2007 IEEE International Conference on
Robotics and Automation
Roma, Italy, 10-14 April 2007

A Dual-Function Wheeled Probe for Tissue Viscoelastic Property Identification during Minimally Invasive Surgery

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***Abstract*—This paper proposes a novel approach for the identification of tissue properties in-vivo using a force sensitive wheeled probe. The purpose of such a device is to compensate a surgeon for a portion of the loss of haptic and tactile feedback experienced during robotic-assisted minimally invasive surgery. Initially, a testing facility for validating the concept ex-vivo was developed and used to characterize two different testing modalities - static (1-DOF) tissue indentation and rolling (2-DOF) tissue indentation. As part of the static indentation experiments a mathematical model was developed to classify tissue condition based on changes in mechanical response. The purpose of the rolling indentation tests was to detect tissue abnormalities, such as tumors, which are difficult to isolate under static testing conditions. During such tests, the test-rig was capable of detecting simulated miniature buried masses at depths of 12mm. Based on these experiments a portable device capable of carrying out similar tests in-vivo was developed. The device was designed to be operated through a trocar port and its key feature is the ability to transition between static indentation and rolling indentation modalities without retracting and changing the tool.**

I. INTRODUCTION

THIS paper presents work carried out to develop a wheeled probe capable of identifying both tissue parameters and abnormalities during Minimally Invasive Surgery (MIS). The purpose is to provide a surgeon with the ability to mechanically probe solid organs within the body with force sensitive instruments and thus classify tissue properties. Work has shown that although 1-DOF indentation tests can provide useful information regarding tissue properties, widely varying boundary conditions often result in inconclusive data. In this paper we present an ex-vivo validation for a device which has the ability to overcome this problem by employing a methodology which performs both rolling indentation and static indentation concurrently to characterise the tissue response. The former tests allow for a large area to be investigated for tissue abnormalities (i.e. areas of increased stiffness or softness) in a short period of time, while the former test allows for a more detailed analysis of tissue properties in a localised setting. In both cases the change of the normal force imparted by the tissue onto the wheel is measured.

The ex-vivo experimental test-rig and in-vivo device described in this paper differ from existing designs in that they employ a wheeled probe and have the ability to transition between static tissue indentation and rolling tissue indentation tests, thus combining the advantages of both modalities.

Manuscript received September 15, 2006.

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II. BACKGROUND

A. Minimally Invasive Surgery

In its simplest terms, MIS, or laparoscopic surgery, can be described as major or minor surgery performed through small incisions. A fibre optic camera is passed through such an incision to provide the surgeon with a field of view, and various laparoscopic instruments are inserted through one or two other incisions to perform the procedure. The primary benefit of this approach is that the use of small incisions results in less trauma for the patient [1], thus decreasing recovery time and hospitalisation costs. However, these advantages are offset by the sharp increase in the technical difficulty of any surgical procedure carried out using a MIS approach. Due to the fact that the procedure is performed from outside the body, the surgeon's ability to both see and touch the operating environment is reduced. These problems are compounded by the fact that distal dexterity is also impaired due to use of long, rigid instruments introduced through a fixed point. Even simple point-to-point movements require extensive training as directions are reversed in-vivo [2]. This reduction of visual, haptic and tactile feedback coupled with dexterity problems can lead to accidental damage of tissue [3]. Furthermore, when performing delicate procedures such as beating heart surgery, the destabilisation induced by respiratory and cardiac motion affects precise tissue-instrument interaction [4].

B. Existing Medical Robotic Devices

Several robotic systems have been developed to surmount the aforementioned problems. The *Heartlander* [5] achieves motion compensation by attaching to the epicardium of a beating heart and thus providing a stable platform for the manipulation of surgical tools. A snake robot design to improve the dexterity of a probes' distal tip is presented in [6],[7] for throat surgery. The *Highly Articulated Robotic*

Probe (HARP) is designed to access hard-to-reach anatomical targets around the heart [8]. To overcome camera vision limitations and facilitate single port biopsy a mobile in-vivo robot was developed at the University of Nebraska, Lincoln [9],[10]. The device consists of two independently drivable wheels, an adjustable focus camera and a biopsy forceps for removing tissue samples. A smart surgical instrument for minimally invasive, computer assisted knee arthroscopy is presented in [11]. The device is an adapted arthroscope which has a 25mm long distal tip which can be steered through 110º in 1-DOF. The device described in [12] is designed to provide a surgeon with information regarding the grasping forces imparted onto tissue during MIS. It consists of a 2-DOF distal tip with an integrated miniature 3-axis Force/Torque sensor.

However, these devices represent techniques to improve aspects of specific surgical procedures, not to provide a complete, integrated surgical robotic system. Such a system is the daVinci™ Surgical System manufactured by Intuitive Surgical. It operates in a master-slave configuration where the slave consists of a tele-operated patient-side cart with three or four robotic arms [13], [14] The system does successfully tackle some of the problems associated with traditional non-robotic MIS – in particular by improving distal dexterity, increasing and magnifying the field of view, motion scaling the surgeon's hand movements for improved control, reversal of the fulcrum effect and tremor removal by filtering hand movements above 6Hz [13]. However, while these represent significant improvement over existing techniques, the surgeon is still severely hindered by a complete loss of tactile and haptic feedback. For example, the ability of a surgeon to palpate tissue when searching for tumors, diseased tissue, arterial blockages or other tissue abnormalities is absent.

C. Tissue Properties

A major area of research to overcome some of the limitations of MIS is that of soft tissue property identification. Such information can be utilized for improving surgical simulations [15],[16], haptic modeling [17] and the identification of tissue abnormalities in solid organs (liver, spleen and pancreas). The device presented in [18] performs uni-axial stretching experiments to measure porcine tissue response both in-vivo (invasively) and ex-vivo. Ottensmeyer developed a device [19] to investigate the in-vivo viscoelastic properties of tissue under uni-axial small deformations. A motorized endoscopic grasper which was used to test abdominal porcine tissues in-vivo and in-situ with cyclic and static compressive loadings is presented in [20]. In [21] Carter developed a hand-held indentation probe capable of producing indentations of about 10 mm travel with force of up to 5 N. A mechanical probe developed by the Harvard Biorobotics Laboratory, [23], attempted to identify the location and properties of tumors based on static indentation tests. The findings concluded that it is necessary to consider a series of distributions measured as the probe slides across the surface of the tissue rather than a single distribution from a static indentation test.

Analysis of these devices revealed that a wheeled probe with the ability to perform both static and rolling indentation tests could potentially overcome some of the disadvantages associated with current techniques. The following sections describe the test-rig and mathematical modeling performed to validate this theory, along with a preliminary design for a device of carrying out such tests in-vivo.

III. Ex-Vivo Test Rig

The testing facility is required to allow for controlled horizontal and vertical movement of a wheeled device across the surface of an excised ovine liver sample. To facilitate this, a test-rig capable of mounting different wheel designs was constructed and attached to the distal tip of a Mitsubishi RV-6SL 6-DOF robotic manipulator. An ATI MINI40 Force/Torque sensor (SI-20-1: real world resolution 0.01N with 16-bit DAQ) was mounted at the interface between the test-rig and the manipulator end effector so as to allow for the measurement of the normal force imparted by the tissue onto the wheel, as shown in Fig. 1.

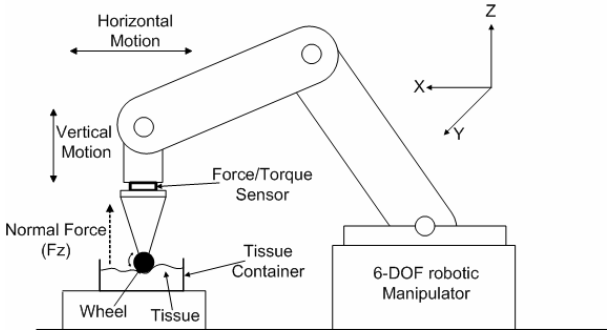

Fig. 1. Schematic of ex-vivo wheeled test rig

The left and right wheel axles are mounted in miniature ball bearings to minimize frictional losses and its rotation is measured using a Kubler Incremental encoder via a belt and pulley arrangement. Data from both the F/T sensor and the encoder was acquired using the National Instruments *LabView 8.0* software package and associated DAQ cards. The tissue is held in a container which contains a Perspex side so that the tissue deformation and wheel-tissue contact areas could be viewed and recorded during testing using a high frame rate camera.

IV. Proposed Methodology

This paper proposes a new methodology to overcome problems experienced by soft tissue models when the boundary conditions change from those used to calculate model parameters. These conditions include changes in tissue thickness and test location between samples. A solution to the problem is to calibrate the model using experimental data for each different sample tested. However, if calibration is to be carried out in this manner, there is no guarantee that the calibration test site consists of healthy, homogenous tissue which is representative of the remainder of the sample, and thus suitable to provide a calibration benchmark for further indentation tests. However, by first performing rolling tests across large areas of an organ's

surface and monitoring the tissue normal response, a healthy region can be located. This region can now be utilized to calibrate the mathematical model for use in more detailed parameter estimation during static indentation tests. Numerous rolling tests have shown that underlying areas of tissue softness and stiffness can be detected by the experimental setup as large changes in the measured F_z (normal force) profile. Consequently, by monitoring this measurement and locating a healthy and homogenous region, accurate static indentation calibration tests can then be carried out. The added advantage of this approach is that rolling indentation tests themselves can provide vital information regarding the underlying tissue structure.

V. Rolling Results

The results presented in this section describe rolling tests carried out using a cylindrical wheel of diameter 15mm and width 15mm on ovine liver samples. Although this will not replicate in-vivo conditions, the overall aim of the tests is to evaluate the feasibility of using a force sensitive wheel to palpate and probe tissue during MIS.

The focus of these specific rolling tests was twofold - to identify the difference between a healthy, homogenous tissue region and one which has underlying abnormalities. The identification of such a healthy region would allow for the calibration of a static indentation model of tissue response. The second focus of the tests was to investigate if the test-rig could detect the presence of artificially inserted solid masses. Such masses are designed to simulate small, buried cancerous tumors which could not be detected by visual means (Fig. 2).

Fig. 2. Healthy liver sample with two sets of three pins spaced 26mm apart at depths of 12mm and 8mm respectively. At these depths neither pins were detectable by visual means. The wheel (15mm diameter x 15mm width) can also be clearly seen.

Numerous tests were carried out using a cylindrical wheel of diameter 15mm and length 15mm mounted on the test rig. An indentation depth of 2.5mm was maintained during all tests and the wheel was rolled over different regions in a linear motion at a speed of 10mm/s. Experimental results showed that peaks or troughs in the measured F_z were due to an inhomogeneous test region. Consequently, healthy regions of tissue samples were identified due to the absence of such peaks and troughs. The dashed line in Fig. 3 shows such a region. In this example the wheel began rolling after 2 seconds and was lifted off the surface after 10 seconds. During this time the force profile indicates a steady decay with no dominant peaks or troughs. Once such a healthy region was identified, tests were carried out to evaluate the capability of the wheeled probe to detect small buried masses.

As can be seen in Fig. 2, during this test a sample of healthy ovine liver was sliced open along two sides and a flap pulled back. Two sets of three miniature pins were then inserted and the flap of liver was replaced such that the pins were no longer visible. The spherical pin heads are each 4.8mm in diameter. Several test runs were performed both with different pin combinations (1, 2, 3 and 4 pins in two sets) as well as with all pins removed. These pins were deemed suitable for the purposes of this feasibility study, however due to the large Young's Modulus of the plastic relative to that of liver tumors (~78GPa) [24], future work will be carried out with more realistic simulated tumors.

Fig. 3 shows the F_z measured during this test run with three pins (solid line) with the same healthy test region superimposed (dashed line). The location of the pins can be clearly identified by the two sharp peaks. Further tests showed that the test rig was capable of positively identifying sets of four, three and two pins. The measured F_z for the single pin tests did indicate the presence of the pins, however, the magnitude of the changing force profile was reduced and so the results can be considered inconclusive.

However, these results do show that a wheeled probe rolled across the surface of biological tissue can both locate regions of healthy tissue and identify underlying tissue abnormalities. Introducing such a force sensitive wheeled probe during MIS could potentially compensate a surgeon for some of the loss of haptic and tactile feedback experienced during those procedures.

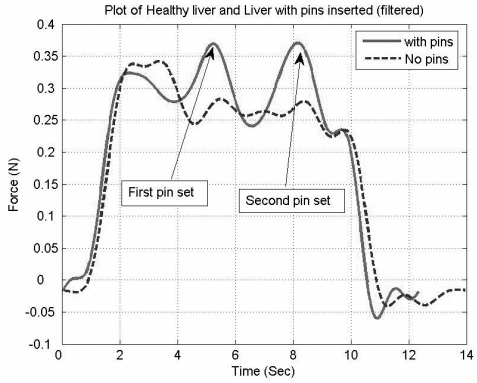

Fig. 3. This plot shows the measured F_z for a rolling test over the same tissue sample both without any simulated masses (dashed line) and with two sets of three pins (solid line) as shown in Fig. 2. The wheel used during the experiment had a diameter of 15mm and width of 15mm. It was rolled at an indentation depth of 2.5mm and a horizontal velocity of 10mm/s.

VI. Development of Static Indentation Mathematical Model

A. Purpose of Development

With the ability to identify underlying tissue abnormalities and healthy regions, a mathematical model to perform localised static indentation tests was developed. The purpose

of developing this model was for it to work in conjunction with a rolling probe to provide further analysis of abnormal tissue regions identified during preliminary rolling tests.

B. Preliminary Study

It is well known that biological soft tissues are viscoelastic materials [18],[22], and as such demonstrate the property of stress relaxation (i.e. a step constant strain results in decreasing stress). Two mechanical models commonly used to represent viscoelastic materials are the Maxwell model and Kelvin model, as shown in Fig. 4.

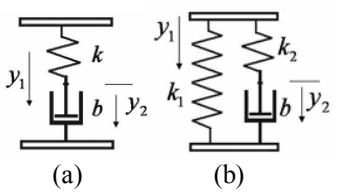

Fig. 4. Maxwell (a) and Kelvin (b) viscoelastic material models

For a unit-area Maxwell model, the stress relaxation is expressed as

$$f = k \cdot H \cdot e^{-\frac{k}{b}t}. \quad (1)$$

For a unit-area Kelvin model, the stress relaxation is expressed as

$$f = k_1 H + k_2 H \cdot e^{-\frac{k_2}{b}t}. \quad (2)$$

Where f is the force, H is the indentation depth, k, k_1 and k_2 represent tissue elasticity, b represents tissue viscosity and t represents time. In order to evaluate the suitability of the models, a series of preliminary ex-vivo static indentation experiments were carried out on ovine liver. The indentation was performed with a cylindrical wheel of diameter 20mm and length 85mm. A larger wheel was used on this occasion because of the small clearance provided by the 15mm diameter wheel and a necessity to increase the measured force signal to overcome noise. The indentation depth was varied from 1.0mm to 6.0mm in 0.5mm increments, for indentation times of 18 seconds with the contact determined by visual means. It was found that the Kelvin model most accurately replicates the behavior of the biological tissue. However, as the indentation depth increased the accuracy of the model, and in particular its ability to model the stress relaxation of the tissue, was reduced.

To investigate this further, the *Matlab* curve fitting toolbox was used to analyze the resulting experimental data. As expected, an equation modeling the Kelvin body, as shown in (3), faithfully mapped the curve for shallow indentations, but errors occurred at deeper indentations.

$$f(t) = Ae^{Bt} + C \quad (3)$$

It was found that an equation representing the sum of two exponential functions, as shown in (4), was the most suitable for modeling the response in terms of maximum accuracy with minimal complexity.

$$f(t) = Ae^{Bt} + Ce^{Dt} \quad (4)$$

In (3) and (4), t represents the time, f is the force, and A, B, C, D are coefficients calculated by the curve fitting algorithm. Table I shows a RMSE comparison of the (3) and (4) with the measured response.

TABLE I
RMSE COMPARISON OF (3) AND (4) WITH MEASURED DATA

Indent Depth	$Ae^{Bt} + C$	$Ae^{Bt} + Ce^{Dt}$
1.5mm	0.01116	0.01075
2.5mm	0.01302	0.0111
3.5mm	0.01424	0.01213
4.5mm	0.01213	0.01163
5.0mm	0.02873	0.01675

Each of the indentation depths produced an equation identical in form to (4). Analysis of resulting coefficient data revealed that the B and D coefficients remained constant at each depth, but that the A and C coefficients varied non-linearly.

Based on these observations, a new model has been developed which accounts for each of the exponential functions in the fitted curve using two Maxwell bodies operating in parallel. Due to the increasing non-linear response of biological tissue, two non-linear polynomials, $P(y_1)$ and $Q(y_1)$, which are functions of the indentation, are added to each Maxwell model. The new model is shown in Fig. 5.

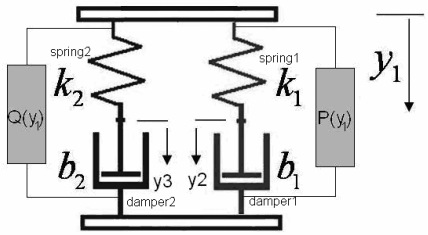

Fig. 5. The proposed model for predicting tissue response during static indentation allowing for the non-linear tissue response as indentation depth increases. Coefficients k_1 and k_2 represent tissue elasticity, b_1 and b_2 represent tissue viscosity, y_1 is indentation depth and $Q(y_1)$ and $P(y_1)$ account for the non-linear portion of the tissue response

C. Parameter Estimation

The parameters for each of the variables in the model shown in Fig. 5 have been calculated from the experimental data. The variation of coefficients A, C is shown in Fig. 6.

Both non-linear variations can be modeled using a polynomial such that (4) becomes

$$f(y,t) = (p_1 y + p_2 y^2 + p_3 y^3)e^{Bt} + (q_1 y + q_2 y^2 + q_3 y^3)e^{Dt}, \quad (5)$$

where, for this set of boundary conditions, the parameter values were calculated as $p_1 = 3.638*10^6$, $p_2 = 0$, $p_3 = 26.08$, $q_1 = 2.478*10^6$, $q_2 = 1.522*10^4$, $q_3 = 88.95$, $B = -0.8$ and $D = -0.012$. As before, f is the predicted force response, y is the indentation depth and t represents the time. Providing the boundary conditions remain similar across different tests (in this case the tissue thickness was *75mm*), the calculated parameters can be used for multiple samples. However, if

the boundary conditions change, the model must be recalibrated using the described methodology (i.e. carrying out a series of static indentations on known healthy tissue to identify the parameters).

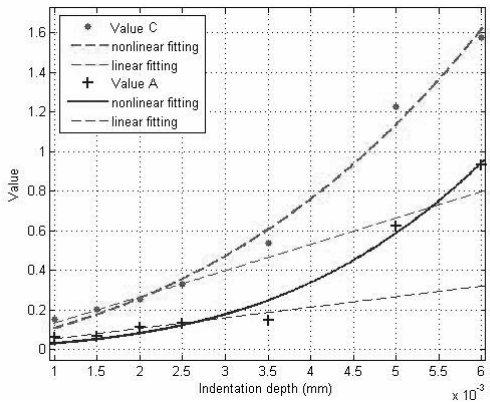

Fig. 6. Plot showing the non-linear variation of coefficient *A* and coefficient *C* with indentation depth. The figure also shows how for shallow indentations (<2.5mm) the response of the tissue is linear.

For constant indentation depth $y_1 = H$, the resulting equation as described by the new non-linear dual-Maxwell model is expressed in as follows

$$f(H,t) = P(H)k_1 \cdot H \cdot e^{-\frac{k1}{b1}t} + Q(H)k_2 \cdot H \cdot e^{-\frac{k2}{b2}t}, \quad (6)$$

where k_1, k_2 represent tissue elasticity, b_1, b_2 represent tissue viscosity, H represents a constant indentation depth, P and Q are polynomials of indentation and t represents the time. By comparing (5) and (6), the parameters of dual-Maxwell model can be defined as $k_1 = 26.08N/m$, $k_2 = 88.95N/m$, $b_1 = 32.6Ns/m$, $b_2 = 7412.6Ns/m$, $P(y_1) = (p_3y_1 + p_2y_1^2 + p_1y_1^3)/p_3$ and $Q(y_1) = (q_3y_1 + q_2y_1^2 + q_1y_1^3)/q_3$ where the coefficient values for $P(y_1)$ and $Q(y_1)$ are the same as (5).

VII. Static Indentation Results

A. Comparison of Predicted and Measured Results

These values were implemented into a simulation which produced a predicted tissue response for a given indentation depth. Fig. 7 shows the measured response and the predicted response of the proposed model, for indentation depths between 1.0 and 2.5mm.

Solid lines represent the simulated curve. Noise can be observed on the measured data. This is caused by vibrations induced by the motors in the robotic manipulator.

By comparing the simulation results and measured experimental data, it can be seen that developed dual-Maxwell model is capable of simulating the stress relaxation behavior of ovine liver.

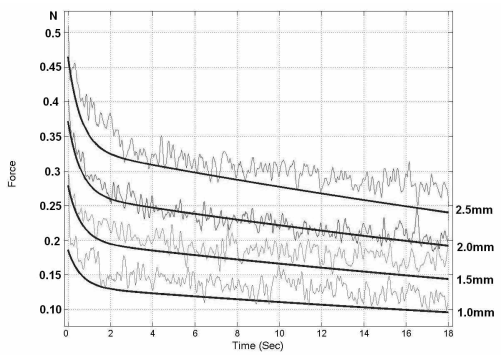

Fig. 7. Simulation results of proposed dual Maxwell Body model for indentation depths between 1.0mm and 2.5mm.

B. Linear Relationship for shallow indentation

As can be seen in Fig. 6, coefficients *A* and *C* increase in a linearly for shallow indentation depths (<2.5mm). Thus, if the wheeled probe was only expected to indent to this depth or less the non-linear polynomials can be ignored. Therefore, for shallow indentation, the dual Maxwell model can be expressed as

$$f(t) = k_1 \cdot H \cdot e^{-\frac{k_1}{b_1}t} + k_2 \cdot H \cdot e^{-\frac{k_2}{b_2}t}. \quad (7)$$

C. Comparison of Healthy Liver and Healthy / Diseased Kidney Tissue

Parameter identification was carried out for healthy kidney and liver tissue using (7) for shallow indentations. The resulting values are shown in Table II. Analysis of these parameters reveals that the kidneys' elasticity values were higher than that of the liver samples, indicating that it was stiffer to touch. Identical calculations were performed on a kidney sample which had a superficial disease. This disease caused its mechanical properties to change and consequently, the kidney was much softer to touch than a healthy sample. This is also evident from the parameter analysis.

TABLE II
COMPARISON OF MODEL PARAMETERS FOR HEALTHY AND DISEASED KIDNEY SAMPLES

Parameter	Healthy Liver	Healthy Kidney	Diseased Kidney
k_1 (N/m)	53.26	67.93	31.05
k_2 (N/m)	132.7	139.8	43.27
b_1 (Ns/m)	66.58	101.6	237
b_2 (Ns/m)	11058	11650	25453

Future work will be carried out to investigate these results using a larger sample size.

The limitation of previous models regarding changing boundary conditions can be overcome by identifying healthy tissue during preliminary rolling tests, and then calculating model parameters at that location for subsequent comparison to suspected diseased tissue. A surgeon could potentially utilize this model during MIS to probe organs which were previously unreachable due to operating constraints.

D. Limitations

Firstly, all testing was carried out ex-vivo when the tissue is not in its true physiological state. It is anticipated that additional challenges will be encountered when attempting similar tests in-vivo. Secondly, the approach of calibrating the model whenever the boundary conditions change could potentially be time consuming. However, this study demonstrates the feasibility of implementing such a device.

VIII. Portable In-vivo Test-Rig

In order to validate these results in-vivo, a portable test-rig has been developed. This device is shown in Fig. 8.

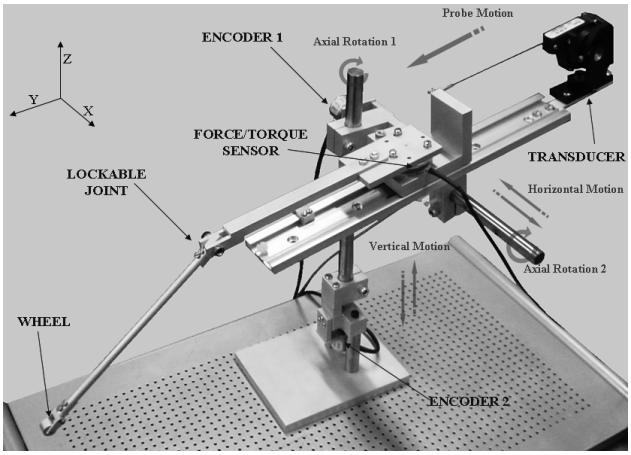

Fig. 8. Force sensitive wheeled probe developed for in-vivo tissue property measurement

The device has 5-DOF to allow for accurate positioning of the wheel mounted on the distal tip of the probe arm. The wheel is rigidly connected to an ATI MINI40 F/T sensor with a range +/-20N and a real world resolution of 5mN. This sensor is mounted on a slider rail and attached to a digital linear position transducer so that indentation depth can be recorded. Locking switches allow a desired indentation depth to be set prior to tissue contact. The position of the wheel can be recorded at all times due to incremental encoders are mounted on both axes of rotation and a scale marked along both the horizontal and vertical planes of translation. The range of travel of the probe wheel is 100mm. Future work to be carried out includes carrying out in-vivo tests to validate the ex-vivo work presented in this paper using this device and to develop a mathematical model capable of predicting tissue response during rolling.

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