2008-03-13

Force Measurement and Evaluation for Surgical Cutting Applications: Development of an Effective Characterisation Testbed

Dean Callaghan
Dublin Institute of Technology, dean.callaghan@dit.ie

Force Measurement & Evaluation for Surgical Cutting Applications: Development of an Effective Characterisation Testbed

Dean Callaghan

Faculty of Engineering, Dublin Institute of Technology

ABSTRACT

Sensorized instruments that cater for accurate measurement of the interaction forces (between biological tissue and instrument end-effector) during surgical procedures offer surgeons a greater sense of immersion during minimally invasive robotic surgery (MIRS). There is much ongoing research into force measurement/evaluation involving surgical graspers. However, comparatively little corresponding effort has been expended in the measurement and subsequent evaluation of forces between scissor blades and tissue. This paper presents the design and development of a force/strain measurement test apparatus, which will ultimately serve as an effective sensor characterisation and evaluation platform. Data acquired from the testing platform can be used to differentiate between tissue samples with differing mechanical properties in a reliable, repeatable manner. PVA cryogel samples which have been exposed to differing freeze-thaw cycles, giving properties similar to those of biological tissue are used. These samples, with a range of stiffness values, allow testbed performance to be evaluated over a wide range of tissue types. Further experimental data/analysis is presented which quantifies the levels of strain generated on the scissor blades during a cutting procedure. The resulting force/strain data correlates well with typical scissor cutting trends. This data is being used to establish a comprehensive set of operational requirements for force sensing transducers which could be incorporated into, or placed on to, a scissor blade end-effector. Future applications of the test equipment will include the assessment of new direct force sensing technologies for telerobotic end-effectors in minimally invasive robotic surgery.

1. INTRODUCTION

Minimally invasive surgery (MIS) is an operating technique during which long slender surgical instruments are inserted into the patient's body through small incisions in the skin. This allows the surgeon to manipulate and treat organs, and tissue within the body while observing the images on a 2-D monitor. The main advantages of this technique include smaller incisions, shorter hospital stays, lower risk of infection, and the convalescence period is reduced [1]. Robot-assisted surgery has revolutionised the way in which surgeons carry out minimally invasive surgical procedures. This robot assistance aids surgeons in overcoming the drawbacks associated with traditional MIS procedures such as hampered dexterity, reduced accuracy, and loss of 3-D visualisation [1]. The layout of a typical master-slave MIRS system is illustrated in Fig. 1 highlighting the three primary communication modalities

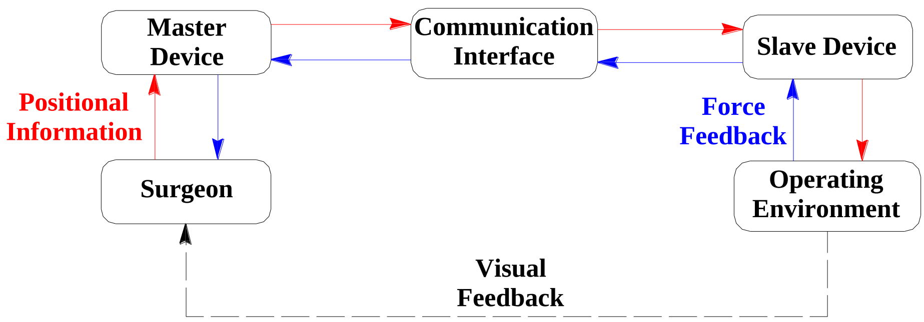


Figure 1: Typical master-slave system incorporating force feedback

(positional, force and visual information feedback) between the surgeon and the remote operating environment. Robotic assistance, as well as overcoming the disadvantages of MIS techniques, offer new advancements in areas such as; provision of additional degrees-of-freedom, tremor filtering and scaling of motions, particularly in the field of microsurgery [2]. Despite all the advantages, progress in the field of robotic assisted surgery is limited by an unresolved problem; the lack of haptic (force and tactile) feedback to the user [3]. One of the most widely used commercially available MIRS systems is the daVinci™ from Intuitive-Surgical® Inc. [4]. This daVinci™ system has been evaluated primarily in the field of minimally invasive heart surgery, but further applications will be established in future. This system offers control over tool position/displacement only; there is no force measurement at the slave side as well as no haptic feedback at the master console [4]. Lack of force feedback leads to difficulties for the surgeon in interpreting organ deformation as a measure of the forces at the slave side [5]. This is quite taxing on the surgeon, which can ultimately lead to reduced levels of concentration and increased fatigue. Complications such as accidental puncturing of blood vessels or tissue damage can also be attributed to lack of haptic feedback [7].

There is little doubt that the inclusion of force feedback in a MIRS system leads to improved performance over a system without haptic feedback. Previous work by Wagner [6] has looked at a blunt dissection task where it was found that force feedback reduces the number of errors that lead to tissue damage by a factor of three. A series of experiments on suture tying were carried out with and without, haptic feedback by Okamura [3]. These experiments conclude that haptic feedback is advantageous, and therefore desirable, in robotic surgical systems.

To date little research has been carried out to investigate the interaction between a pair of scissor blades and the tissue being cut with a view to developing a real-time force feedback solution. Work has been carried out by Greenish [8] to measure the forces generated while cutting a range of anatomical tissue. It was concluded from this work that exact quantitative measures for the forces required to cut tissues remained indeterminate. This data was subsequently used by Mahvash [9] and Chial [10] for virtual cutting simulations.

The work presented here focuses on the measurement of contact forces between a surgical instrument tip, in this case a pair of scissor blades, and the tissue with which it makes contact. In order to facilitate this, a force measurement evaluation testbed has been developed that will be used in the determination of the actual forces being generated at the instrument/tissue interface.

The remainder of the paper is organised as follows. An overview of the current methods used in measuring contact forces as well as a description of the design and development of the test apparatus for force measurement employed in this work are given in Section 2. Following this is a description of the synthetic tissue samples that were used in the cutting experiments. Detail on the experimental procedures as well as results from these experiments are presented in Section 3. The paper finishes with a discussion of these results followed by conclusions drawn from the work.

2. MATERIALS AND METHODS

2.1 Direct and Indirect Measurement

The decision as to where to locate the instrument/tissue interaction force measurement sensors in order to achieve maximum transparency is an issue for current researchers in this field. The options are to measure the forces directly by placing the force sensor as close as possible to the jaws of the end effectors (direct measurement) or placing the sensor at a location away from the instrument/tissue interaction i.e. external to the patient (indirect measurement). Research has been carried out, using both direct and indirect measurement methods, using laparoscopic instruments with a grasper as the end-effector [11-14].

Small, sterilisable force sensors which could be inserted into the patient are still a requirement for effective direct force sensing [11]. The most suitable solution, but technically the most challenging, is to integrate within the instrument, a miniaturised force sensor in close proximity to the instrument-tissue interface [14]. Experiments were carried out by Tholey [12] that validated the requirement for direct force sensing. These tests highlighted the extent to which the indirect technique overestimated the tissue grasping force by an order of magnitude when compared to the direct sensing method.

2.2 Scissor/Tissue Interaction

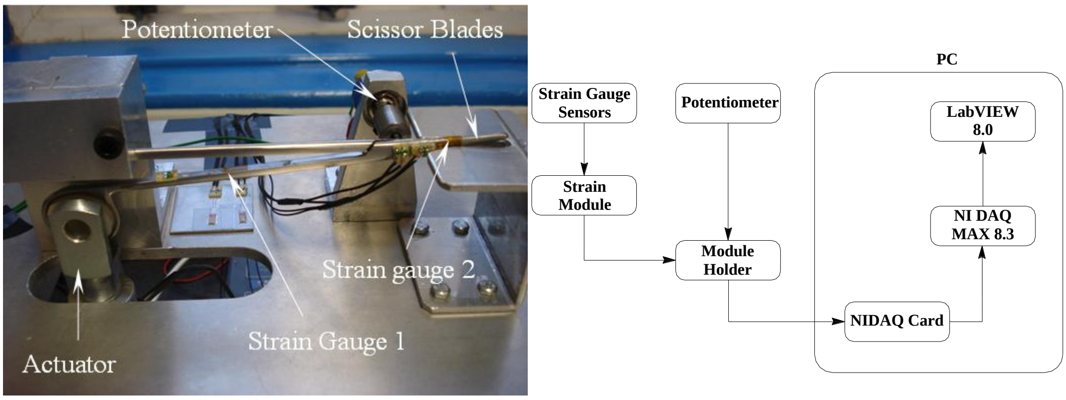
Cutting and grasping of tissues are two fundamental tasks that are commonplace in any surgical procedure [15]. Scissor dissection is one of the most frequently performed operations carried out using Laparoscopic surgery. The scissor blades can be either used to cut tissue by aligning the tissue to be cut between them or as a blunt dissector whereby the closed scissor tips are used to create incisions in the tissue. A further blunt dissector method involves the surgeon using the scissor tips in tissue separation techniques in which the closed tips are inserted into an opening in a piece of tissue, then opened to spread the tissue apart. Experimental work carried out by Greenish [8] showed that experienced surgeons typically will not use a scissors past 25° opening, except when spreading tissue apart, as it results in less control during cutting. These tasks can be carried out with increased effectiveness as well as enhanced safety if the surgeon is able to feel the forces experienced by the blades.

2.3 Test Equipment Design

A test apparatus incorporating a pair of scissor blades is proposed and implemented to investigate forces and strains experienced by the blades during a cutting procedure. The primary requirements of the testbed were to,

1) Indirectly measure interface forces using existing sensing technology.
2) Allow distinction to be made between materials of known elastic properties.
3) Allow characterisation of strains experienced by blades during cutting.

The equipment used to carry out the tissue cutting experiments consists of two primary units, a scissor-cutting unit, and a data acquisition unit (Fig. 2). Scissor blades that are geometrically similar to a typical pair of scissor blades used in MIS surgery were examined during characterisation of the cutting process. The complexity of the system was minimized by mounting the scissors in a rigid fixture allowing one degree-of-freedom movement only. This approach was appropriate in this investigation as only forces perpendicular and transverse to the edges of the scissor blades are of interest at present. A pair of 18 cm straight blade Metzenbaum-Nelson scissors was used as the cutting instrument for this investigation as shown in Fig. 2(a). These scissors was deemed appropriate as they have been used in previous force measurement investigations by Greenish [8] and therefore data is available for



(a) (b)

Figure 2: (a) Experimental setup (b) Data acquisition architecture

comparative purposes. Standard strain gauges (Radionics, 120 Ω, 2mm foil gauge) were bonded to the mid point of the lower scissor arm (gauge 1) and to the top edge of the same scissor blade (gauge 2). The upper arm of the scissors is securely fixed while the lower arm is free to rotate about its fulcrum. The actuation of the lower arm is achieved by means of a 32 mm double acting pneumatic cylinder. A single turn conductive plastic precision potentiometer (Vishay Spectrol® 357) was fixed to the scissor fulcrum via a coupling device for the measurement of the angular displacement of the blades. Fully closed, the scissor cutting edges form an included angle, θ, of 6.5°, but the inclusion of strain gauge 1 limits this angle to 7.5°. The cutting range of the scissor blades is from 40° to 7.5° with the cutting process completed at 10°. The design and construction of the cutting assembly offers a robust arrangement ensuring that the forces measured by the strain sensor are only those generated during the blades/tissue interaction.

2.4 Scissor Calibration

Calibration of the force measurement system was carried out to ensure meaningful force readings were obtained from the measured strain values. To achieve this, a relationship was established between the point of contact, C, and the strain readings, ε, obtained at strain gauge 1 (Fig. 3). This was accomplished by placing a miniature button load cell (model SLB-25 from Transducer Techniques®) between the blades using two specially designed securing devices which allowed the forces, F_L, to be directed perpendicular to the load cell surfaces. The upper arm of the scissors was secured in a clamping mechanism while a series of forces, F_A, were applied to the scissor arms. The strain gauge and the load cell readings were recorded for each force increment applied to the scissor arms. This resulted in a value, *k*, in

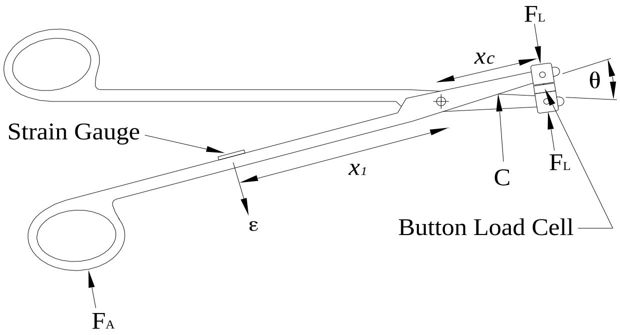


Figure 3: Scissor calibration arrangement

N/$\mu\varepsilon$, between the load cell output at x_c (mm) along the scissor blades and the strain gauge readings at the fixed point x_1, being obtained. This procedure was repeated with the load cell placed at three different locations along the blades resulting in three different k values. However, the dimensions of the load cell placed a restriction on how near the load cell could be placed to the fulcrum with the result that k values close to the fulcrum could not be measured directly. The unknown k values were obtained as follows; the three known values were plotted and extrapolated over the working length of the scissor blades resulting in the linear equation,

$$k = -0.0059x_c + 0.3072 \tag{1}$$

This equation yields a k value for any position x_c along the cutting edges of the blades. The angle θ can be measured directly, hence, a relationship between this angular displacement and x_c allows a k value to be assigned to any angle and consequently any distance x_c along the length of the blade. Both θ and x_c were recorded at incremental steps of 2° over the cutting range of the blades (40° to 7.5°). When plotted the results yielded the following power relationship,

$$\theta = 560.38x_c^{-1.1069} \tag{2}$$

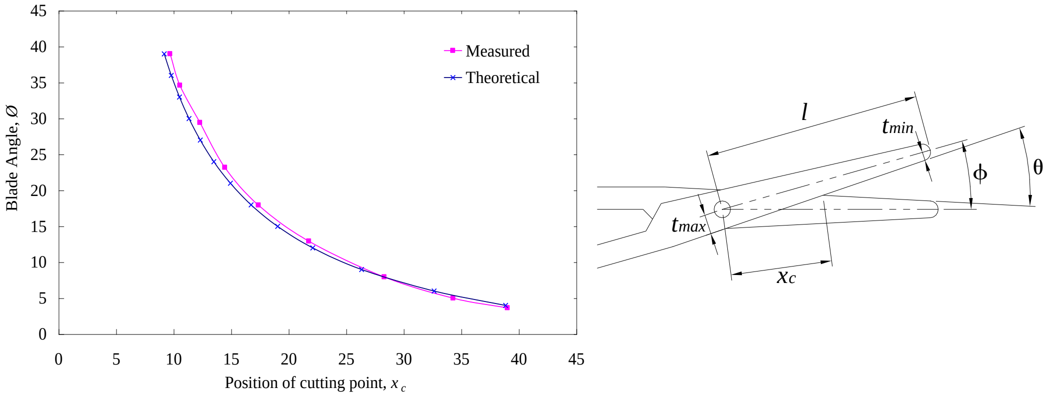
A comparison can be made between the empirically derived relationship in (2) and the theoretical relationship by examination of the blade geometry in Fig. 4 using,

$$x_c = \frac{t(l)}{\sin(\frac{\phi}{2})} \tag{3}$$

Where $t(l)$ is the thickness of the scissor blades which varies along its length and ϕ is the angle between the centre lines of the blades. It is assumed that t varies linearly between t_{max} and t_{min} [16], thus yielding,

$$t(l) = t_{\max}\left[\frac{Tan(\frac{\phi}{2})}{Tan(\frac{\phi}{2}) + \frac{t_{\max} - t_{\min}}{l}}\right] \tag{4}$$

Comparing both the theoretical and measured values (Fig. 4(a)) it can be observed that there are small variations due to the assumption that the blade cutting edges are perfectly straight.



(a) (b)

Figure 4: (a) Variation in cutting point location with angular displacement
(b) Scissor blade geometry

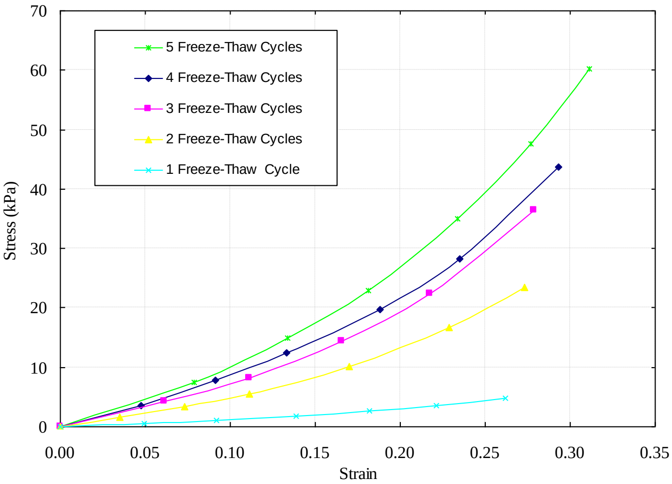
Examination has shown that there is a slight irregular curvature along the length of the blade. A database was created wherein the appropriate k value was assigned to every angle in increments of 0.001° between 40° and 7.5°. The resultant force F_C at the point of contact, C, is expressed as,

$$F_C = \text{Measured Strain} \times k\ (\theta) \tag{5}$$

where k is a function of the measured blade angle θ.

2.5 Tissue Preparation

Synthetic tissues are used in the assessment of instrument/tissue interaction forces as a practical substitute to *in vivo* or *in vitro* biological tissue samples [12]. A range of homogeneous samples of differing elastic properties are produced using a carefully mixed solution and exposing them to a varying number of freeze-thaw cycles. A 10 wt.% Polyvinyl Alcohol (PVA) cryogel solution was achieved by dissolving 40g of PVA powder (Sigma-Aldrich, 99+% Hydrolyzed) in 360g of de-ionized water [17]. The solution was mixed on a magnetic stir plate at 90°C for 30 minutes, then removed and stirred for a further 30 minutes until cooled to room temperature. Once cooled the solution was weighed and additional de-ionized water added to ensure a 10 wt.% was achieved. Five samples were prepared by pouring the PVA solution into five identical moulds giving a sheet of PVA tissue 180 mm × 80 mm × 3 mm. A cylindrical test sample, measuring 22 mm × 20 mm diameter, was exposed to the same freeze-thaw cycle as each of the tissue sheets. These cylindrical specimens were included so that mechanical properties could be obtained using appropriate testing procedures. An average Young's Modulus value was obtained for each cylindrical sample by subjecting them to standard compression testing using a materials testing machine (Lloyd Instruments™ LRK30) with a 500 N load cell. The stress/strain relationship for each cylindrical sample is plotted in Fig. 5(a) and the mean Young's modulus values are summarised in Fig. 5(b). This range of Young's Modulus values (13.37 to 155.17 kPa) is comparable to that of a range of *intra vitam* bovine biological tissues such as Liver, Spleen and Kidney (10 to 85 kPa) that were measured by Maaß [18].



Number of Freeze-Thaw cycles	Young's Modulus (kPa)
5	155.17 ± 36.45
4	102.04 ± 24.6
3	84.98 ± 23.4
2	57.55 ± 15.2
1	13.37± 2.7

(a) (b)

Figure 5: (a) Stress/strain data for PVA samples (b) Mean Young's modulus values

3. RESULTS

Cutting measurements were obtained from each of the samples immediately after they had been removed from the de-ionized water to prevent dehydration, which would result in changes in tissue properties. The angular velocity at which all samples were cut was maintained constant at 22.7 deg·s^{-1} to ensure the conditions would be as consistent as possible, and allow comparisons to be made between samples. This speed was chosen based on data collected by Greenish [8] which found that the average speed during actual surgical cutting procedures was found to be in the range from 7 deg·s^{-1} to 44 deg·s^{-1} depending on the material being cut. Each sample was subjected to one complete cutting cycle involving the blades closing from 40° to 7.5° and returning to the fully open position again, resulting in a 32 mm long cut. Data from three different cuts from each sample set were compared to establish the repeatability of the measurement system. An example of the closeness of agreement after three cuts for one sample set is illustrated in Fig. 6. Each cutting cycle consists of a number of different stages [19]. Engagement is the initial phase, where by the blades make contact with the tissue. This is illustrated in Fig. 7 by the sudden rise in the force reading at t = 0.01 s. There is then a phase from t = 0.01s to 0.06s during which the tissue goes through elastic

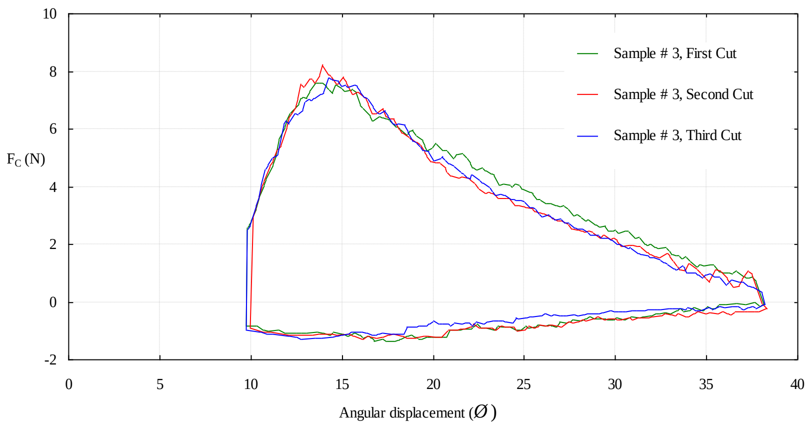


Figure 6: Repeatability analysis for a given sample

deformation. From this point onward there is a combination of phases such as plastic deformation and intercellular fracture followed by separation along the line of the scissor blades. This process continues along the cutting edges, until the end of the cut is reached at 10°. Beyond this angle it is not feasible to measure the forces using the current method as the forces generated from 10° to the fully closed position of 7.5° are not as a result of contact forces but are due to frictional forces between the blades only.

Force/displacement curves were created using the data collected from the sensorised scissors

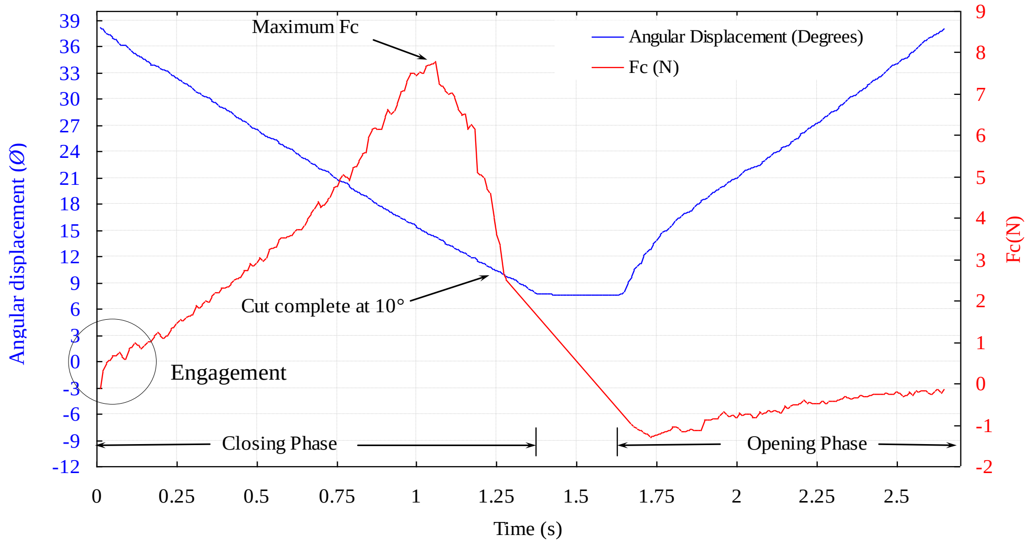


Figure 7: Typical cut characteristics

while cutting each of the five different tissue samples. It is observed that the force distribution along the blades for the softest sample (one freeze-thaw cycle) followed a similar profile to that for the empty cut with only a slight increase in magnitude. This was expected as the Young's modulus value for this sample was quite low. For each of the other samples the contact force continuously increases along the blade length, with a significant increase in the maximum force towards the end of the cutting cycle. This may be due to the fact that as the angle of the scissor blades decreases it has the effect of squeezing the tissue resulting in an extended plastic deformation phase before fracture and separation stages [8]. It is reasonable to assume that the forces generated due to deformation of the tissue with the blades would be greater than those generated during fracture, warranting further investigation. It was expected that the location of the maximum contact force would be at the tip of the blades corresponding to an angle of 10°, however, it can be seen from Fig. 8 that the maximum force occurs before the end of the cut. An explanation for this may come from the fact that, during cutting, the tissue is being pushed forward slightly due to the longitudinal components of the forces acting on the blades. Towards the end of the cut the tissue may slide off the blade tip instead of being cut, accounting for the sudden drop in contact force after the maximum is obtained.

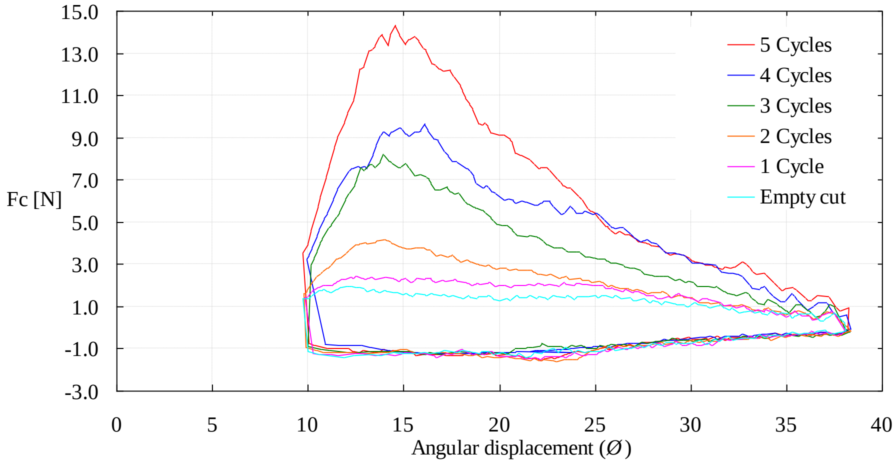
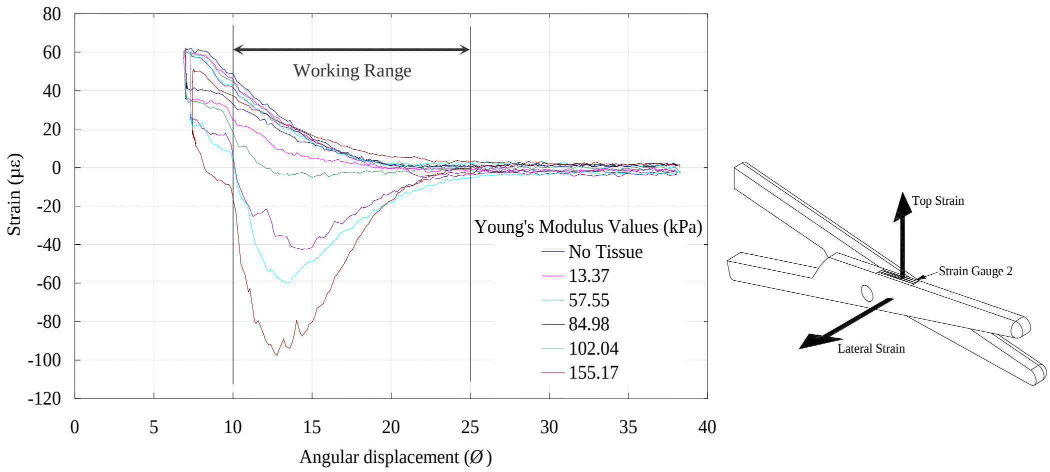


Figure 8: Force-displacement curves for five PVA samples

3.1 Blade Analysis

A miniature strain gauge (gauge 2) was bonded to the upper edge of one blade to directly measure the strains experienced by the blades during a cutting procedure. The strains ranging from +62 $\mu\varepsilon$ to -121 $\mu\varepsilon$ were measured and are graphically presented in Fig 9(a). These results are consistent with the results of Fig. 8 which employs an indirect method of



(a) (b)

Figure 9: (a) Strain variation during cutting cycle (b) Direction of strains on blade

estimating the strains and hence the forces experienced by the blade during a cutting cycle. It is important to note that in terms of sensor selection, the resolution of the measurement system is of greater significance over the typical working range (25° to 10°) used by a surgeon. The selectivity of the measurement system, developed in this work, within this working envelope, is highlighted in Fig. 9(a). Although the strain gauge primarily caters for the measurement of the strain experienced on top of the scissor blade, it inadvertently experiences some lateral strain (Fig. 9(b)) as a result of the curvature of the blade along its length. Future work will involve dual measurement of the strains on the cutting face as well as the cutting edge in an attempt to fully quantify the interface characteristics.

4. DISCUSSION

The force/displacement curves obtained from the test equipment exhibited typical scissor cutting characteristics such as tissue engagement, elastic deformation, plastic deformation, fracture and separation. It was observed that the maximum force consistently occurred before the cutting process was completed, indicating possible tissue slippage towards the tips of the blades. The force displacement profiles are in general agreement with those published in other literature. It is difficult to compare directly as data regarding Young's modulus values and sample thicknesses were neither measured nor indicated for all samples in the available literature.

Measurement of the true forces experienced during an actual cutting procedure is of importance to researchers involved in the design and development of virtual surgical training systems. A major difficulty in the design of safe surgical systems is the lack of good data on soft tissue behaviour [2]. It is envisaged that the data collected using this apparatus will assist in advancing the development of reality based cutting models to be incorporated into 3-D virtual simulators.

A future application of this force measurement testbed will involve assessing the feasibility of new direct force sensing technologies for applications in telerobotic surgery. This current testbed allows new sensing technologies to be readily placed at the location where the forces are being generated while allowing a series of controlled experiments to be carried out. The effects of different blade geometries on force distribution can be analysed by changing the type of scissors being examined.

5. CONCLUSIONS

The purpose of force sensing in MIRS systems is to overcome the reduced tool-tissue sensation that is otherwise present in normal open surgery with a view to improving the quality of surgical interventions. Miniature force sensors need to be placed as close as possible to the interaction site to allow surgical instrument end effectors to measure these interaction forces.

The main aim of this work was to develop a test apparatus that caters for the collection of force and strain data which could be used in the characterisation of the tissue-blade interface during cutting. Experiments have shown that the testbed can be used to distinguish between materials of differing properties as well as highlighting typical scissor cutting trends. It is evident that decoupling of the direct and lateral forces requires further investigation so as to fully quantify the interface characteristics. This data will be used to establish a set of parameters for the development of an appropriate force measurement strategy incorporating a sensor that is biocompatible as well as electromagnetic compatible, sterilisable and has orthogonal force measurement capability without affecting the functionality of the instrument.

REFERENCES

1. A. R. Lanfranco, A. E. Castellanos, J. P. Desai, and W. C. Meyers, "Robotic surgery: a current perspective," *Ann. Surg.,* vol. 239, pp. 14-21, Jan. 2004.
2. F. Tendick, S. S. Sastry, R. S. Fearing, and M. Cohn, "Applications of micromechatronics in minimally invasive surgery," *IEEE/ASME Trans. Mechatronics,* vol. 3, no. 1, pp. 34-42, March 1998.
3. A. M. Okamura, "Methods for haptic feedback in teleoperated robot-assisted surgery," *Ind. Rob.,* vol. 31, no. 6, pp. 499-508, Dec. 2004.
4. G. S. Guthart, and J. K. Salisbury Jr., "The Intuitive™ telesurgery system: overview and application," *Proc. IEEE Int. Conf. Rob. Autom.*, San Francisco, CA, pp. 618-621, 2000.
5. T. J. Ortmaier, "Motion compensation in minimally invasive robotic surgery," Ph.D. dissertation, Tech. Univ. Munich, Germany, 2003.
6. C. R. Wagner, N. Stylopoulos, and R. D. Howe, "The role of force feedback in surgery: analysis of blunt dissection," *Proc.10th Symp. on Haptic Interfaces for Virtual Environment and Teleoperator Sys.,* pp. 68-74, 2002.
7. M. Tavakoli, A. Aziminejad, R. V. Patel, and M. Moallem, "Tool/tissue interaction feedback modalities in robot-assisted lump localization," *28th IEEE Int. Conf. in Med. and Bio. Society,* New York, pp. 3854-3857, 2006.
8. S. Greenish, "Acquisition and analysis of cutting forces of surgical instruments for haptic simulation," Masters Thesis, Dept. of Elec. and Comp. Eng., McGill Univ., Montréal, Canada, 1998.
9. M. Mahvash, and A. M. Okamura, "A fracture mechanics approach to haptic synthesis of tissue cutting with scissors," *Proc. 1st Conf. and Symp. Haptic Interfaces for Virt.*

Environment and Teleoperator Syst., pp. 356-362, 2005.

10. V. B. Chial, S. Greenish, A. M. Okamura, "On the display of haptic recordings for cutting biological tissues," *10th Symp., Haptic Interfaces for Virtual Environment and Teleoperator Systems*, Orlando, pp. 80-87,2002.
11. N. Zemiti, G. Morel, T. Ortmaier, and N. Bonnet, "Mechatronic design of a new robot for force control in minimally invasive surgery," *IEEE/ASME Trans. Mechatronics*, vol. 12, no. 2, pp. 143-153, Apr. 2007.
12. G. Tholey, A. Pillarisetti, and J. P. Desai, "On-site three dimensional force sensing capability in a laparoscopic grasper," *Ind. Rob.*, vol. 31, no. 6, pp. 509-518, Dec. 2004.
13. B. Kuebler, U. Seibold, and G. Hirzinger, "Development of actuated and sensor integrated forceps for minimally invasive robotic surgery," Int. J. Med. Robotics and comp. assisted surg. vol. 1, pp. 96-107, Sept. 2005.
14. J. Peirs, *et al*, " A micro optical force sensor for force feedback during minimally invasive robotic surgery," *Sensors and Actuators A*, vol. 115, no. 2-3, pp. 447-455, Sept. 2004.
15. G. Tholey, T. Chanthasopeephan, T. Hu, J. P. Desai, and A. Lau, "Measuring grasping and cutting forces for reality-based haptic modeling,"*17th Int. Congress and Exhibition Computer Assisted Radiology and Surgery,* London, U.K., vol. 1256, pp. 794-800, 2003.
16. A. M. Okamura, R. J. Webster III, J. T. Nolin, K. W. Johnson, and H. Jafry, "The Haptic Scissors: Cutting in Virtual Environments," *Proc. 2003 IEEE Int. Conf. on Robotics & Automation*, Taipei, Taiwan, pp. 828-833, 2003.
17. A. Kharine, *et al*, "Poly (vinyl alcohol) gels for use as tissue phantoms in photoacoustic mammography," *Phys. Med. Biol.* Vol.48, pp. 357-370, 2003.
18. H. Maaß, and U Kühnapfel, "Noninvasive Measurement of Elastic Properties of Living Tissue" *13th Int. Congress on Computer Assisted Radiology and Surgery (CARS '99)*, Paris, pp 865-870, June 23-26, 1999.
19. S. M. Shimi, "Dissection techniques in laparoscopic surgery: a review," *J. Roy. Coll. Surg. Edinb.*, vol. 40, no. 4, pp. 249-259, Aug. 1995.