Abstract—This paper presents a novel miniaturized force-indentation depth (FID) sensor designed to conduct indentation on soft tissue during minimally invasive surgery. It can intra-operatively aid the surgeon to rapidly identify the tissue abnormalities within the tissue. The FID sensor can measure the indentation depth of a semi-spherical indenter and the tissue reaction force simultaneously. It make use of with fiber optical fiber sensing method measure indentation depth and force and is small enough to fit through a standard trocar port with a diameter of 11 mm.

The created FID sensor was calibrated and tested on silicone block simulating soft tissue. The results show that the sensor can measure the indentation depth accurately and also the orientation of the sensor with respect to the tissue surface whilst performing indentation.

I. INTRODUCTION

There have been notable advances in robotic-assisted minimally invasive surgery (MIS) or laparoscopic surgery in recent years. The most well known robotic surgical systems includes the Zeus™ Surgical System from Computer Motion (Computer Motion was merged with Intuitive Surgical in 2003) and the da Vinci™ Surgical System (Intuitive Surgical, Inc.[2]). Distinct advantages of these robotic systems include high distal dexterity, enhanced 3D vision, motion scaling and tremor filtering as well as direct hand-eye coordination eliminating reversed tool motions. These advances result in improved ergonomics and also allow complex surgical procedures, which are usually difficult to be conducted by conventional minimally invasive means such as coronary artery bypass grafting [3] and mitral valve repair[4] to be effectively performed. However, the lack of haptic feedback in current robot-aided minimally invasive surgery systems is still one of the major downsides of such systems.

II. LITERATURE SURVEY

The indentation depth of an indenter is given by the relative displacement of the tip of indenter with respect to the soft tissue surface. Hence the indentation depth can be measured by acquiring the local tissue deformation when the wheel rolls over the soft tissue. So far, a number of techniques have been developed to intra-operatively acquire the tissue deformation during a MIS.

Intra-operative three-dimensional computed tomography (CT) technique can provide information regarding soft tissue morphology and structure. This technique is particularly suitable for differentiating the bones from other tissues and thus has been often used in the image-guided surgery such as oral surgery [7], orthopedic surgery [8], neurosurgery [9] as well as the prostate brachytherapy[10]. However, due to the cost and the radiation exposure associated with such techniques, they are not ideal for intraoperatively measuring the tissue deformation during a minimally invasive surgery.
A relatively inexpensive and practical technique to measure the tissue deformation is to use the stereo vision. The stereo vision utilizes two or more cameras to interpolate the three-dimensional (3D) information of a subject including the shape, texture and depth, based on computer vision techniques and the known geometry coordinates between the cameras [11]. Using a high speed camera, this technique can extract information in real time, and thus it can be used to recover the 3D structure of the operating field [12] and track the temporal motion of deformable tissue surfaces [13] during a minimally invasive surgery. While the applications of optical vision method in minimally invasive surgery are promising and progressing, one major downside with this technique is by the difficult in extracting features on curved and specular surfaces. In medical application, especially during a surgical procedure where the internal organs are all surrounded in fluids and have curved surfaces, the stereo vision method is not always effective [11].

To extract features on curved surfaces, the laser line scanner can be used. While this type of system provides high resolution 3D images, its speed is usually not real-time. Hence it is difficult to require feature information from moving objects using such a system. A more practical way to obtain the 3D information of a curved object in real time is the structured light technique.[11],[14],[15] This technique utilizes a projector to project a known light pattern onto an object. The light pattern on the object is then viewed from a camera mounted next to the projector. Given the known position and orientation of both camera and projector, the information including the shape and the distance of object can be calculated. However to use such a technique to infer the real time wheel indentation depth is problematic. First, while this technique works effectively for a curved object, it has difficulty in extracting information on specular surfaces which are often encountered in the internal body. Second, it is difficult to integrate the wheeled probe with the projector and the camera while maintaining the small size of the whole device.

In view of limitations associated with the above intra-operative 3D imaging techniques, a new technique to measure the indentation depth is proposed in this paper. Our method is to develop a fiber optic FID sensor to concurrently measure the wheel indentation depth and the tissue reaction force during the wheel-tissue interaction. The advantages of utilizing such a method are twofold: 1) the method is effective and low cost as it can accurately measure the wheel indentation depth in real-time without the need for a high speed camera system and time consuming imaging processing techniques; 2) the FID sensor can be easily miniaturized and integrated with the force-torque sensor, still small enough to fit through a standard trocar port.

III. THE DESIGN OF THE MINIATURIZED FID SENSOR

The miniaturized FID sensor consists of a fiber optical force sensor (FS) and four fiber optical displacement sensors (DS-1, DS-2, DS-3 and DS-4). Fig.1 and Fig.2 indicate the structures of the miniaturized FID sensor. The basic form of the optic sensing principle consists of a pair of transmitting fiber and receiving fiber as well as a reflector. The transmitting fiber carries the light generated from a light source and projects the light to the reflector. The reflector then reflects the light into the receiving fiber, and finally the intensity of reflected light is detected by an optical detector. Since the change of the displacement between the reflector and fiber tips \( \phi \) varies the reflected light intensity, the motion of reflector can be detected through the measuring the reflected light intensity [17]. The transmitting fiber and receiving fiber can be placed with an inclined angle \( \phi \) to increase the sensitivity. It has been demonstrated in [17] [18] that, along the increase of the bent angle \( \phi \) from 0° to 90°, the sensing sensitivity increases and displacement sensing range decreases.

![Fig.1. The miniaturized FID sensor consisting of one fiber optical fiber force sensor (FS) as well as four fiber optical fiber displacement sensors (DS), including DS-1, DS-2, DS-3 and DS-4; a coin of one pound sterling is used to indicate the size of the miniaturized FID sensor. All units are in “mm”.

![Fig.2. The structures of the miniaturized FID sensor; the main components of the device include a optic-fiber force sensor (FS), and four fiber optical fiber displacement sensors, DS-1, DS-2, DS-3 and DS-4, each consisting of a sliding rod (SR) attached to a reflector and a pair of transmitting fiber and receiving fiber. SR(DS-1), SR(DS-2), SR(DS-3) and SR(DS-4) represent the sliding rod of displacement sensor DS-1, DS-2, DS-3 and DS-4, respectively. D1, D2, D3 and D4 indicate the contactors connecting with the displacement sensors; E indicates the indenter connecting with the FS.](image-url)
reflector and a fiber pair including a transmitting fiber and a receiving fiber, as shown in Fig. 3(a). The helical-cut sensing structure behaves like a spring with high elastic modulus; when an axial force is imparted, the helical-cut sensing structure is compressed, causing a small movement of reflector towards the fiber tips. By measuring the displacement of the reflector, the applied axial force can be interpreted. The transmitting fiber and receiving fiber are used to detect the distance between the fiber tips and the reflector. The tips of both transmitting fiber and the receiving fiber are bent with an angle of 15° and placed towards each other to increase the sensitivity of the FS, Fig. 3 (a) (inclined angle $\varphi = 30^\circ$). The shield prevents the shape of helical-cut sensing structure being altered by lateral forces.

The four displacement sensors, DS-1, DS-2, DS-3 and DS-4, are arranged in an equal space inside the shield. SR(DS-1), SR(DS-2), SR(DS-3) and SR(DS-4) represent the sliding rod of displacement sensor DS-1, DS-2, DS-3 and DS-4, respectively. Four semi-spherical contactors, $D_1$, $D_2$, $D_3$ and $D_4$, connect with the sliding rods of the four displacement sensor. The shield guides each sliding rod to move freely along its axial direction. Each displacement sensor consists of a sliding rod attached with a reflector and a pair of transmitting and receiving fibers, Fig. 3 (b). The transmitting fiber and receiving fiber are used to detect the distance between the fiber tips and the reflector. The two fibers are arranged in parallel with no bent angles to increase the sensing range of the displacement sensor.

Five high-intensity LEDs are used for generating transmitting light, the change of light intensity for DSs and FS are measured using photo detectors, the received signals are amplified using the Op-Amp AD620 which has ultra-low input current noise. The optic circuit is placed inside metal box to reduce the signal noises. A 16-bit data acquisition card (NI PCI 6013) and the LabView™ 8.0 software package are used to acquire the measured signals. The sampling rate is 100 Hz.

IV. Measuring the Indentation Depth and the Sensor Orientation

During a indentation conducted using the miniaturized FID sensor, the fiber optical FS measures the forces acting on the indenter E while the four semi-spherical contactors ($D_1$, $D_2$, $D_3$ and $D_4$) slide over the tissue surrounding the indenter. When the indenter indents into the tissue surface the contactors are free to slide axially and therefore remain on the surface of the tissue. Due to soft tissue's high local deformability, the indentation contour raised from the indentation point to the surface following an exponential curve. Therefore, by measuring the distance between the deepest indentation point and the positions of the contactors, the indentation depth of the indenter and the orientation of the indenter with respect to the tissue surface can be established.

![Fig. 3. Schematics of the fiber optical force sensor (a) and the displacement sensor (b); in both sensors, a pair of transmitting fiber and receiving fiber is used to measure the distance between the fiber tip and the reflector. In the force sensor, the reflector is attached to a helical-cut sensing structure; in the displacement sensor, the reflector is attached to a sliding rod which has free movement axially.](image1)

![Fig. 4. Acquisition of the indentation depth and the orientation of the indenter with respect to the tissue surface during indentation using the miniaturized FID sensor; positions of contactors, $D_1$, $D_2$, $D_3$ and $D_4$ are (0, -$a$, -$h_1$), (0, $a$, -$h_2$), (0, $a$, -$h_3$) and (-$a$, 0, -$h_4$) respectively in the coordinate system $(x, y, z)$, where $a$ is the radius of the miniaturized FID sensor ($a = 5.5$ mm), $h_1$, $h_2$, $h_3$, and $h_4$ are the distances between the contactors and the $x$, $y$–plane; $\alpha_0$ indicates the average position of the four contactors.](image2)

It is defined the $(x, y, z)$ to be the globe Cartesian coordinate system. The coordinate system $(x_i, y_i, z_i)$ is attached to the miniaturized FID sensor as indicated in Fig. 4 ($z_i$ is align with the central axis of the sensor). When the sensor indents into the soft tissue, the four contactors can configure four planes, $D_1D_2D_3$, $D_2D_3D_4$, $D_3D_4D_1$ and $D_4D_1D_2$. The normal direction of the contacted tissue surface can be approximated by averaging the normal vectors of the planes $D_1D_2D_3$, $D_2D_3D_4$, $D_3D_4D_1$ and $D_4D_1D_2$.

As seen in Fig. 4, defining the $n_{123}$ to be the unit vector normal to the plane $D_1D_2D_3$, the $n_{123}$ can be calculated as:

$$n_{123} = \frac{D_2D_3 \times D_1D_2}{|D_2D_3 \times D_1D_2|}$$
Defining the positions of contactors, D1, D2, D3 and D4 to be 
(0, -a, -h1), (a, 0, -h2), (0, a, -h3) and (-a, 0, -h3) respectively
in the coordinate system (x, y, z), where a is the radius of the
miniaturized FID sensor (a = 5.5 mm), h1, h2, h3 and h4 are the
distances between the corresponding contactors and the x, y, z
plane, the vector D2D1 and D3D4 can be calculated as:

\[
\begin{align*}
D_2D_1 &= oD_1 - oD_2 = -ai + aj + (h_2-h_3)k, \\
D_3D_4 &= oD_3 - oD_4 = -ai - aj + (h_2-h_1)k.
\end{align*}
\]

Thus

\[
D_2D_3 \times D_1D_4 = \begin{vmatrix} i & j & k \\ -a & a & h_2 - h_3 \\ -a & -a & h_2 - h_1 \end{vmatrix} = a(2h_2-h_1-h_3)i - a(h_1-h_3)j - 2a^2k,
\]

where i, j and k are the unit vectors along x, y, z axis, respectively. Hence

\[
n_{123} = \frac{a(2h_2-h_1-h_3)}{r_{123}}i + \frac{a(h_3-h_1)}{r_{123}}j + \frac{2a^2}{r_{123}}k,
\]

where \(r_{123}\) is the norm of the vector \(n_{123}\).

\[
r_{123} = \sqrt{a^2(2h_2-h_1-h_3)^2 + a^2(h_3-h_1)^2 + 4a^4}.
\]

Following the above procedures, the unit vector \(n_{124}\) which
is normal to plane \(D_1D_2D_4\) can be calculated as:

\[
n_{124} = \frac{a(h_2-h_3)}{r_{124}}i + \frac{a(h_2+h_4-2h_1)}{r_{124}}j + \frac{2a^2}{r_{124}}k,
\]

where \(r_{124}\) is the norm of the vector \(n_{124}\),

\[
r_{124} = \sqrt{a^2(h_2-h_4)^2 + a^2(h_2+h_4-2h_1)^2 + 4a^4};
\]

the unit vector \(n_{134}\) which is normal to plane \(D_1D_3D_4\) is

\[
n_{134} = \frac{a(h_1+h_3-2h_4)}{r_{134}}i + \frac{a(h_3-h_1)}{r_{134}}j + \frac{2a^2}{r_{134}}k,
\]

where \(r_{134}\) is the norm of the vector \(n_{134}\),

\[
r_{134} = \sqrt{a^2(h_1+h_3-2h_4)^2 + a^2(h_3-h_1)^2 + 4a^4};
\]

the unit vector \(n_{234}\) which is normal to plane \(D_2D_3D_4\) is

\[
n_{234} = \frac{a(h_2-h_4)}{r_{234}}i + \frac{a(2h_3-h_2-h_4)}{r_{234}}j + \frac{2a^2}{r_{234}}k,
\]

where \(r_{234}\) is the norm of the vector \(n_{234}\),

\[
r_{234} = \sqrt{a^2(h_2-h_4)^2 + a^2(2h_3-h_2-h_4)^2 + 4a^4}.
\]

The normal vector to the contacted tissue surface, \(n_r\), can be
approximated by the resultant vector of \(n_{123}, n_{124}, n_{134}\), and
\(n_{234}\),

\[
n_r = n_{123} + n_{124} + n_{134} + n_{234}.
\]

The angle \(\theta_e\) between the normal vector of the tissue surface
and the z axis (central axis of the sensor) can be determined as:

\[
\cos \theta_e = 2a^2 \left( \frac{1}{r_{123}} + \frac{1}{r_{124}} + \frac{1}{r_{134}} + \frac{1}{r_{234}} \right),
\]

and the distance between the deepest indentation point and
the positions of the contactors along the normal direction of
contact, \(d_m\) is determined as:

\[
d_m = \cos \theta_e \left( \frac{h_e - h_1 + h_2 + h_3 + h_4}{4} \right),
\]

where \(r_{123}, r_{124}, r_{134}\) and \(r_{234}\) are the vector norms of \(n_{123},
\(n_{124}, n_{134}\), and \(n_{234}\), respectively. \(h_e\) is the distance between the
tip of the indenter E and the \(x, y, z\)-plane.

Since the helical-cut sensing structure of the fiber optical
fiber FS has high elastic modulus, the change of distance \(h_e\)
during indentation is reasonable to be neglected. Hence
distance \(h_e\) can be considered as a constant value. Moreover,
the distance \(h_1, h_2, h_3, h_4\) can be measured using the
corresponding displacement sensors, DS-1, DS-2, DS-3 and
DS-4. Thus the \(d_m\) and orientation of the indenter with respect
to soft tissue surface can be detected using the miniaturized
FID sensor. To obtain the indentation depth, \(d_m\), the
relationship between the \(d_m\) and \(d_m\) need to be calibrated on
different tissue sample. The calibration can be then used to
interpret the indentation depth \(d_m\) from measuring the \(d_m\).

VI. CALIBRATION OF THE FID SENSOR

To calibrate the four displacement sensors and the force
sensor of the FID sensor, the entire unit is attached to the
distal tip of a Mitsubishi RV-6SL, a 6-DoF (degree of
freedom) robotic manipulator, to allow for accurate motion
control. Fig.5 shows the calibration setup using the FID
sensor.

For the force sensor calibration, the fiber optic FS is
compressed against an ATI Mini40 sensor. During the
calibration, the fiber optic FS was slowly advanced towards
and retrieved from the Mini40, varying the interaction force
between the two sensors. The signals measured from the fiber
optic FS and the corresponding force values measured from
the Mini40 were recorded. The calibration results are shown
in Fig.6. For the calibration of the four DSSs, each sliding rod of
DS was moved individually from its minimum length to the
maximum length over a range of 16 mm. The displacement
\(h_1, h_2, h_3, h_4\) and the corresponding signal from each
sliding rod were recorded respectively. The calibration results
are shown in Fig.7.
To calibrate the relationship between the indentation depth $d_{in}$ and the distance $d_m$, a silicone block, with the dimension of 30 mm in height, 50 mm in width and 100 mm in length, was used. During the calibration, the FID sensor was vertically indented into the silicone sample at a speed of 0.1 mm/s until $d_{in}$ reached 6 mm. The value of $d_m$ and the corresponding $d_{in}$ were recorded. The calibration results are shown in Fig.8. From the calibration, it was found that the relationship between the indentation depth $d_{in}$ and the distance $d_m$ can be expressed as

$$d_{in} = 0.44(e^{0.66d_m} - 1).$$  \hspace{1cm} (3)

VI. PERFORMANCE ANALYSIS

In order to analysis the performance of the FID sensor, the robotic manipulator was used to drive the sensor to slide across the silicone sample along y axis by a distance of 40 mm at different angle $\theta_z$ (the angle between the normal vector of the tissue surface and the central axis of the sensor). There were four angle selected, $\theta_z = 0^\circ$, $5^\circ$, $10^\circ$ and $15^\circ$. The indentation depth was linearly increased from from 0 mm to 6 mm while the angle $\theta_z$ was kept constant at $0^\circ$. Indentation depth increased linearly from 0 mm to 5 mm while the angle $\theta_z = 5^\circ$, $10^\circ$ and $15^\circ$. During the experiments, the distance $d_m$ was calculated using Eq.1, Eq.2 and then the indentation depth, $d_{in}$ was determined from the $d_m$ using Eq.3. Fig.9 shows the comparison between the measured indentation depth and ground truth data.
measuring mechanism outward in order to guarantee contact between the contactor and the tissue. Future work will aim at studying other mechanisms in order to ensure that indentation depth can be measured independently of the direction of gravity, as it is needed if the tool is to be moved arbitrarily with regards to lines of gravity.

VII. CONCLUSION

This paper describes a miniaturized FID sensor which fits through a standard trocar port with an inner diameter of 11 mm. The validation experiment conducted on a silicone sample demonstrates that the developed miniaturized FID sensor can accurately and concurrently measure both the indentation depth of the indenter and the tissue reaction force during tool-tissue interaction. Since the stiffness can be acquired using the tissue deformation and the corresponding tissue reaction force, it is expected that the stiffness distribution of the investigated soft tissue can be identified from by sliding the FID sensor over a large area of tissue without prior tissue surface registration. This indicates the feasibility of using the miniaturized FID sensor for tissue abnormalities identification during minimally invasive surgery. The proposed FID sensor has potential to aid surgeon in locating tumors during minimally invasive surgery.

It is noted that, although using a semi-spherical indenter instead of a rolling wheel simplified the manufacturing of the, the spherical indenter may cause large tangential (shear) force while sliding across tissue, potentially damaging the underlying tissue. Furthermore, the interaction dynamics between the fixed spherical indenter and soft tissue has not been investigated in this study. Thus future experimental study of interaction dynamics would be required for implementing the spherical indenter.

It is also noted that at this stage of the research into indentation depth measuring, gravity is used to extend the

REFERENCES