatible teleoperated needle insertion with haptic feedback

W.B. (Willem) Hoitzing

BSc Report

Committee:

Prof.dr.ir. S. Stramigioli
Dr.ir. M. Abayazid
Dr. H.J.B. Muijzer-Witteveen

July 2017

019RAM2017
Robotics and Mechatronics
EE-Math-CS
University of Twente
P.O. Box 217
7500 AE Enschede
The Netherlands

Abstract

In this assignment a prototype of a 1-degree-of-freedom teleoperated master-slave system was developed. The slave is able to insert a needle into a phantom inside an MRI-scanner, while the master provides the user with haptic feedback. A PHANTOM Omni (Sensable Technologies) serves as a master providing force feedback to the user. The slave is a pneumatic stepper motor, equipped with a force sensitive resistor. The master and slave were interfaced with MATLAB R2015b running on a desktop and an Arduino Uno. The MRI-compatibility of the slave and the implementation of haptic feedback were evaluated. Images taken with a 0.25T scanner showed reasonable image quality, but no full MRI-compatibility was achieved. A user's response study revealed basic haptic feedback was implemented successfully, but the transparency while inside the phantom was inadequate. Results showed the used sensor proved to be a bottleneck for the system's functionality, calling for a better MRI-compatible force sensing alternative.

Contents

1	Introduction	4
1.1	Aim	4
1.2	Research Plan	4
1.2.1	Workflow	5
1.2.2	Contribution and Novelty	5
1.3	Organization	5
2	Literature Review	6
2.1	Haptic Feedback	6
2.1.1	Kinesthetic Feedback	6
2.1.2	Tactile Feedback	8
2.2	MRI-Compatible Force Sensors	9
2.2.1	Fiber-Optic Force Sensors	9
2.2.2	High Elasticity Fabric Force Sensor	11
2.2.3	Pneumatic Pressure Deduced Force Estimation	11
3	Materials and Methods	13
3.1	Teleoperated System	13
3.1.1	Master	13
3.1.2	Slave	14
3.1.3	Interface	16
3.1.4	Remote Control	16
3.1.5	Implementation Haptic Feedback	17
3.2	Experimental setup	18
3.2.1	Validation Actuator Displacement and Velocity	18
3.2.2	Calibration Force Sensor	20

3.2.3	Evaluation MRI-compatibility	20
3.2.4	Evaluation of Haptic Feedback	21
4	Results	23
4.1	Validation Actuator Displacement and Velocity	23
4.2	Calibration Force Sensor	24
4.3	Evaluation MRI-compatibility	25
4.4	Evaluation of Haptic Feedback	25
5	Discussion	30
6	Conclusion	32
A	Overview Fiber-Optic Force Sensors	36

1 Introduction

Currently, clinicians performing biopsies or ablations can make use of one of several imaging modalities such as Magnetic Resonance Imaging (MRI) [21], Computed Tomography (CT) [4], and Ultrasound (US) [7]. These modalities provide visual feedback that can be used to accurately position the needle with the help of a robotic system, while keeping the procedure minimally invasive. Even palpations, traditionally performed by hand [25], are carried out by machines in some research [19] using sensors designed to be capable of sensing inside for example the challenging environment of an MRI- or CT-scanner.

Image-guided robotic needle insertion systems are not used in the current clinical field [15]. This may be because they are simply not available, as compatibility with an imaging technique can be challenging to achieve. Besides, safety concerns arise with a fully actuated robotic system where a human operator is not directly controlling their instrument in the loop [20].

1.1 Aim

This research focuses on MRI-guided needle insertion. The aim is to develop a prototype, capable of safely and remotely inserting a needle into a phantom in an MRI-environment. The system controlling the distant actuator, called the master, should provide the user with feedback on the needles environment and motion. Such feedback in this scenario is known as haptic feedback, and would increase the safety of the system. In order to allow for haptic feedback, the slave's end-effector should be equipped with an MRI-compatible force or pressure sensor. Real-time operation of the system, including haptic feedback and control of the slave, is required.

1.2 Research Plan

One of the many applications of using haptic feedback is to improve teleoperation in Magnetic Resonance Imaging (MRI). Think of MRI-guided biopsies to accurately steer the needle into position. As the setup of an MRI system leaves very limited open space for a surgeon to perform any operations, the use of MRI-guided invasions call for the accurate and dexterous tools of surgical robotics. Teleoperating such a surgical robotic system in turn calls for haptic feedback to improve the surgeons perception of the end-effector. As previously mentioned, a teleoperated surgical robot will consist of a master-slave system. The master can be placed outside the MR-bore where it will not interfere with the MRI's functionality independent of its type of system. The slave system, however, has to be placed at least partially inside the MR-bore. This requires (parts of) the slave system to be MRI-compatible. MRI-Compatibility implies the ability to function well within an MR-bore, as well as not to interfere with the scanner's imaging quality. At a bare minimum, only the slave's end-effector could be placed inside the bore and be manipulated with actuators outside the sensitive area. Concerning the MRI-compatibility of the slave's actuator, possibilities

are: an electromagnetic actuator placed outside the bore and a pneumatic or hydraulic actuator that can be placed inside the bore.

1.2.1 Workflow

A current clinical workflow for liver biopsies performed in a closed-bore high-field MRI system has been reported in 2016 by Moche et al. [15]. Due to the surgeon’s inability to get real-time feedback on their needle positioning, an iterative and time-consuming approach is reported. Initially the patient will be in the MRI and a rough access point and corresponding orientation is assessed. The next step is to turn off the scanner and move the patient out of the bore again to be able to perform the previously determined operation. Following these cycles of scanning and positioning, the needle is step-by-step guided through the lesion.

The expected workflow after integration of a teleoperated surgical robot with haptic feedback in a closed-bore MRI system would have greatly increased efficiency. Once the patient is being scanned inside the MRI, the surgeon can start positioning the needle through the master-side of the robot. Making use of real-time MRI and the accuracy of a surgical robotic system, the needle can be guided through the lesion towards the targeted tissue with greater precision and in less time while the patient is still inside the MRI’s closed bore.

1.2.2 Contribution and Novelty

This research serves as a proof of concept which, if successful, may develop into an MRI-compatible teleoperated surgical robotic system with haptic feedback used for needle insertion. Individual aspects of the entire system may be incorporated in other systems. The force or pressure sensor’s MRI-compatibility is a relevant aspect for other applications within the medical field.

1.3 Organization

The report follows the outline as discussed here. After the introduction, section 2 includes a literature review on haptic feedback and MRI-compatible force or pressure sensors. Section 3 presents the system, its components, and the used methods for conducted experiments. Next, the results from these experiments are shown in section 4. The discussion in section 5 describes the limitations of the system and some points for future research, referring to the results in the previous section. Lastly, section 6 discusses the conclusions drawn from the results with respect to the objective.

2 Literature Review

In this section, a concise review on haptics in robotics, and available MRI-compatible force sensors is presented.

2.1 Haptic Feedback

As industrial robotics are still developing, so is the surgical application of these robots. The earliest use of a surgical robot was in 1985, where it was used to accurately guide a needle for a brain biopsy [6]. This was the start of a new era, as surgical robotics promise the surgeon better accuracy and dexterity than the human equivalent.

In robotics, a system is said to be teleoperated when it is operated from a distance. In surgical robotics this usually takes the form of a master system being controlled by a user - the surgeon in our case - and a non-rigidly connected slave following every command the master gives. A controlled end-effector (e.g. a needle) would be on the slave-side of the entire system. A frequently reported flaw of these systems is their lack of haptic feedback [17].

Proper haptic feedback is a key aspect to the safety of a teleoperated surgical robotics system. Direct interaction between instrument and patient is no longer sensed by the surgeon due to the remote-control nature of the system [17]. Haptic feedback supplies the surgeon with cues about the end-effector's motion, giving them some insight into their actions. In practice, haptic feedback takes multiple forms which will be discussed here.

2.1.1 Kinesthetic Feedback

a) *Concept*

Kinesthetic feedback, also referred to as force feedback, supplies the master system with a force that should be exerted on the user. Ideally, on impact of the slave's end-effector, the master system would mimic a sensation similar to the sensation felt by the user if the master system was rigidly connected to the end-effector [10]. Typically, the forces the end-effector apply to the patient are either measured or calculated [17]. After processing, a resolved force to be felt by the user is realized by the actuators of a haptic display.

b) *Properties*

Using force feedback, object-collision may be mimicked. Any surface collision can be detected and a resolved force calculated. This resolved force can differ in size and direction based on the surface's stiffness and orientation. However many degrees of freedom the master system has, can also be used in force feedback. Any actuators realizing movement in these degrees of freedom can be (partially) blocked by exerting an opposite force to the user's input in order to imitate the sense of bumping into a wall.

c) *Application*

As said before, force feedback works well for giving the user sensations related to collisions. Some practical applications of this in surgical robotics would be sensing a needle puncturing the skin or a muscle [9], or running into a bony structure. It is also a preferred feedback mechanism for suture knot tying, promoting the user's sense of gripping force [17]. Force feedback is also used in training simulations for surgeons operating on a virtual patient [17].

These are some examples of haptic devices that integrated force feedback:



Figure 1: SensAble Technologies' PHANTOM Omni, part of the PHANTOM® series [27, 2].



Figure 2: The Omega 6 by Force Dimensions, which developed the Sigma.x, Omega.x and Delta.x series [27, 1].



Figure 3: The CyberGlove [11].



Figure 4: The Da Vinci surgical system [12].

Unlike the two previously mentioned series, the CyberForce and CyberGrasp by CyberGlove are no 'desktop robots', but a tight-fitting glove with force feedback to each individual finger [27, 11].

Although the Da Vinci surgical system is a surgical robot rather than a haptic device, research has been done using force feedback integrated in the Da Vinci robot [16].

2.1.2 Tactile Feedback

a) *Concept*

Whereas kinesthetic feedback uses brute forces (partially) restraining certain movements, tactile feedback relies on receptors closer to the skin's surface. These receptors sense pressure, heat and vibration to enable detection of texture and more subtle shapes of objects [13].

b) *Properties*

Tactile feedback enables the user to feel the end-effector's contact location and pressure distribution [17]. This type of feedback allows for either intuitive or more symbolic feedback. Think of localized vibrating nodes that can pinpoint a measured contact point on the end-effector. On the other side of the spectrum, the symbolic feedback, the possibility to code certain events or states of the end-effector into specific vibration patterns in the haptic display.

c) *Application*

An example of medical application of tactile feedback [17] would be the detection of several indicators of tissue health, such as compliance, viscosity and surface texture. A surgeon could also directly interpret the supplied pressure distribution or tissue deformation.

There are multiple different devices using tactile feedback as their means of transferring a haptic sensation to the user [10]. The simplest is a vibrotactile device, whose function can be compared to the buzzing of a mobile phone. Whereas force feedback directly mimics a force exerted on the end-effector, vibrotactile feedback focuses more on the perception and interpretation of certain vibrotactile codes or stimuli. Another tactile device is a surface display. The idea here is that a real surface is moved under a user's finger. In fact this device might not transfer a measured or processed haptic sensation as seen before, but instead it attempts to recreate the end-effector's surroundings. The last device is called a tactile display, which provides sensations that are spatially distributed.

An already researched application of haptic feedback would be palpation [17]. Successful experiments have been conducted using force feedback [14], but it is also suggested that fingerpad deformation plays an important role in palpation [18]. This latter observation implicates an incentive to use tactile feedback for the purpose of palpation.

As previously stated, force feedback has been integrated in the Da Vinci surgical system, but also an extension of the robot including tactile or vibratory feedback has been made [3]. This in vivo study provides promising results concerning the use of tool vibration feedback during robotic minimally invasive surgeries.

2.2 MRI-Compatible Force Sensors

To supply haptic feedback to the surgeon, the slave should also be able to gain insight into the forces and/or pressure the needle experiences. MRI-compatible force or pressure sensors are available, but there are multiple types to choose from:

2.2.1 Fiber-Optic Force Sensors

One type of force sensor is the so-called fiber-optic force sensor [23], making use of an interaction between a deformation and different aspects of light. Three types of fiber-optic force sensors are distinguished as they each use a different aspect of light to correlate with deformation (and hence force). These three distinct types are listed below:

2.2.1.1 Intensity Modulated Fiber-Optic Force Sensors

Fiber-optic force sensors based on intensity modulation rely on a change in light-intensity due to a force. This can be interpreted as a distance-dependant attenuation where the distance the light travels is affected by a force-induced deformation. There are two different variants of intensity modulated sensors: reflective and transmissive sensors. Most importantly they differ in the sense that for a reflective sensor, a light source and intensity sensor are placed on the same side of the force sensor, whereas a transmissive sensor places one on either side of the force sensor. Figure 5 shows the basics of these variants on the basic principle of intensity modulation.

The simple design, low cost, and easy signal interpretation of an intensity modulated fiber-optic force sensor does have its drawbacks. Light source instability, fiber bending or fiber mismanagement can cause intensity fluctuations and light exiting the fiber causes optical loss. Several solutions have been proposed to overcome signal instabilities (see review of Su et al. [23]).

2.2.1.2 Wavelength Modulated Fiber-Optic Force Sensors

Another fiber-optic force sensor is based on wavelength modulation. It is often referred to as a fiber Bragg grating or FBG sensor. If a load is applied on the sensor, these gratings will change and cause a different wavelength to be reflected back from the one that is sent towards the sensor through an optical fiber. Equation 1 shows an expression for the wavelength shift change, where λ_B represents the reflected wavelength shift, k_ϵ the coefficient for strain ϵ and k_T the coefficient for the temperature change ΔT .

$$\frac{\Delta\lambda_B}{\lambda_B} = k_\epsilon \cdot \epsilon + k_T \cdot \Delta T \quad (1)$$

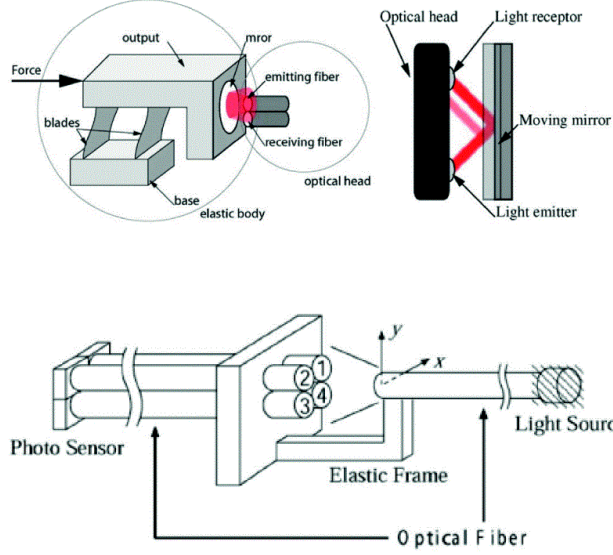


Figure 5: The two implementations of intensity modulation implementation: reflective (upper) and transmissive (lower) [23].

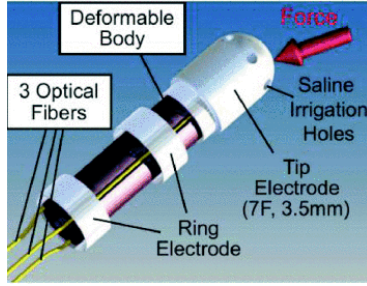


Figure 6: FBG sensor Tacti-Cath [28].

Figure 6 shows an example of a wavelength modulated fiber-optic force sensor used for atrial fibrillation. Compared to an intensity modulated sensor, an FBG sensor can achieve better resolution. Another benefit is the usage of small diameter fibres. Elayaperumal et al. [5] have managed to integrate an FBG sensor into an MRI-compatible needle. Due to the temperature-sensitive nature of an FBG sensor, it does require adequate compensation. Key drawbacks of an FBG sensor are its costs and complicated system setup.

2.2.1.3 Phase Modulated Fiber-Optic Force Sensors

Phase modulated fiber-optic force sensors are based on measuring a relative phase shift between light beams through interferometry. A phase shift can be related to a difference in optical paths that different beams take. A typical sensor uses a multiple beam configuration such as Fabry-Perot interferometers (FPI). The following description is given by Su et al. [23]:

'Inside a Fabry-Perot strain sensor, light propagates between a pair of par-

tially reflective mirrors that form a Fabry-Perot cavity. A portion of light exits and the rest remains inside the cavity. Multiple beams with different optical path lengths exiting the Fabry-Perot cavity are superimposed, generating destructive and constructive interference that can be observed in the spatial domain or spectral domain.’

Observing interference, or rather, a change in interference patterns, a phase shift can be seen. This phase shift relates to a strain caused by an axial force. FPI sensors’ advantages are their high resolution and large operable temperature range. A relatively low-cost FPI sensor is one designed by Shang et al. [22], but their premium counterparts can be very costly. A drawback for an individual FPI instrument is the possibility it may require repeated calibration over time. Su et al. reviewed fibre-optic force sensors and produced an overview of different designs and some aspects like their working principal, sensory range and resolution, see Figure 27 in Appendix A.

2.2.2 High Elasticity Fabric Force Sensor

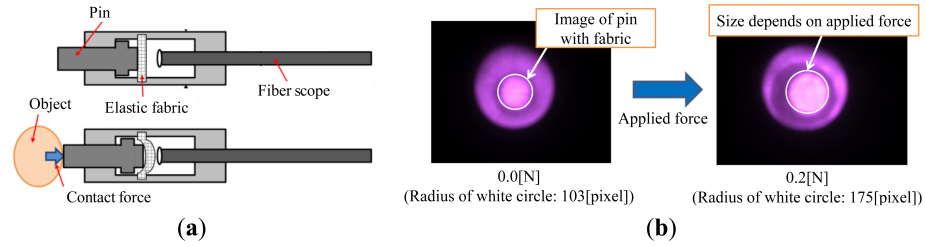


Figure 7: (a) Schematic overview of working principle, (b) visualization of deformation caused by an applied axial force [26].

A force sensor that relies on a force transformed into deformation which is visualized by an optical fiber was developed by Watanabe et al. [26]. As can be seen in Figure 7, an axial component of a contact force will deform a highly elastic fabric. In the visualization by the fiber scope, a bigger area will be highlighted due to the deformation. The radius of the highlighted circle is related to the applied force. The experiment conducted used pantry stocking fabric and tested applied forces ranging from 0 to 0.2 N. Results state a high resolution of < 0.01 N.

2.2.3 Pneumatic Pressure Deduced Force Estimation

A 4-DOF forceps system with force sensing [24] has also yielded satisfactory results. In this study, a master-slave system with pneumatic actuators was used. Calibrating the system without a load for different position signals produced a so-called neural network which specifies system dynamics. When the master system is manipulated, a certain reference driving force is then sent to the slave system’s PD and feedforward controller to be realized. When the end-effector

is experiencing external forces, resisting its movement, additional air pressure (compared to the neural system's reference values) is required to compensate for these unexpected factors. It is assumed the following relationship between force F , pressurized area A , and differential pressure ΔP in the actuator's cylinder is abided by: $F = A \cdot \Delta P$. When the slave's pressure signal does not correspond to the neural network's reference value, it is attributed to the external force and calculated according to the previously mentioned relationship.

Noteworthy is how this sensing method does not rely on mechanical deformation, but steering signals of the teleoperated robotic system. It should be considered that when making use of a pneumatic stepper motor, varying pressure signals do not relate to an external force.

3 Materials and Methods

This section lists the materials that have been used to create the setup, as well as some methods used to measure or process data, or to conduct an experiment.

3.1 Teleoperated System

The system consists of three environments: the human environment, the control interface, and the MRI-room. The human operator sends control signals from the human environment, through the control interface, to the MRI-room. In the MRI-room, these signals are expressed in the form of linear motion, attributing to the insertion of a needle. The MRI-room responds with a signal correlating to an axial force, measured at the actuator's end-effector. As the control interface receives these signals from the slave in the MRI-room, it reacts by sending a control signal towards the master, located in the human environment. This control signal prompts the master to provide the user with adequate force feedback. Figure 8 shows the aforementioned environments, with the corresponding individual components placed inside these environments. The flow of signals between components is shown with labeled arrows pointing in the direction of this flow. The following sections go further into detail on some of the important individual, or combination of, components.

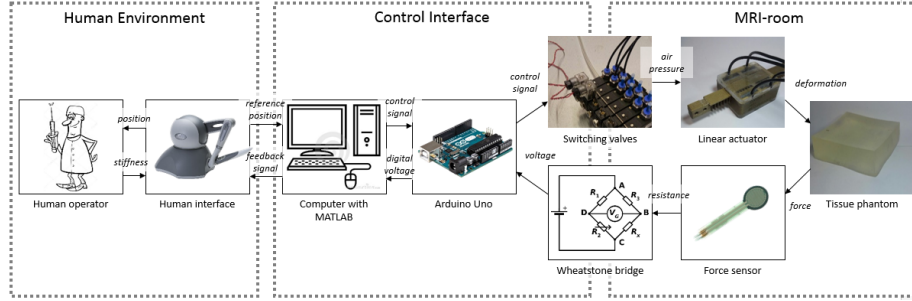


Figure 8: Block diagram of the complete teleoperated system.

3.1.1 Master

The master system should be capable of sending control commands to the slave system, while at the same time providing haptic feedback to the user. To achieve this, a PHANTOM Omni, from SensAble Technologies was used. The Omni is a haptic device with the ability to provide force feedback to the user (see Figure 9). In this study, the Omni is communicating with the computer with an IEEE-1394 FireWire connection. Figure 9 shows the three rotational joints capable of providing the user with force feedback.



Figure 9: The PHANTOM Omni, with the three joints capable of providing force feedback labeled.

3.1.2 Slave

As previously mentioned, the slave system should be able to actuate a needle and push it into a phantom, while operating inside an MR-bore. Meanwhile it should take record of the forces the needle experiences during insertion or retraction to be provided to the user as haptic feedback through the master system. A pneumatic linear stepper motor as designed by Groenhuis and Stramigioli [8] is used (see Figure 10). The specifications for the used stepper motor are found in Table 1. The actuator uses two pistons in a double-acting cylinder, hence four pneumatic tubes are used. These tubes are 3m long, with a 1mm inner diameter. The air supply is switched between the two sides of the piston using solenoid switching valves. A quadrature-encoded waveform acts as a control signal for forward or backward movement. Refer to the paper of Groenhuis and Stramigioli for more information on the design and operation of this type of pneumatic motor.

Table 1: Specifications for the pneumatic linear stepper motor.

supplied air pressure	6 bar
step size	1 mm
max. velocity	15 mm/s

The linear actuator's end-effector should hold the insertion needle, whose experienced forces need to be deduced. As this research serves as a proof of concept, instead of a needle, a 3D-printed cylinder with a 1 cm diameter acts as a contact-point. To measure the axial forces acting on the cylinder, an Interlink FSR400 was used. Due to the bad availability and high costs of the other MRI-compatible force sensing options as researched in section 2.2, a merely MRI-safe

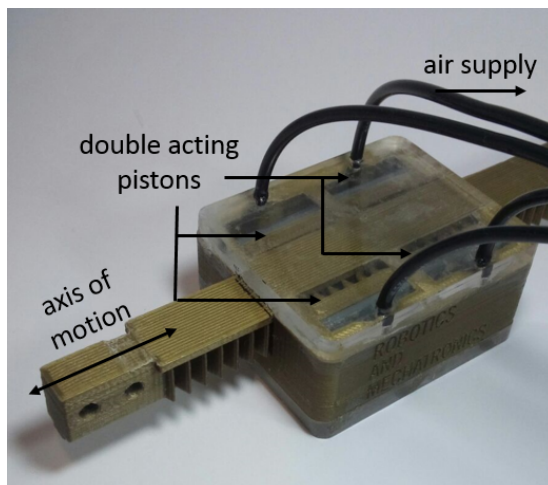


Figure 10: The linear actuator: a pneumatic stepper motor.

sensor was chosen. This Force Sensitive Resistor (FSR) was integrated into a Wheatstone bridge and a simple voltage divider according to the configurations as shown in Figure 11. The Wheatstone bridge uses two analog pins to measure the voltage across the bridge due to the ground being connected to the node halfway through the bridge. The voltage across the bridge (U_b) is calculated by subtracting the voltage measured on analog pin 2 from the voltage on analog pin 1 ($U_b = U_1 - U_2$). The FSR (located inside the MRI-room) is connected to the Wheatstone bridge or voltage divider with 3m long shielded twisted-pair cabling. The voltage across the bridge or the voltage divider was measured with an Arduino Uno, acting as an A-D converter. The Wheatstone bridge is configured to be (approximately) balanced when the FSR is experiencing zero force. When a force is applied, the bridge is not attempted to be rebalanced by a variable resistor; instead the force is deduced from a calibrated mapping between the voltage across the bridge and axial force exerted.

In order to place the sensor correctly, an attachment for the actuator has been designed which holds the sensor and the 'needle' in such a way it translates the needle's axial forces adequately. Figure 12 shows two views of the disassembled attachment in SOLIDWORKS. The SOLIDWORKS model of the attachment has been 3D-printed using a Stratasys Fortus 250mc, using the material ABS. The assembled attachment is shown in Figure 13. Nylon bolts and nuts (M4) are used to tighten everything together.

The sensor's needle pushes into a gelatin-like cuboid, the phantom. The phantom used for the complete setup was made of 70 % plastisol with dimensions $100 \times 100 \times 35 \text{ mm}$ ($L \times W \times H$). For the experiments inside the MRI-scanner, a different phantom was used to improve the reference image quality. This phantom will be discussed later.

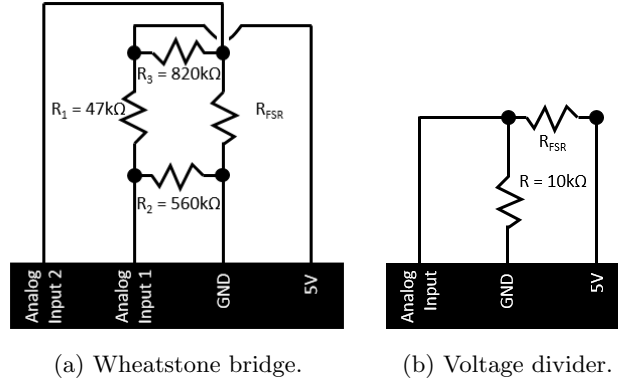


Figure 11: The connections between the FSR, Arduino Uno, and two measuring circuits. The black rectangle signifies pins on the Arduino Uno.

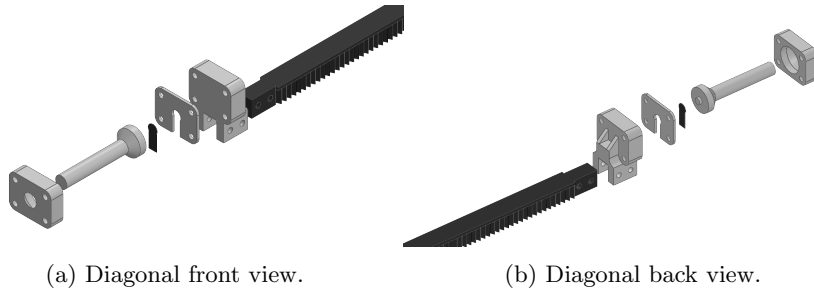


Figure 12: SOLIDWORKS assembly of the actuator's end-effector (dark grey), Interlink FSR400 force sensor (dark grey), and the disassembled sensor attachment (light grey).

3.1.3 Interface

The system has been interfaced with MATLAB and an Arduino Uno to allow control of the switching valves (which are in turn controlling the actuator) with the PHANTOM Omni. Due to the required IEEE-1394 FireWire port for the Omni, a desktop computer running on windows 7 was used. Sensable's PHANTOM Omni driver software version 5.1.7 was installed. 64-bit MATLAB 2015b together with an add-on for Arduino support enabled control of the valves through the Arduino Uno.

3.1.4 Remote Control

The slave's actuator should be controlled through the master in order to make the system teleoperated. As previously mentioned, the master has three joints capable of providing force feedback to the user. This allows the control of

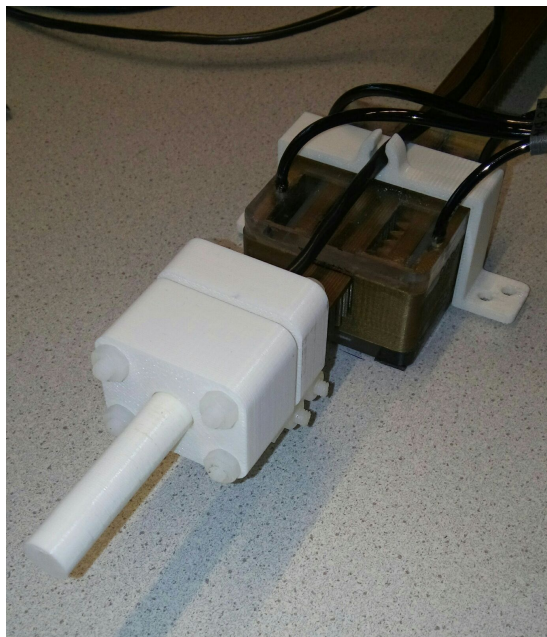


Figure 13: The sensor attachment, including FSR and wiring, attached to the actuator’s end-effector.

the slave through any of, or a combination of, these joints. The slave being a linear actuator asks for merely one or two joints being used. A combination of two joints would allow for an intuitive linear motion in Cartesian coordinates, whereas the use of one joint would increase the possible workspace. A single joint, J_1 as seen in Figure 9, was used in a position-control scheme, where an angular shift in J_1 of $0.04[rad]$ maps to one step ($1mm$) of the slave. For comparison, the full ranges of joint J_1 and the linear actuator are $2rad$ and $140mm$, respectively; although the actuator’s full range is deliberately not used as this would make control through the Omni too sensitive. Instead, the actuator is able to move up to $50mm$ with the parameters used.

3.1.5 Implementation Haptic Feedback

Kinesthetic feedback should be employed to make the user, remotely controlling the slave, more aware of the needle’s surroundings. The haptic feedback control works together with the control allowing teleoperation of the slave, and runs at $20Hz$. Haptic feedback could not be implemented in the ideal way where the master mimics the slave’s experienced forces at a 1:1 ratio. Instead, after some observation, it seemed viable to skip the calibration step, and map the obtained range of voltages linearly to the range of forces the master is capable of providing. The voltage range was chosen to be $[0.5V\ 3.0V]$. A threshold value

of $0.5V$ was chosen to ignore small voltage peaks that spontaneously occurred. A maximum feedback force of $2.5N$ was used, resulting in a force range of $[0 \ 2.5]$. Figure 14 shows the mapping function as previously described.

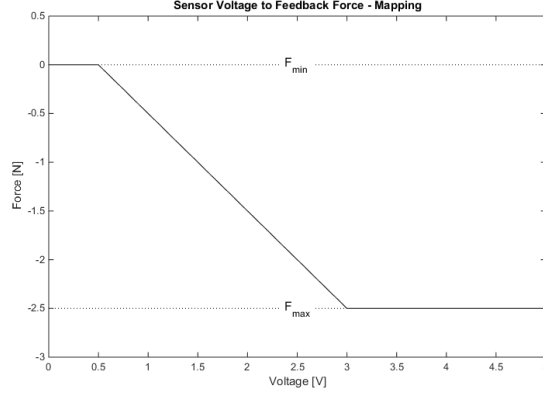


Figure 14: Mapping of the measured voltage at the actuator’s end-effector to the feedback force supplied by the PHANTOM Omni.

The slave is controlled by rotation in one of the joints of the Omni. Yet the Omni takes control signals specifying the forces in Cartesian coordinates. The feedback that is supplied should be opposite to the direction of motion, and orthogonal to the Omni’s arm in top view. Figure 15 shows this top view with the labeling of variables. The force $F_{feedback}$ that is to be provided as haptic feedback to the user has to be expressed in two Cartesian forces, F_x and F_y . The following equations are found:

$$\begin{aligned} F_x &= -\sin(q(t)) * F_{feedback} \\ F_y &= \cos(q(t)) * F_{feedback} \end{aligned}$$

Signs are configured to use a positive value for $F_{feedback}$, or the Euclidean norm. The vertical force in the 3D workspace of the Omni, F_z , will never have a value assigned.

3.2 Experimental setup

3.2.1 Validation Actuator Displacement and Velocity

To allow the validation of the actuator’s actual displacement and velocity, these values were measured and compared to their reference values in the control signals.

The actuator has been equipped with two markers, as seen in Figure 16a. One marker has been placed on the stationary ‘core’ of the actuator, whereas the other has been placed on the moving end-effector. Recording perpendicular

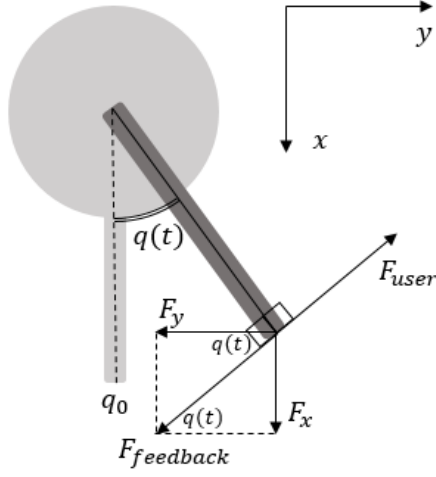


Figure 15: Schematic top view of the omni where the arm rotates around the z -axis (rotation in J_1). The top right shows the reference axes as used in the MATLAB drivers for the PHANTOM Omni.

to the axis of movement allows tracking of the moving marker's centroid. In MATLAB, the camera (Logitech C922 Pro Stream) takes snapshot at a frame-rate of 3 Hz. Using the snapshot, a mask is created - distinguishing the markers from the environment - and both marker's centroid positions are stored to be viewed later, all at real-time. Figure 16b shows the setup as seen from the camera and its image after MATLAB processing. In the bottom part, x_0 is the reference distance between the two marker's centroids and x represents the actuator's displacement (relative to starting position x_0).

In order to correlate the centroid's position in the image to the real-life position, the pixel-spacing has to be obtained. The pixel-spacing is a measure of scaling between a real distance (in mm) and the amount of pixels in the image it consists of. The pixel-spacing is found to be $0.3764mm/pixel$ and was obtained by manually moving the mobile marker over a known distance across the same axis (i.e. same distance from camera) it would move when attached to the actuator's end-effector. The actual displacement in $[mm]$ has been differentiated using a numerical two-point backward algorithm to obtain the actuator's velocity at each point.

To obtain an error-margin (in the x -direction), a stationary marker was recorded for 4 seconds. The minimum x -coordinate of the marker's centroid was subtracted from the maximum. An error-margin of $0.3291mm$ was found.

A reference signal was used which prompted the actuator to move for 4 seconds with velocities of 5, -5, 10, -10, 15, -15, 20 and -20 mm/s .

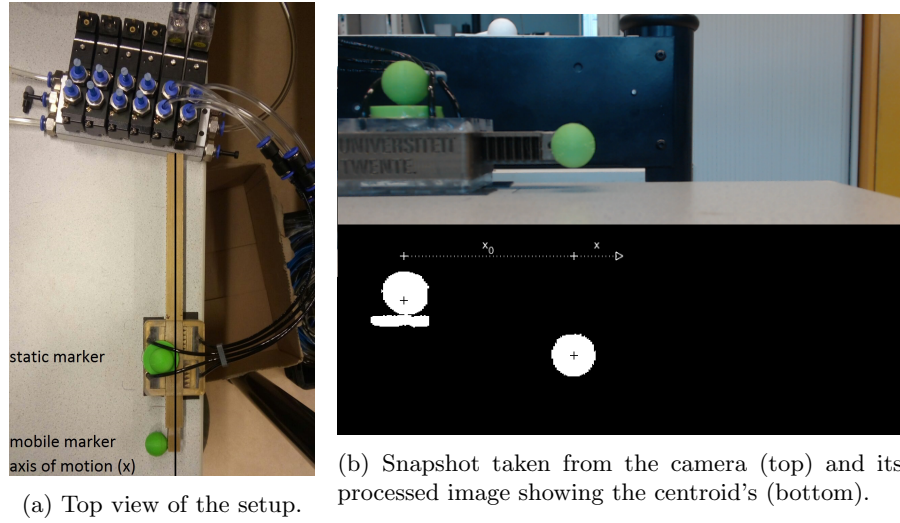


Figure 16: Experimental setup of actuator displacement and velocity validation.

3.2.2 Calibration Force Sensor

A calibration of the FSR has to be done to map the measured voltage across the Wheatstone bridge to the applied axial force on the needle. To obtain the mapping, a series of known masses are placed on the actuator's end-effector tip - or the needle - which has been positioned vertically upwards in order to convert the gravitational forces of the masses to axial forces exerted on the FSR. The axial forces can be deduced from the known masses. Figure 17 shows this setup.

The calibration of the FSR used 13 different masses ranging from 112.2g to 643.5g. A minimum of two and a maximum of six measurements were done with each mass.

3.2.3 Evaluation MRI-compatibility

To evaluate the MRI-compatibility of the slave system, the linear actuator with the sensor attached were placed inside an MRI-scanner (0.25T G-scan Brio, by Esaote). As a reference, a breast phantom made of soft polyvinyl chloride (PVC) plastic with small fish oil capsules inside was used. Three 3D images are taken using a 3D Hyce sequence, an imaging sequence relatively sensitive to metallic artifacts. Firstly, only the phantom is placed inside the scanner. Next the slave system is added in proximity of the breast phantom, with its end-effector being inside the RF-coil. The slave will not move yet in the second image. In the third image, the motor continuously moves back and forth over a distance of 20mm with a speed of 6mm/s, not touching the phantom in maximally extended position. See Figure 18 for the used setup. The images taken will be analyzed for their artifacts, using the first image as a baseline as no component of the

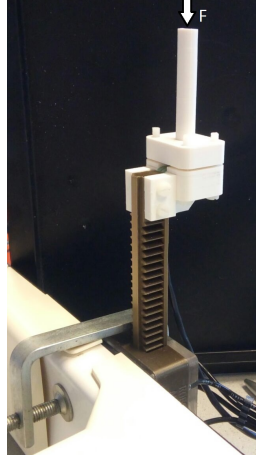


Figure 17: Vertical configuration of the actuator for FSR calibration.

system will be present yet.

3.2.4 Evaluation of Haptic Feedback

The implementation of haptic feedback in the completed system has been evaluated by comparing needle insertion with the teleoperated system to manual needle insertion. To allow direct comparison, manual insertion was performed with a similar quasi-needle; the smooth back-end of a drill bit with a diameter equal to the sensor's needle within a margin of half a millimeter. Both manual and teleoperated insertions were performed twice; once with and once without visual feedback. The presence of visual feedback in this scenario envelops the ability to view the slave system and phantom. This means a total of four sub-experiments are conducted per person, in random order. The four possible combinations of insertion and visual feedback are labeled A-D corresponding to the configuration as shown in Figure 19.

Four test subjects were asked to poke around in the phantom for approximately 15 seconds. In the case of teleoperated insertion, up to a minute was allowed, to let everyone familiarize with the control of the system a little more. Every subject was asked the following questions. The first subset of questions were asked after every single sub-experiment, the second subset was asked only once after the entire experiment had been finished.

1. *Post sub-experiment*

- (a) To what extent can you understand the needle's surroundings? (1-10)
- (b) Does operating the system require a lot of attention and focus? (1-10)
- (c) How would you rate the transparency of the system? (1-10)

2. *Post experiment*

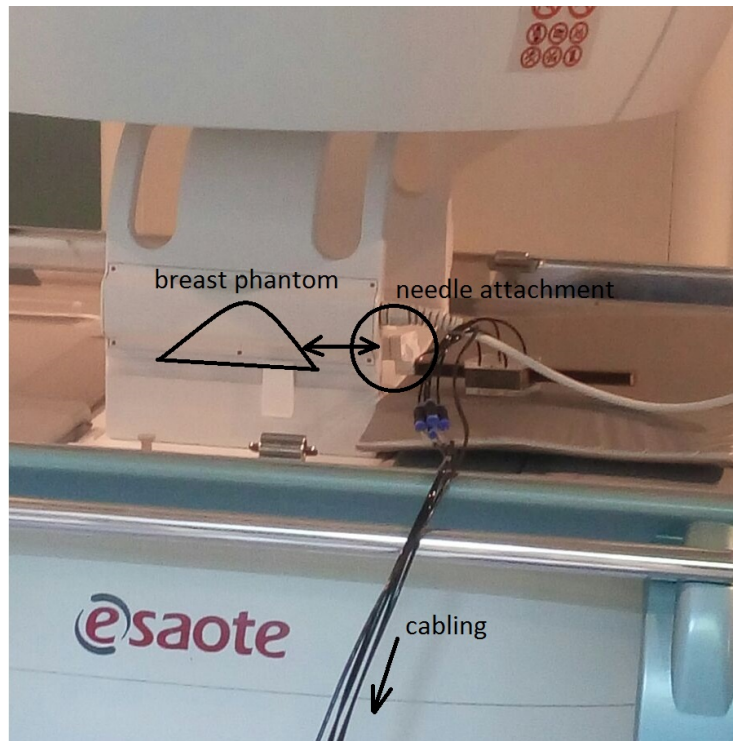


Figure 18: Side-view of the slave system inside the MR-bore. As the high magnetic field did not permit risk-free entering with a camera, the breast phantom's position inside the RF-coil is sketched. The range of motion of the actuator's end-effector (circled) has been sketched with a two-way arrow.

- (a) Is the system intuitive to use? (1-10)
- (b) Does visual feedback on the needle tip's position improve the user's experience? (Y/N)
- (c) Would you prefer using the teleoperated system rather than manual insertion when working in a confined space? (Y/N)
- (d) Do you have any additional feedback?

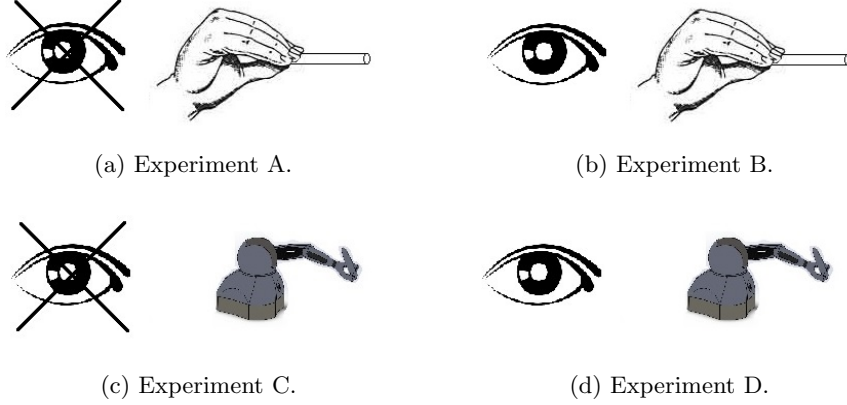


Figure 19: Setup and naming of the sub-experiments.

4 Results

In this section, the results of the experiments as described in section 3.2 are presented.

4.1 Validation Actuator Displacement and Velocity

Figure 20 shows the results found from this experiment. The validation experiment has been done twice, yielding minimal differences between the two iterations. The displacement showed no extraordinary fluctuations and the shaky velocities showed a similar distribution in both cases.

As can be seen from the figure, the displacement does not always reach the expected value. Differences between expected displacement and actual displacement as measured with the camera are shown in Table 2, where the mean of the two measurements has been used.

As the reference velocity increases, the actuator finds it harder and harder to reach the expected displacement when extending (positive velocities). Seeing how on the way back (negative velocities) the mobile marker always returns to its original position ($x = 0$), the cause seems to be a faulty velocity. Although the combination of small displacement errors from the camera and the numerical approach for differentiation causes the velocities to be distributed rather than constant, the results are clear enough. The distributed individual measured velocities are distributed more or less evenly with the reference velocity as its center for small speeds ($5mm/s$). When you compare this centerline for the maximum speed of $20mm/s$, the reference velocities clearly cannot be realized. For this reason, a maximum velocity of $15mm/s$ was chosen for the actuator in the complete system. Velocities measured near the time where the actuator was told to start or stop moving are anywhere between the new and old reference velocity. This is nothing to worry about since these values are most likely due

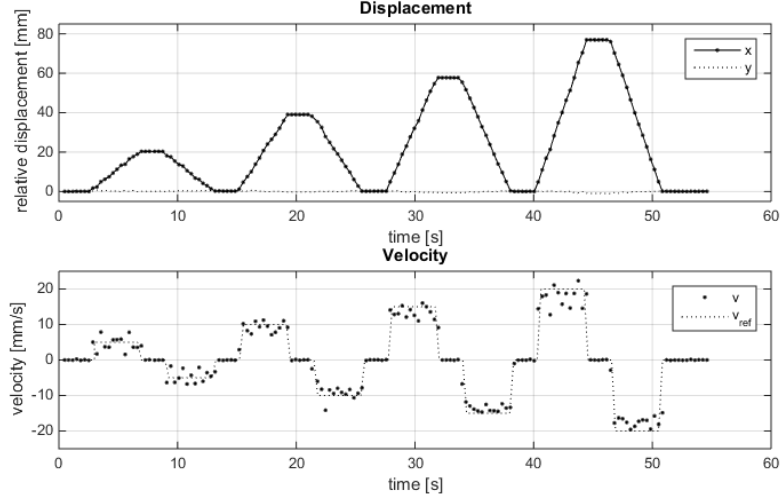


Figure 20: The measured actuator displacement and velocity in the time domain, with the reference velocity signal plotted in the background.

to the relatively slow frame-rate of the camera.

4.2 Calibration Force Sensor

After a total of 48 calibration measurements, the results as shown in Figure 21 were obtained. The known masses whose gravitational forces were exerted on the FSR plotted against the resulting voltage measured across the Wheatstone bridge. It can be observed that near the asymptote at a voltage of 4.5V the maximum voltage is reached, and an increasing force does not significantly change the voltage anymore.

As is seen in Figure 21, the voltage caused by a mass can differ greatly between separate measurements of the same mass. Observations throughout multiple attempts at measuring calibration data showed the surface of the FSR that should be actuated by the back-end of the needle could not be actuated in a constant manner. The pressure was not uniformly distributed across the surface, causing small positional changes of the needle with respect to the sensor's casing to result in unreliable calibration measurements. Figure 22 shows the effect of the little play in the sensor's needle that was allowed. Two masses were put on the needle in turn. The masses used are labeled inside the figure. The heavy disturbances just as a mass is put on the needle are ignored as they are due to some time spent balancing the mass.

Table 2: Differences between expected and measured displacements ($\Delta x = x_{measured} - x_{expected}$). Positive and negative values in the last column signify extra and skipped steps, respectively.

Velocity [mm/s]	Deviation in displacement (Δx) [mm]	Extra/Skipped steps
5	+0.1535	0
-5	+0.1323	0
10	-1.1464	-1
-10	-1.0727	-1
15	-2.3203	-2
-15	-2.3039	-2
20	-3.1227	-3
-20	-3.1075	-3

4.3 Evaluation MRI-compatibility

The MRI-images of the phantom are shown in Figure 23. The images shown are 2D coronal slices, at the same height from the floor as the slave’s sensor. The same artifacts are present in both the 3D and 2D images, hence only results referring to the 2D images are discussed.

It can be seen that the addition of the static slave system did not influence the image quality. The internal oil capsules can still be seen well, with a seemingly equal contrast and resolution. The rightmost image does show artifacts. Immediately noticeable is the decreased contrast. The image shown is already manually brightened, whereas the other two are not. This is required to clearly see the image, as the MRI-scanner automatically lowered the RF-power for this last image. Another artifact only present when the actuator is moving are horizontal lines in the image, like the one indicated by the white arrow in the figure. Each line represents a specific frequency that leaked into the MR-room, resulting in a horizontal line after image reconstruction. While scrolling through the (coronal) slices in the 3D image, more of these lines are seen.

4.4 Evaluation of Haptic Feedback

Figure 24 shows the course of some of the most important signals during needle insertion ($t \approx 3s$) and retraction ($t \approx 17s$). The variable x_{q1} stands for the master configuration (q_1 [rad]) expressed in the corresponding actuator displacement [mm]. Circles in the top plot mark a step backward and immediately forward again (or vice versa), while the trend of the master signal (represented by x_{q1}) follows a more constant, slowly stepping signal. These sudden two steps back and forth are felt by the user as if a sudden disturbance in the phantom pushes against the needle in a tiny burst. In the bottom plot, arrows indicate irregularities in the measured voltages, usually occurring as the actuator performs a step while the needle tip is inside the phantom, and thus experiencing

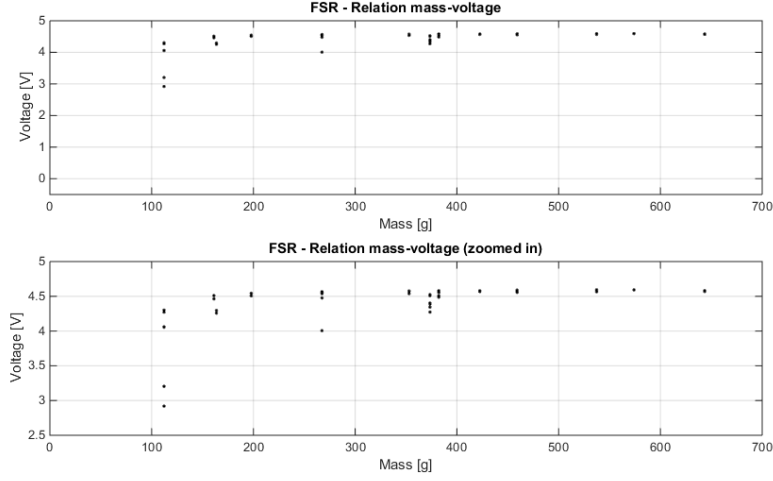


Figure 21: Results of the FSR calibration. Top and bottom plots concern the same data, only the bottom plot is zoomed in on the vertical axis.

forces. The plots show no notable time-delays in the transmission of the needle's forces to the user.

The results of the first part of the questionnaire are shown in Figure 25. Each subplot corresponds to a different sub-experiment, where the black lines represent the mean values for each subquestion. These mean scores are also shown in Table 3.

Some observations are made, using the two manual insertions (experiments A and B) as a reference value. The first question, concerning the ease with which the haptic feedback is interpreted, showed basic haptic feedback is available to the user, who is able to mark the moment contact between needle and phantom is either made or lost. More refined haptic feedback is lacking, as during insertion and retraction (while inside the phantom) the score given, as well as the transparency-score (question 1c) decreased in a similar fashion.

Table 3: Mean scores given to the first set of questions asked. Rows distinguish between questions, columns distinguish sub-experiments.

	A	B	C	D
1a	6.75	9.50	4.00	7.00
1b	3.25	3.25	5.25	2.75
1c	8.75	9.25	4.00	6.75

The results of the questions asked after all the experiments are shown in Figure 26. The mean score for question 2a is 8.25. Supplemented explanations stated a translational movement in the master would further increase the intuitiveness of the system, compared to the used method where rotational motion

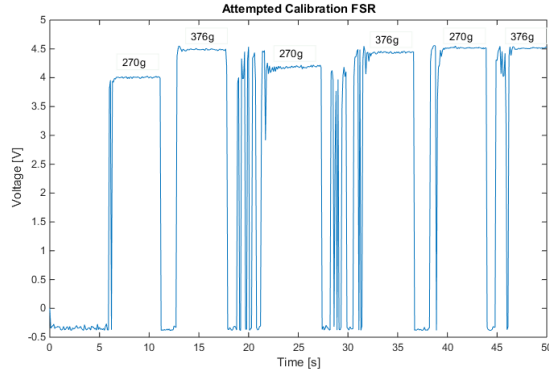


Figure 22: Voltage measurements across the Wheatstone bridge as two masses are alternately put on the sensor.

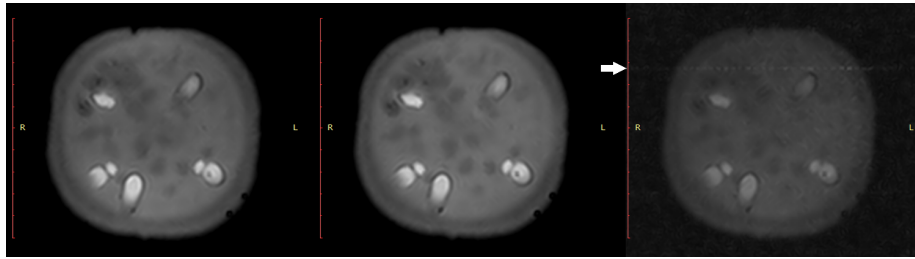


Figure 23: 2D Coronal cross-sections of the MRI-images taken of (from left to right): the breast phantom, the phantom and the static slave, the phantom and the moving slave. The white arrow indicates one of the artifacts in the form of a horizontal line.

controlled the slave's linear motion. The results of the first set of subquestions support the unanimous decision that visual feedback is required for functional operation of the teleoperated system.

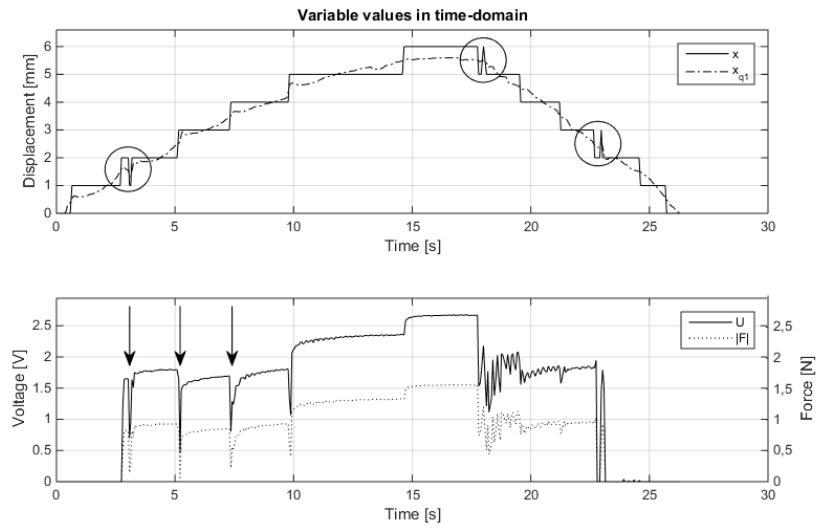


Figure 24: MATLAB plot of variables during teleoperated needle insertion and retraction. The top figure shows the displacement of the actuator, expressed in reference signal (x_{q1}) and actual displacement (x). The bottom figure shows the voltage measured by the sensor and the corresponding feedback force the Omni exerts. Circles and arrows indicate interesting results that will be discussed.

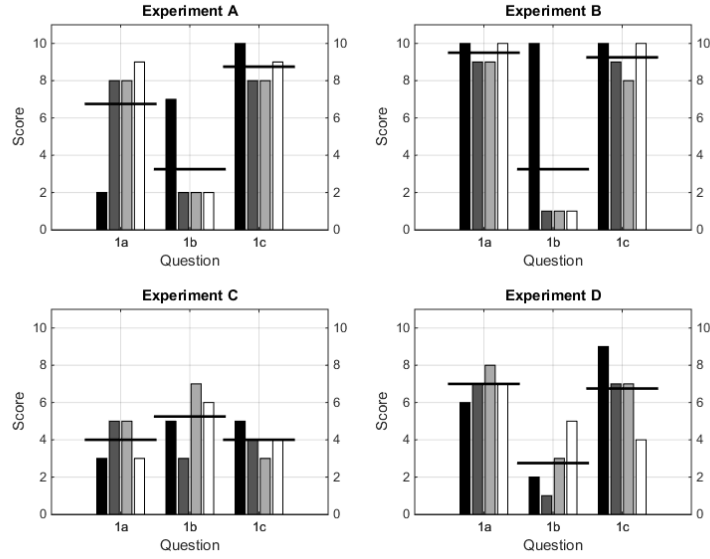


Figure 25: Results of the first set of questions, asked after each sub-experiment. The questions were: 1a) To what extent can you understand the needle's surroundings? 1b) Does operating the system require a lot of attention and focus? 1c) How would you rate the transparency of the system?

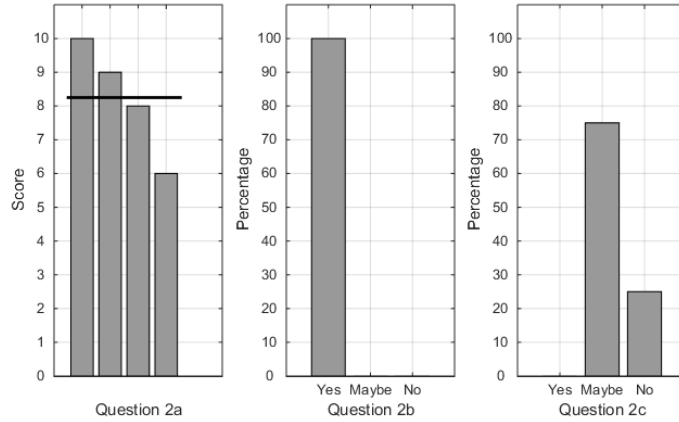


Figure 26: Results of the second set of questions. The questions were: 2a) Is the system intuitive to use? 2b) Does visual feedback on the needle tip's position improve the user's experience? 2c) Would you prefer using the teleoperated system rather than manual insertion when working in a confined space?

5 Discussion

This section interprets the previously presented results. Possible logic behind unexpected results, limitations of the complete system, and recommendations for future research are presented.

The validation of the motor displacement and velocity showed that with increasing velocity, the amount of steps being skipped also increases in a linear fashion. As a perfect linear increase can be found within the allowed error bounds of $\pm 0.3291mm$, this hints at an error in the pixel-spacing causing all measured displacements, and consequently all velocities, to differ a factor from their real values.

The calibration results shown used a Wheatstone bridge, which excels at detecting minor changes. This high sensitivity proved to be too much for this setup, hence a single voltage divider was used for the system evaluation. The calibration of the FSR using a voltage divider was not attempted, as the fault did not seem to lie in the measuring circuit but the sensor attachment. A trade-off is present, between reducing friction between the needle and casing on the one hand and tightly encasing the needle on the other. Meaning a balance has to be found between being able to exert the exact forces of the needle to the user at a 1:1 ratio, and achieving a constant FSR actuation, decreasing fluctuations in the force to voltage transmission.

The reasoning behind lowering the RF-power for the third MRI image is unknown, as the manufacturer discloses the algorithm from users. Therefore this decision internally made by the scanner software cannot be interpreted. The horizontal lines on the other hand can be explained as they are due to electric interference. The MR-room is closed off and acts as a Faraday cage with a sliding door. The cabling supplying the pressurized air and connecting the sensor had to be led through the door, leaving the door open for approximately 2 cm. This small cavity in the Faraday cage allows electrical interference to enter the MR-room and distort the RF-signals for specific frequencies that have entered. As the static motor does not require any electrically active components (the laptop controlling the Arduino and generator supplying pressurized air were still shut off), the noise was only introduced as the motor was turned on. This suggests moving these electrically active components farther away from the MR-room may be a solution. Longer cabling would be a necessity, however, decreasing the slave's functionality by limiting its maximum velocity.

After evaluation of the results in section 4.4 it showed the questionnaire requires some updating. Question 1b received answers ranging from very high to very low ratings, especially in experiment B. It was not clarified to the test subjects whether they should or should not lay their free hand on the phantom to obtain more insight into the location of the phantom relative to the needle they were holding. This shows clearly in the distribution of scores for experiments A and B.

The quality of haptic feedback during needle insertion and retraction suffered from inadequate force sensing. The arrows in Figure 24 show that with

every step of the motor, the measured voltage is peaking downwards, resulting in a sudden, drastic change in force the user is pushing against. The sudden steps made, marked with the circles, may be caused by the phenomenon just described, as a user is not expecting this drastically fluctuating force. Filtering of the measured voltage could eliminate these peaks from the signal and consequently improve the user's experience. A mechanical remedy may be to make use of a stepper motor with smaller step size, or not even a stepper motor at all but one capable of providing more smooth motion as both may decrease the artifacts in the sensor's signal.

Future research may be done into increasing the system's transparency. One approach could be to make use of a fiber-optic force sensor, consequently capable of providing full MRI-compatibility. Another approach is to use a motor allowing smoother motion.

6 Conclusion

After construction of the master-slave system, several experiments have been conducted. The stepper motor’s velocity and displacement have been validated. The force sensitive resistor could not be actuated constantly, causing the measured voltage (relating to a tissue-force) not to be able to be calibrated properly. The slave, including this electrically-active sensor, was imaged using a low-field MRI-scanner, yielding reasonable image-quality but no full MRI-compatibility. Lastly, four test subjects were asked to perform a number of tests with the tele-operated system or a manual form of insertion. Several questions these subjects answered proved successful implementation of the basic functionalities of the system. Space for future research lies at improving force sensing at the needle.

In this section conclusions are drawn from the results and their interpretations. The conclusions refer back to the objective, presented in the introduction.

The validation of the stepper motor’s velocity and displacement showed the system is capable of remotely inserting a needle into a phantom. The accuracy of the realization of desired motion does not seem to be guaranteed for higher velocities, although this may have been due to a faulty error margin.

Even though the moving slave system in the MRI did not distort the images drastically, it can be concluded the image quality is not optimal compared to the image of the plain phantom. Concluding, the slave-side of the system is not marked as MRI-compatible for the imaging sequence used, but merely MRI-safe. This means the system will not introduce safety hazards for a patient nor the equipment, yet it will have a negative impact on the imaging quality.

The force sensor, essential for the master’s implementation of haptic feedback, did not live up to expectations, consequently limiting the system’s quality of haptic feedback. Basic feedback is successfully implemented, but improving the system’s transparency is required to guarantee a patient’s safety. The required real-time aspect of the feedback is successfully implemented.

References

- [1] Force dimension haptic devices. <http://www.forcedimension.com/products>. Accessed: 25-04-2017.
- [2] Geomagic haptic devices. <http://www.geomagic.com/en/products-landing-pages/haptic>. Accessed: 25-04-2017.
- [3] K. Bark, W. McMahan, A. Remington, J. Gewirtz, A. Wedmid, D. I. Lee, and K. J. Kuchenbecker. In vivo validation of a system for haptic feedback of tool vibrations in robotic surgery. *Surgical endoscopy*, 27(2):656–664, 2013. doi: 10.1007/s00464-012-2452-8.
- [4] E. M. Boctor, M. A. Choti, E. C. Burdette, and R. J. Webster Iii. Three-dimensional ultrasound-guided robotic needle placement: an experimental evaluation. *The International Journal of Medical Robotics and Computer Assisted Surgery*, 4(2):180–191, 2008. doi: 10.1002/rcs.184.
- [5] S. Elayaperumal, J. H. Bae, B. L. Daniel, and M. R. Cutkosky. Detection of membrane puncture with haptic feedback using a tip-force sensing needle. In *Intelligent Robots and Systems (IROS 2014), 2014 IEEE/RSJ International Conference on*, pages 3975–3981. IEEE, 2014. doi: 10.1109/IROS.2014.6943121.
- [6] R. A. Faust. *Robotics in surgery: history, current and future applications*. Nova Publishers, 2007. ISBN 978-1-60021-386-1.
- [7] G. Fichtinger, A. Deguet, K. Masamune, E. Balogh, G. S. Fischer, H. Mathieu, R. H. Taylor, S. J. Zinreich, and L. M. Fayad. Image overlay guidance for needle insertion in ct scanner. *IEEE transactions on biomedical engineering*, 52(8):1415–1424, 2005. doi: 10.1109/TBME.2005.851493.
- [8] V. Groenhuis and S. Stramigioli. Laser-cutting pneumatics. *IEEE/ASME transactions on mechatronics*, 21(3):1604–1611, 2016. doi: 10.1109/TMECH.2015.2508100.
- [9] W. Harris. How haptic technology works. <http://electronics.howstuffworks.com/everyday-tech/haptic-technology.htm>. Accessed: 25-04-2017.
- [10] V. Hayward and K. E. MacLean. Do it yourself haptics: part i. *IEEE Robotics & Automation Magazine*, 14(4), 2007. doi: 10.1109/M-RA.2007.907921.
- [11] C. S. Inc. Featured hardware. <http://www.cyberglovesystems.com/>. Accessed: 25-04-2017.
- [12] I. Intuitive Surgical. The da vinci® surgical system. <http://www.davincisurgery.com/da-vinci-surgery/da-vinci-surgical-system/>. Accessed: 01-05-2017.

- [13] S. D. Laycock and A. Day. Recent developments and applications of haptic devices. In *Computer Graphics Forum*, volume 22, pages 117–132. Wiley Online Library, 2003. doi: 10.1111/1467-8659.00654.
- [14] M. Mahvash, J. Gwilliam, R. Agarwal, B. Vagvolgyi, L.-M. Su, D. D. Yuh, and A. M. Okamura. Force-feedback surgical teleoperator: Controller design and palpation experiments. In *Haptic interfaces for virtual environment and teleoperator systems, 2008. haptics 2008. symposium on*, pages 465–471. IEEE, 2008. doi: 10.1109/HAPTICS.2008.4479994.
- [15] M. Moche, S. Heinig, N. Garnov, J. Fuchs, T.-O. Petersen, D. Seider, P. Brandmaier, T. Kahn, and H. Busse. Navigated mri-guided liver biopsies in a closed-bore scanner: experience in 52 patients. *European radiology*, 26(8):2462–2470, 2016. doi: 10.1007/s00330-015-4097-1.
- [16] O. Mohareri, C. Schneider, and S. Salcudean. Bimanual telerobotic surgery with asymmetric force feedback: A davinci® surgical system implementation. In *Intelligent Robots and Systems (IROS 2014), 2014 IEEE/RSJ International Conference on*, pages 4272–4277. IEEE, 2014. doi: 10.1109/IROS.2014.6943165.
- [17] A. M. Okamura. Haptic feedback in robot-assisted minimally invasive surgery. *Current opinion in urology*, 19(1):102, 2009. doi: 10.1097/MOU.0b013e32831a478c.
- [18] W. J. Peine and R. D. Howe. Do humans sense finger deformation or distributed pressure to detect lumps in soft tissue. In *Proc. ASME Dynamic Systems and Control Division, DSC*, volume 64, pages 273–278. Citeseer, 1998.
- [19] P. Puangmali, H. Liu, L. D. Seneviratne, P. Dasgupta, and K. Althoefer. Miniature 3-axis distal force sensor for minimally invasive surgical palpation. *IEEE/ASME Transactions on Mechatronics*, 17(4):646–656, Aug 2012. ISSN 1083-4435. doi: 10.1109/TMECH.2011.2116033.
- [20] J. M. Romano, R. J. Webster, and A. M. Okamura. Teleoperation of steerable needles. In *Robotics and Automation, 2007 IEEE International Conference on*, pages 934–939. IEEE, 2007. doi: 10.1109/ROBOT.2007.363105.
- [21] R. Seifabadi, S.-E. Song, A. Krieger, N. B. Cho, J. Tokuda, G. Fichtinger, and I. Iordachita. Robotic system for mri-guided prostate biopsy: feasibility of teleoperated needle insertion and ex vivo phantom study. *International journal of computer assisted radiology and surgery*, 7(2):181–190, 2012. doi: 10.1007/s11548-011-0598-9.
- [22] W. Shang, H. Su, G. Li, and G. S. Fischer. Teleoperation system with hybrid pneumatic-piezoelectric actuation for mri-guided needle insertion with haptic feedback. In *Intelligent Robots and Systems (IROS), 2013 IEEE/RSJ International Conference on*, pages 4092–4098. IEEE, 2013. doi: 10.1109/IROS.2013.6696942.

- [23] H. Su, I. I. Iordachita, J. Tokuda, N. Hata, X. Liu, R. Seifabadi, S. Xu, B. Wood, and G. S. Fischer. Fiber-optic force sensors for mri-guided interventions and rehabilitation: A review. *IEEE Sensors Journal*, 17(7): 1952–1963, 2017. doi: 10.1109/JSEN.2017.2654489.
- [24] K. Tadano and K. Kawashima. Development of 4-dofs forceps with force sensing using pneumatic servo system. In *Robotics and Automation, 2006. ICRA 2006. Proceedings 2006 IEEE International Conference on*, pages 2250–2255. IEEE, 2006. doi: 10.1109/ROBOT.2006.1642038.
- [25] L. Vorvick. Palpation. <https://medlineplus.gov/ency/article/002284.htm>. Accessed: 30-06-2017.
- [26] T. Watanabe, T. Iwai, Y. Fujihira, L. Wakako, H. Kagawa, and T. Yoneyama. Force sensor attachable to thin fiberscopes/endoscopes utilizing high elasticity fabric. *Sensors*, 14(3):5207–5220, 2014. doi: 10.3390/s140305207.
- [27] P. Xia. Haptics for product design and manufacturing simulation. *IEEE transactions on haptics*, 9(3):358–375, 2016. doi: 10.1109/TOH.2016.2554551.
- [28] K. Yokoyama, H. Nakagawa, D. C. Shah, H. Lambert, G. Leo, N. Aeby, A. Ikeda, J. V. Pitha, T. Sharma, R. Lazzara, et al. Novel contact force sensor incorporated in irrigated radiofrequency ablation catheter predicts lesion size and incidence of steam pop and thrombusclinical perspective. *Circulation: Arrhythmia and Electrophysiology*, 1(5):354–362, 2008. doi: 10.1161/CIRCEP.108.803650.

A Overview Fiber-Optic Force Sensors

Author	Principle (# of fibers)	MRI compatible	DOF	Dimension: OD/L (mm)	Range	Resolution	Application
Hirose et al., 1995 [28]	Intensity (6)	No	6	76/40	980N	2.94N	General
Takahashi et al., 2003 [29]	Intensity (6)	Yes	6	NA	20N ^a	0.3N	Neuroscience
Peirs et al., 2004 [36]	Intensity (3)	No	3	5/11	2.5N (AX), 1.7N (RA)	0.01N	Laparoscopy
Chapuis et al., 2004 [42]	Intensity (2)	Yes	1	NA	5Nm	0.07Nm	Neuroscience
Tada et al., 2005 [43]	Intensity (4)	Yes	3	25/18	15N (AX), 8N (RA)	0.24N	General
Tokuno et al., 2008 [32]	Intensity (2)	Yes	1	25/11	3N	0.048N	General
Yokoyama et al., 2008 [2]	FBG (3)	Yes	3	3.5/NA	0.5N	10mN	Cardiac ablation
Su et al., 2009 [38]	Intensity (9)	Yes	3	25/35	10N	0.3N	Breast cancer
Iordachita et al., 2009 [44]	FBG (3)	Possible	2	0.5/NA	6.5mN	0.25mN	Ophthalmology
Yip et al., 2010 [45]	Intensity (2)	Possible	1	5.5/12	4N	0.13N	Cardiology
Su et al., 2011 [46]	FPI (1)	Yes	1	12×5×4 ^b	10N	1mN	Prostate cancer
Tan et al., 2011 [35]	Intensity (6)	Yes	3	49.5×48.3×50.8	7N	0.7N	Breast cancer
Kesner et al., 2011 [31]	Intensity (6)	Possible	1	6/NA	10N	0.2N	Cardiology
Polygerinos et al., 2011 [16]	Intensity (4)	Yes	1	3/18	0.85N	0.01N	Cardiology
Puangmali et al., 2012 [34]	Intensity (8)	Yes	3	10	3N (AX), 1.5N (RA)	0.02N	General
Liu et al., 2012 [47]	FPI (3)	Possible	3	0.5/NA	25mN	0.25mN	General
Moerman et al., 2012 [48]	FBG (3)	Yes	1	45/NA	15N	0.043N	General
Polygerinos et al., 2013 [37]	Intensity (4)	Yes	3	4/24.5	0.85N (AX), 0.45N (RA)	0.01N	Cardiology
Monfaredi et al., 2013 [49]	FBG (4)	Yes	2	15/20	20N (AX), 200Nmm (AX)	0.1N/1Nmm	Cardiology
Nakagawa et al., 2013 [50]	Spring deflection	No	3	3.5/NA	0.4N	10mN	Cardiac ablation
Su et al., 2013 [17]	FPI (1)	Yes	1	50×25×3.5 ^b	20N	1mN	Prostate cancer
Elayaperumal et al., 2014 [51]	FBG (3)	Yes	3	1.02/NA	±0.5N	0.043N	Surgical haptics
Turkseven et al., 2015 [52]	Intensity (2)	Yes	1	NA	7N	NA	Rehabilitation
Butzer et al., 2015 [33]	Intensity (2)	Yes	1	10×29.7×60	3N	0.01N	Neuroscience
Qiu et al., 2016 [53]	FPI (1)	Possible	1	1.88/NA	1.2N	0.25mN	Tissue mechanics
Noh et al., 2016 [40]	Intensity (2)	Yes	3	3.5/13	1N (AX), 0.5N (RA)	0.25mN	Cardiology
Xu et al., 2016 [54]	FBG (3)	Yes	2	2/NA	NA/(AX), 1.57N (RA)	0.01N	Continuum robot

Figure 27: Copy of Su et al.'s table, source [23] and references mentioned there.