A Force Measurement Evaluation Tool for Telerobotic Cutting Applications: Development of an Effective Characterization Platform

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Abstract—Sensorized instruments that accurately measure 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. Although there is ongoing research into force measurement involving surgical graspers little corresponding effort has been carried out on the measurement of forces between scissor blades and tissue. This paper presents the design and development of a force measurement test apparatus, which will serve as a sensor characterization and evaluation platform. The primary aim of the experiments is to ascertain whether the system can differentiate between tissue samples with differing mechanical properties in a reliable, repeatable manner. Force-angular displacement curves highlight trends in the cutting process as well the forces generated along the blade during a cutting procedure. Future applications of the test equipment will involve the assessment of new direct force sensing technologies for telerobotic surgery.

Keywords—Force measurement, minimally invasive surgery, scissor blades, tissue cutting.

I. INTRODUCTION

MINIMALLY invasive surgery (MIS) is an operating technique in 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, muscle and tissue within the body while observing the images on a 2-D monitor. The primary advantages of this technique include smaller incisions, shorter hospital stays, lower risk of infection, and convalescence is also significantly reduced [1].

Robot-assisted surgery has revolutionized the way in which surgeons carry out minimally invasive surgical procedures. It assists surgeons in overcoming the drawbacks associated with traditional MIS procedures such as hampered dexterity, reduced accuracy, and a loss of 3-D visualization [1]. As well as overcoming the disadvantages of MIS techniques robotic assistance offers new advancements in areas such as; provision of additional degrees-of-freedom, tremor filtering

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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 minimally invasive robotic surgery (MIRS) systems is the daVinci™ from Intuitive-Surgical® Inc. [4]. This system has been evaluated mainly in the field of minimally invasive heart surgery, but further applications will be established in future. This type of 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]. This lack of force feedback leads to the difficult task for the surgeon of interpreting organ deformation as a measure of the forces at the slave side [5]. This in turn is quite taxing on the surgeon, which can 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 in [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 in [3], with and without, haptic feedback. These experiments indicated conclusively that haptic feedback is advantageous, and therefore desirable, in robot-assisted surgical systems. Thus, for any dexterous manipulations such as cutting, grasping, suturing or dissection, force sensing needs to be incorporated into the surgical instruments being used.

To date little research has been undertaken 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 in [16] 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 in [19], [20] for virtual simulation. Experiments were carried out in [21], [22] using scissors to ascertain the fracture toughness of various biological tissues. Forces acting on the blades were measured by the placement a load cell between the scissor

handles and an actuator.

This work focuses on the direct 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 test-bed has been developed that will allow for the direct measurement of the forces and determination of the actual forces being generated at the instrument/tissue interface.

The remainder of this paper is organized as follows. An overview of the current methods used in measuring contact forces between instrument end-effectors and the environment with which they are interacting is dealt with in Section II. This is followed by a more specific look at the role of cutting with scissor blades and why this area was chosen for this particular work. A description of the test apparatus used to measure the contact forces experienced along the scissor blades is included in Section III. Following this we present the data collected from our experiments and highlight the capability of the test apparatus in distinguishing between materials with differing mechanical properties. A discussion and review of the main results of the current work and a description of ongoing and future work is contained in Section V.

II. FORCE MEASUREMENT

A. Direct and Indirect Measurement

A major obstacle to the provision of force control in MIRS systems lies in the actual measurement of the forces [8]. It is this problem specifically which has motivated this research endeavor which aims to explore the possibility of enhancing the force measuring capabilities of commonly used end-effectors such as graspers and scissors. The decision of 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. outside of 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 [8]-[10], [12]-[14].

As previously outlined none of the currently available MIRS systems provide kinesthetic feedback due to a lack of appropriate force sensing capabilities with the instruments being used. Indeed small and sterilizable force sensors, which could be inserted into the patient, are still a requirement [8]. Experiments were carried out in [9] that validated the requirement for direct force sensing. They highlighted how the indirect technique overestimated the tissue grasping force by an order of magnitude compared to the direct sensing method. The most suitable solution, but technically the most challenging, is to integrate a miniaturized force sensor at the instrument tip [12]. The German Aerospace Centre (DLR) [10] has developed one of the most promising direct sensing

methods. The sensor is incorporated between a gripper and the trocar shaft and allows measurement of instrument/tissue interaction forces. An indirect measurement approach has been adopted in [8] which locate a six-axis force/torque sensor outside the patient. The frictional effects have been reduced by a novel double-barreled trocar design. For both groups, experimental work involving the use of scissor end-effectors does not appear to have been the subject of investigation as of yet.

B. Scissor/Tissue Interaction

Cutting and grasping of tissues are two fundamental tasks that are commonplace in any surgical procedure [11]. 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. In order to carry out these tasks effectively and safely it is desirable for the surgeon to be able to feel the various forces being experienced by the blades. During open surgery this sense of feel is not a problem as the scissors are in direct contact with the surgeon's hands. In MIS surgery these forces cannot be relayed to the surgeon, with any real accuracy, due to the interference in the trocar. This interference makes tissue manipulation more heavily dependent on reduced 2-D visualization. However, MIRS systems, as mentioned previously, can overcome these drawbacks encountered by traditional MIS, as well as introducing a number of further advantages. Force measurement and haptic feedback allows precise movements, the application of optimum forces, identification of tissue abnormalities or other anatomical information, as well as giving the surgeon a sense of immersion during the surgical procedure.

Although there are a wide range of tools available for various surgical procedures this paper focuses specifically on the interaction forces between a pair of scissor blades and the tissue being cut. To date little research has been carried out on the measurement of the forces generated at the point of contact between a pair of scissor blades and the tissue with which they are interacting. We have carried out a set of experiments that allow measurement of the force distribution along a set of Metzenbaum scissor blades while cutting through a series of Polyvinyl alcohol (PVA) cryogel samples with differing mechanical properties.

III. EXPERIMENTAL SETUP

A. Test-Bed Development

A number of primary requirements were laid out at the test-bed design stage so that the arrangement could be deemed a suitable force measurement, characterization, and evaluation

tool. It is necessary that the real-time forces and angular displacement generated during the cutting process be measurable. This test-bed will be used in the future to assess new sensing technologies in the measurement of the contact forces at the instrument-tissue interface. It is therefore logical to take initial force readings using tried and tested sensing methods such as bonded resistance strain gauges. It is imperative that the test equipment can differentiate in a quantifiable manner between tissue samples with differing mechanical properties. To achieve this, a series of PVA cryogels of increasing elastic moduli were cut and the force/angular displacement characteristics recorded using appropriate technologies. A number of cutting experiments were performed on each of the different tissue samples in order to establish the repeatability of the measurements.

Various test-bed arrangements have been previously employed to establish various properties of different biological and synthetic materials as well as the force generated at the interaction between tissue and instrument. A test apparatus was developed in [23] to estimate the mechanical properties of different materials by subjecting them to a series of uni-axial stretching tests. A load cell was used to measure the force that the tissue was subjected to while its linear displacement was monitored. This apparatus was shown to be a very effective means of collecting data to establish design parameters for teleoperative surgical systems as well as for surgical simulation applications.

An automated laparoscopic grasper for the characterization of grasping and cutting tasks was developed in [11]. To ensure accurate measurement of the cutting forces, the jaws of the grasper were first calibrated by placing a load cell against one of the jaws and forming a relationship between forces and actuating current. The grasper set-up was evaluated by grasping hydrogels of varying elasticity and distinguishing between each one based on the force feedback from the grasper. Further tests were carried out on samples to assess repeatability of the measurement arrangement.

Equipment for measurement of the forces and torque exerted on a scalpel blade during one degree-of-freedom tissue cutting were also designed and developed in [11]. The acquired force displacement graphs highlighted a characteristic deformation followed by a localized crack extension pattern. This data was later used to verify finite element models that would be used to create a reality-based model for real-time medical simulation.

B. Test Equipment Design

The equipment used to carry out the tissue cutting experiments consists of two primary units, a scissor-cutting unit, and a data acquisition unit. In order to characterize the cutting process, scissor blades that are geometrically similar to a typical pair of scissor blades used in MIS surgery were examined. 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 the contact forces perpendicular to

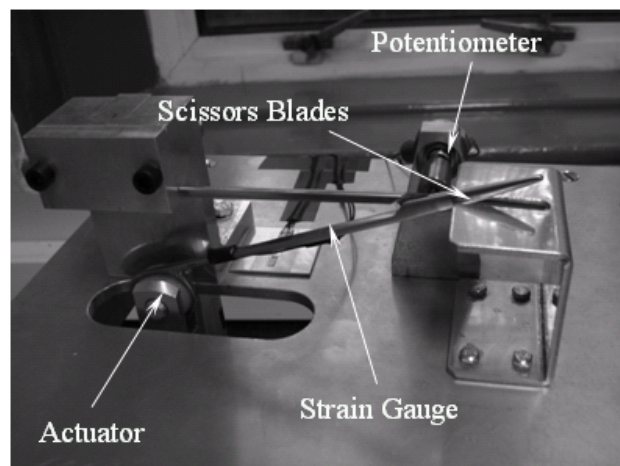the edges of the scissor blades are of interest.

Fig. 1 Scissor cutting unit

A Pair of 18 cm straight blade Metzenbaum-Nelson scissors (nopa® instruments) were used as the cutting instrument for this investigation (Fig. 1). These scissors were deemed appropriate as they have been used in previous force measurement investigations [16] and therefore data is available for comparative purposes. A standard strain gauge (from Radionics, stock no. 632-168) was bonded to the lower arm of the scissors at its midpoint. The gauge has a nominal resistance of 120 Ω and base dimensions of 9mm $\times$ 3.5 mm, small enough to be attached to the inner surface of the scissor arm. 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 diameter double acting pneumatic cylinder (Festo, model DSNU-32-100-P-A) with a maximum force output of 322 N. Adjustment of a unidirectional flow control valve at the entry to the cylinders' upper and lower chambers controls the linear velocity of the piston rod. A single turn conductive plastic precision potentiometer (Vishay Spectrol® model 357), for the measurement of the angular displacement of the blade cutting edges, was fixed to the scissor fulcrum via a coupling device. Fully closed, the scissor cutting edges form an included angle, θ (Fig. 3) of 6.5°, but with the strain gauge bonded to the inner surface of the lower blade this angle is increased to 7.5°. The cutting angle 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 attained during the blades/tissue interaction.

The data acquisition unit used in this system collected analogue signals from the active strain gauge and the precision potentiometer. As a result of limited space on the scissor arm the strain gauges were arranged in a quarter bridge configuration with a dummy gauge for temperature compensation. This three-wire arrangement was connected to

a National Instruments® (NI) SCC-SG02 strain gauge module, which provided bridge completion. The bridge filtered the signal through a 1.6 kHz lowpass filter and amplified the signal by 100 to give readable strain values. Bridge offset nulling was also included in the module by adjustment of a built-in potentiometer. This strain gauge module was inserted into an NI SCC 68 module holder, which also accepted analogue output signals from the precision potentiometer (Fig. 2).
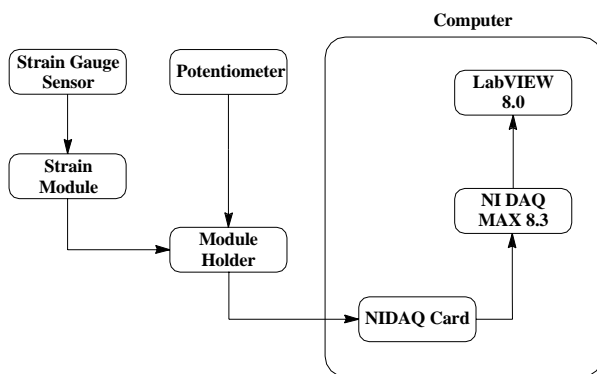

Fig. 2 Data acquisition architecture

These analogue signals were converted into useable digital readings by connecting the SCC 68 module holder to a PCI-6221 NIDAQ card, which was installed on a standard PC (Dell optiplex GX150). The software used included NIDAQ MAX 8.3, which configured the devices, sub-devices and channels, as well as set up tasks. LabVIEW® 8.0 was used to condition and display the acquired data.
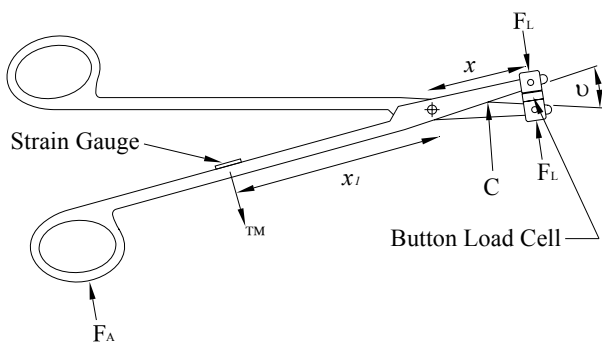
C. Scissor Calibration

Fig. 3 Scissor geometry

Calibration of the force measurement system was carried out to ensure that correct force values along the blade as it cut through the various tissue samples were obtained. The aim of this calibration process was to establish a relationship between the point of contact, C, and the strain readings, ε, obtained at the strain gauge (Fig. 3).This was achieved 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 readings and the load cell readings were both recorded for each force increment applied to the scissor arms. This resulted in a value, k in N/$\mu\varepsilon$, between the load cell output at x (mm) along the scissor blades and the strain readings at the fixed point x_1, being obtained. This procedure was carried out with the load cell placed at three different locations along the blades resulting in three different k values. However, the height of the load cell placed a restriction on how close 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,

$$k = -0.0059x + 0.3072 \qquad (1)$$

This equation yields a k value for any position x along the cutting edges of the blades. To assign the correct k value for a given distance x, a relationship between the angular displacement of the blade edges, θ, and x was established. The single turn precision potentiometer was used to measure angular displacement while a Vernier Calipers was used in the measurement of the distance x. The plotted results yielded the power relationship,

$$\theta = 560.38x^{-1.1069} \qquad (2)$$

A database was created wherein the appropriate k value was assigned to every angle in increments of 0.001° between 7.5° and 40° (which is the measuring range of the scissor blades used). The resultant force F_C at the point of contact, C, is expressed as,

$$F_C = \text{Measured Strain} \times k\,(\theta) \qquad (3)$$

where, k is a function of the measured blade angle θ. The force-displacement relationship and the strain-displacement relationship for a typical cut are illustrated in Fig. 4. The scissors are initially in the fully open position at 40°; they are then closed to 7.5° and again opened fully to complete one cut cycle. It is observed that the contact force, F_C, as expected is not proportional to the strain reading given that the k values closer to the fulcrum are greater than those towards the tip of the blades.
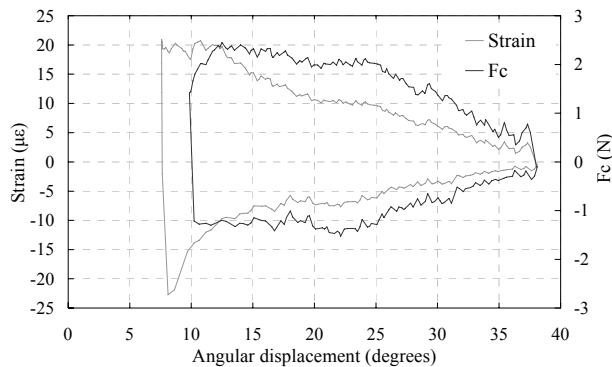
Fig. 4 Force and strain comparison

Fig. 5 Stress strain data for PVA samples

D. Tissue Preparation

A number of homogeneous tissue samples of varying stiffness were prepared using Polyvinyl Alcohol (PVA) cryogel. By exposing each of the samples to differing freeze-thaw cycles, a range of samples were obtained. A 10 wt.% PVA solution was achieved by dissolving 40 g of PVA powder (Sigma-Aldrich, 99+% Hydrolyzed) in 360 g 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, with the mean Young's modulus values given in Table I.

IV. EXPERIMENTAL RESULTS

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 keep the results as consistent as possible. This speed was chosen based on data collected by [16] which found that the average speed during cutting was found to be in the range from 7 deg·s^{-1} to 44 deg·s^{-1} depending on the material being cut. During cutting, each sample was clamped to a support platform that aligned the lower plane of each tissue sample with the cutting edge of the fixed lower scissor blade. Despite the samples being securely clamped to the support platform, there was a
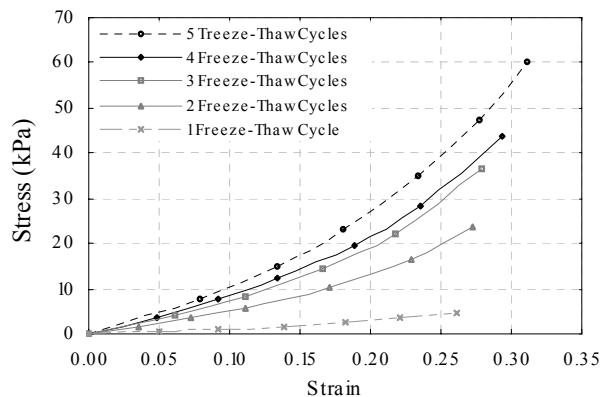
TABLE I
MEAN YOUNG'S MODULUS VALUES

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

tendency for the tissue to slide along the scissor blades at the initial point of engagement due to a small degree of local elastic deformation of the sample. This resulted in the cutting process initiating when the cutting blade edges were at approximately 38° and not 40°, the angle at which the blades are fully open. 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 assess the repeatability of the system. An example of the closeness of agreement after three cuts for one sample set is illustrated in Fig. 6.
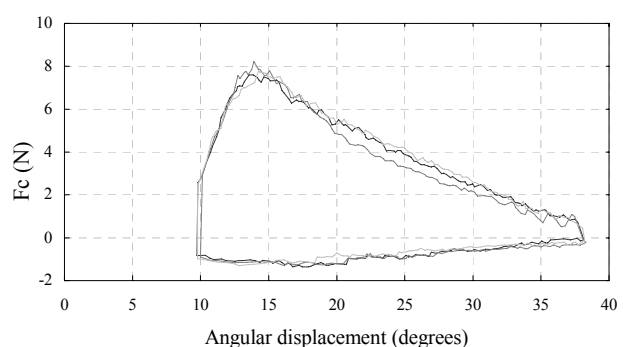

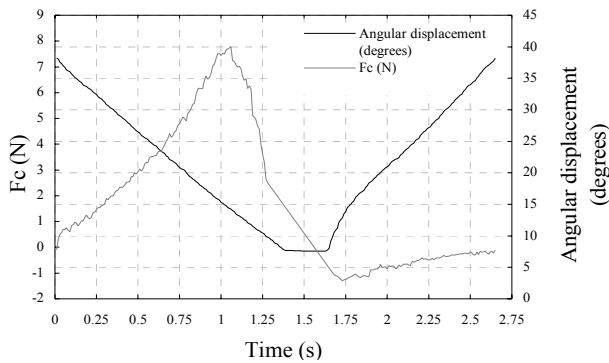Fig. 6 Repeatability analysis for a given sample

Fig. 7 Displacement and force data

Fig. 8 Force-displacement data

Each cutting cycle consists of a number of different stages [18]. Engagement is the initial phase whereby the blades make contact with the tissue. This is illustrated in Fig. 7 by the sudden rise in the force reading at 0.01 s. There is then a phase from 0.01s to 0.06s in which the tissue goes through elastic 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. However the frictional effects can be approximated using the current method if it is assumed that the friction occurs at the point where both blade edges meet.

Force-displacement curves were created using the data collected from the sensorized scissors while cutting each of the five different samples (Fig. 8). These curves clearly highlight the ability of the test equipment to distinguish between material samples with a range of elastic moduli values. It is observed that the force distribution along the blades for the softest sample (1 freeze-thaw cycle) followed a similar profile to the empty cut with 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 that fact that, as the angle of the scissor blades decreases it has the effect of squeezing the tissue [16] resulting in an extended plastic deformation phase before the fracture and separation stages. 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,
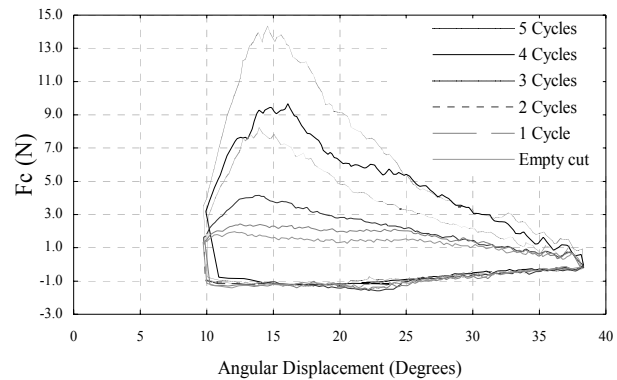
during cutting, the tissue is being pushed forward slightly due to the longitudinal components of the forces acting on the blades. Towards the end of the cut the tissue may slide off the blade tip instead of being cut, accounting for the sudden drop in contact force after the maximum is obtained.

V. DISCUSSION AND FUTURE WORK

A force measurement evaluation apparatus has been designed and developed which can cater for the characterization of scissor-cutting procedures on synthetic tissue samples with known elastic properties. Measurement and acquisition of the forces experienced by the scissor blades, as well as the angular displacement of the blades, during the cutting procedure, were the major system requirements. The test apparatus had to be robust, accurate and capable of discriminating between homogeneous tissue samples of varying mechanical properties. Calibration of the scissors was carried out so that a correlation could be established between the forces experienced along the blade edges and the measured strain readings on the scissor arms. Results from tests carried out at a constant angular velocity of 22.7 deg·s⁻¹ revealed that the repeatability of the force measurements for samples with the same properties were good. 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 appeared before the cutting process was completed, indicating possible tissue slippage towards the tips of the blades. Data showed that the cutting apparatus was clearly able to distinguish between the varying tissue samples. The force displacement profiles are in general agreement with those published in other literature. It is difficult to compare directly as there was limited data available regarding Young's modulus values and sample thicknesses.

Future development of the test-bed will involve the introduction of an electrical linear actuator to open and close the scissor blades. This actuator's advanced capabilities will allow cutting speeds to be varied which will permit the blades

to operate at velocities similar to those applied by a surgeon during a laparoscopic cutting procedure. This experimental work was carried out using isotropic, homogenous materials of the same thickness to maintain consistency during each cut. In reality this is not the case, therefore, the force distribution while cutting real tissue will be acquired thus enhancing the applicability of the data.

The interaction between scissor blades and tissue is an extremely complex one with each cut consisting of intricate contact and fracture mechanics due to continually changing blade angles. The problem is compounded further when the blade type is changed from being smooth and straight to having serrated or curved edges. It has been widely acknowledged by a number of researchers that the ideal location for measurement of real forces between tissue and instrument is by placing the force sensor as close as possible to the instrument tip. In terms of scissor blades, especially those with irregular geometries, the perfect location for force sensors would be on, or even in, the blades themselves. Locating sensors on or in the blades would allow a true reflection of the forces experienced during cutting.

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 behavior [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 test-bed will involve assessing the feasibility of new direct force sensing technologies with applications in telerobotic surgery. This current test-bed 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 analyzed by changing the type of scissors being used.

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