Force Sensor Integrated Surgical Forceps for Minimally Invasive Robotic Surgery

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Abstract—This paper presents a novel surgical instrument integrated with a four-degree-of-freedom (DOF) force sensor. By adopting the capacitive transduction principle, the sensor enables the direct sensing of normal and shear forces at surgical instrument tips. Thus, three-DOF pulling forces and a single-DOF grasping force can be measured for haptic feedback control of robotic minimally invasive surgery. The sensor consists of four capacitive transducers, and all the transducers including analog signal processing units are embedded in small surgical instrument tips. The four-DOF force sensing is enabled thanks to the four capacitive transducers by using the force transformation method. In this study, the instrument is designed and manufactured to be adaptable to the open-source surgical robot platform, called Raven-II. In addition, the sensing system is experimentally validated through its application to the Raven-II by using a reference force sensor.

Index Terms—Capacitive sensor, force sensor, haptic feedback, minimally invasive surgery (MIS).

I. INTRODUCTION

T

he application of robotic systems to minimally invasive surgery (MIS) has introduced a number of advantages in terms of postoperative pain, recovery time, and overall costs of medical treatment [1]–[4]. Although minimally invasive robotic surgery (MIS) is advantageous for patients, it also involves some limitations. MIS mainly relies on a surgeon’s skill to operate a master device to manipulate the surgical robot [5]–[7], and is there always the possibility of surgeon’s damage to the tissue due to excessive grasping force or possible slips during the operations [8]–[10]. To overcome this problem, the force-sensing system designed for MIS has been studied by many research groups [11]–[18]. However, the difficulty in integrating the sensing system still hampers its introduction to surgical instruments in practical applications.

Fig. 1 shows some of the locations on an instrument, which are deemed to bear possibility at the onset of our experiment, which are the gripper, articulated joint, instrument shaft, and joint actuation unit. However, all but the gripper are not an appropriate candidate due to the following limitations. The sensor at the shaft and joint actuation unit may provide inaccurate force information due to variables such as inertia, friction, backlash, and gravity [9]. The articulated joint is also not an appropriate location because there are numerous mechanical disturbances such as driving mechanisms, friction, and backlash. On the contrary, force sensors placed at the tip of the instrument, which are collocated ones, can measure accurate force information without the aforementioned problems. The sensor can read multidimensional contact forces accurately because there is direct contact with tissue; thus, we can prevent excessive grasping forces and slipping during operations.

Recently, a number of studies have been performed to integrate force sensors at the tips of surgical instruments. King et al. [19] designed a piezoresistive tactile sensor integrated into the surgical instrument tip. Although this sensor can measure direct contact forces, it does not provide the essential functionality of a surgical gripper. Qasaimeh et al. [20] developed a PVDF-based microfabricated tactile sensor mounted on the
surgical grasper with preserved grasper shape. However, the approach provided only grasping force information, which is insufficient for deducing contact force information. Hong and Jo [21] fabricated a compliant forceps using two flexure hinges with the capability to sense two axial forces. Sensorized forceps can measure single-axis pulling and grasping forces. However, measurements of single-axis pulling force do not provide sufficient information to allow force feedback control for surgical robotic systems. Lee et al. [22] developed a multiaxis contact force sensor integrated into the surgical forceps with multiaxis force-sensing capability. However, it is difficult to measure accurate contact force information due to nonlinearity and complicated calibration of the sensor.

In our previous research [1], the concept of sensorized forceps consisting of two grasper-integrated force sensors was presented. On each side of the grasper, capacitive transducers with triangular prism structure are attached. From the readings of the transducers, the tissue handling forces, which are a three-degree-of-freedom (DOF) pulling force and a single-DOF grasping force, are extracted in real time. The sensing system is simple in its structure and, thus, appropriate for disposable instruments with low cost. Although the paper deals with the conceptual design of the sensorized surgical forceps, however, thorough analyses of the sensing principle and the force transformation method are not addressed. In addition, since the forceps were not integrated with an actual surgical robot, the performance verification had not been performed.

This study extends the previous research including the complete evaluation of the sensorized forceps and further performance analysis of the proposed sensing system. The forceps is validated via experiments at the instrument level. In addition, the instrument is designed to be adaptable to the open-source surgical robot platform, called Raven-II, for verification in the actual robotic system. In the actual surgical robot operating environment, the manufactured prototype instrument is evaluated through experiments using a reference force sensor. The force transformation method also is completely verified by reflecting the change of the included angle between two graspers during the operation. In addition, additional experiments about grasping force sensing are conducted for verifying the sensing’s feasibility. And, data analyses of the sensor are conducted to estimate the sensor’s performances such as force range, resolution, RMS error, repeatability, and hysteresis through additional experiments.

This paper is organized as follows: The development of sensorized forceps is introduced in Section II. Verification of sensorized forceps is explained in Section III. In Section IV, verification of the sensorized integrated system is presented in an experimental setup. Finally, discussion and the conclusion are given in Section V.

II. DEVELOPMENT OF THE SENSORIZED INSTRUMENT

A. Overview of Sensing System

Measuring contact and shear forces at surgical instrument tips is important to improve surgeon’s skill for preventing failures in robotic surgery [1], [23], [24]. Thus, the surgical graspers should be able to measure contact and shear forces with enough sensitivity to detect changes of force in handling soft tissue. And, the sensor’s force range should be capable of the maximum force applied during various surgical tasks using a surgical robot. In [5], the maximum stress applied to tissues such as liver was measured as about 200 kPa before those failures. The stress can be calculated as 5 N, assuming that the grasper’s surface is 25 mm². Forces applied to sutures using a surgical robot were detected to be less than 4 N [25]. To satisfy these requirements, a sensorized grasper composed of two grasper-integrated force sensors is designed as shown in Fig. 2. By specifically arranging capacitive-type sensor cells, contact and shear forces can be measured on the tips of the sensorized grasper while preserving the surgical grasper’s shape. Capacitive-type sensors consisting of a dielectric elastomer and two electrodes allow reduction in size, enhanced sensitivity, and proper force range [9], [26], [27]. The grasper provides information regarding all directional contact and shear forces applied to the tip. The force elements, i.e., two normal forces and two shear forces, can be transformed into a three-DOF Cartesian manipulating force and one-DOF grasping force by their geometric relation to the triangular prism.

B. Sensing Principles of Capacitive Transducer Cell

In general, the capacitance between two parallel conductors is obtained by

$$C = \varepsilon_0 \varepsilon_r \frac{A}{t}$$

where $C$ is the capacitance between the upper and lower electrodes, $\varepsilon_0$ represents the dielectric constant of air, and $\varepsilon_r$ is the relative permittivity of the dielectric elastomer. $A$ is the overlapping area, and $t$ is the distance between the two electrodes or the thickness of the polymer layer. Thus, it is noted that the capacitance depends on the distance between the upper and lower electrode layers and the overlapping area of the electrode layers.

As depicted in Fig. 3, the proposed sensor is composed of a dielectric elastomer substrate and two electrode layers. The sensor can measure two directional forces as changes in the...
capacitance change between the electrodes occurs. Fig. 3(a) illustrates the movement of the proposed sensor when a vertical force is applied to the sensor surface. Vertical displacement causes changes in the dielectric elastomer’s thickness \( t \), which alters the capacitance value. Since the applied force is directly related to displacement, changes in the capacitance enable the applied force to be computed. When the horizontal force is applied to the sensor surface, changes in the overlapping area between two electrodes occur along with a change in capacitance as presented in Fig. 3(b). In this case, the change of capacitance is very small, and accurate measurement cannot be guaranteed [22]. In this study, we have removed the influence of horizontal force by designing the lower electrode smaller than the upper electrode as shown in Fig. 3(c). Thus, any influence of capacitance changes caused by changes in the overlapping area is eliminated, and only pure vertical force sensing can be observed.

According to Hooke’s law, the relationship between the thickness and the applied force is estimated to be in line with each other; if the change of thickness is small, change in capacitance is also related to that of thickness. Thus, the relationship between capacitance and applied force with the thickness \( t \) is represented as

\[
C = \varepsilon_0 \varepsilon_r \frac{A}{t(1 - \frac{F}{E})}
\]

where \( C \) is the capacitance, and \( F \) is the force applied on the sensor’s surface. \( A' \) is the sensor area exposed to an external force, and \( E \) is Young’s modulus of the dielectric elastomer substrate. Fig. 4 illustrates the sensing principles of a grasper-integrated sensor. The sensor is composed of two capacitive-type sensor cells, a triangular prism, and a bottom structure. The two cells are adhered to the gap between the triangular prism and the bottom structure as shown in Fig. 4(a). The positive electrode is placed at the bottom surface, and the triangular prism structure is used as a common ground. As shown in Fig. 4(b), a normal force applied on the sensor’s surface deforms the two cells with symmetric capacitance responses. On the contrary, a shear force deforms the two cells with asymmetric capacitance changes as shown in Fig. 4(c), and thus, normal and shear force can be discerned. The combination of the force information measured by two capacitive-type sensor cells allows the derivation of the normal and shear forces applied to the surface of the grasper including the magnitude and direction of the external force.

C. Force Transformation of Sensorized Forceps

As mentioned above, the four capacitance values measured by the sensorized forceps give normal and shear forces at the contact point and they need to be transformed into a three-DOF manipulating force and one-DOF grasping force. The three-DOF manipulating force indicates the 3-D Cartesian force, which can be detected while handling tissues. The one-DOF grasping force represents the squeezing force which grasps the tissue. As displayed in Fig. 5, the upper grasper-integrated force sensor measures normal and shear forces, and the lower force sensor measures normal and shear forces in a direction orthogonal to the upper ones. The geometry of the triangular prism structure
is described by a transformation matrix that relates force data from capacitive sensor cells to two normal forces and two shear forces as

$$[F_{\text{nor}}, F'_{\text{nor}}, F_{\text{she}}, F'_{\text{she}}]^T = T_{\text{NS}} \cdot [F_{\text{cell1}}, F_{\text{cell2}}, F_{\text{cell3}}, F_{\text{cell4}}]^T$$

(3)

where $[F_{\text{nor}}, F'_{\text{nor}}, F_{\text{she}}, F'_{\text{she}}]^T$ represents the vector of each normal and shear force, and $[F_{\text{cell1}}, F_{\text{cell2}}, F_{\text{cell3}}, F_{\text{cell4}}]^T$ denotes the vector of four output forces measured by four cells as shown in Fig. 5. $T_{\text{NS}} \in \mathbb{R}^{4 \times 4}$ transforms the four output forces into two normal and two shear forces. Based on the geometric configuration of the triangular prism, $T_{\text{NS}}$ is obtained by

$$T_{\text{NS}} = 
\begin{bmatrix}
\cos \theta & -\cos \theta & 0 & 0 \\
\sin \theta & \sin \theta & 0 & 0 \\
0 & 0 & \sin \theta & \sin \theta \\
0 & 0 & \cos \theta & -\cos \theta 
\end{bmatrix}
$$

(4)

where $\theta$ is the angle of the four output forces on the coordinate based on normal and shear force vectors. Since the cross section of the triangular prism is a right-angled isosceles triangle, $\theta$ is 45°. With Cartesian coordinates, the direction vectors of the normal and shear forces are determined using the included angle ($\alpha$) between two graspers. These values are transformed into three-DOF pulling and one-DOF grasping forces such as

$$[F_{\text{px}}, F_{\text{py}}, F_{\text{pz}}, F_{\text{G}}]^T = T_{\text{PG}} \cdot [F_{\text{nor}}, F'_{\text{nor}}, F_{\text{she}}, F'_{\text{she}}]^T$$

(5)

where $[F_{\text{px}}, F_{\text{py}}, F_{\text{pz}}, F_{\text{G}}]^T$ indicates the vector of pulling and grasping forces. $T_{\text{PG}} \in \mathbb{R}^{4 \times 4}$ transforms the normal and shear forces into manipulating and grasping forces. $T_{\text{PG}}$ is formulated as

$$T_{\text{PG}} = 
\begin{bmatrix}
\cos \alpha & -\sin \alpha & -\sin \alpha & 0 \\
\sin \alpha & \cos \alpha & -\cos \alpha & 0 \\
0 & 0 & 0 & 1 \\
0 & 0 & 0.5 & 0.5
\end{bmatrix}
$$

(6)

where grasping force is defined as the average of two normal forces at the sensorized grasper, and pulling force is defined as the three-DOF pulling force with Cartesian coordinates. $\alpha$ is the angle between the graspers. Depending on $\alpha$, the manipulating and grasping forces are determined by the two normal and two shear forces of the graspers. The calculated transformation matrices can be used to calibrate the force values of the four sensing cells.

D. Fabrication of Grasper-Integrated Force Sensor

As shown in Fig. 6(a), the grasper-integrated force sensor consists of a triangular prism structure, a flexible printed circuit board (FPCB), an insulator layer, a base structure, and a chip case. Two electrodes are printed, and a capacitance-digit-converter (CDC) chip is soldered on the FPCB. The FPCB is placed on the V-shaped insulator, which isolates capacitance from the base structure. The triangular prism, base structure, and chip case are connected to the ground to minimize outer electromagnetic interference, since the electrodes and the CDC chip need to be shielded due to the capacitive-type sensor’s sensitivity to changes in outer electrical energy (noise) [30], [31].
Fig. 7. Three-dimensional model of the developed sensorized surgical instrument. (a) Four-DOF articulated joint in sensorized forceps. (b) Joint actuation unit equipped with four driving pulleys. (c) Developed sensorized surgical instrument.

Fig. 8. Pictures of the developed sensorized forceps. (a) Two semiassembled grasper-integrated force sensors. (b) Fully assembled sensorized forceps. (c) Developed sensorized surgical instrument.

The sensorized instrument is also tested via integration into the Raven-II. The Raven II has been proven as a reliable robotic surgery platform through various experiments [29].

The proposed sensor is fabricated as presented in Fig. 8. Fig. 8(a) shows a picture of the semiassembled grasper-integrated force sensor. The triangular prism, base structure, and chip case are made of aluminum alloy that possesses good electrical conductivity for passive shielding to minimize outer electromagnetic interference with capacitance. The AD7147A (2.1 × 2.3 mm$^2$) produced by Analog Devices was selected as a small CDC chip. The AD7147 chip can measure capacitance with 650-Hz speed and 16-bit resolution. Using the I$^2$C communication interface, the digitized capacitance is sent to the outer signal processing device without any additional device. The FPCB is 4 mm wide with four lines (two lines for I$^2$C communication and two sensor power lines) that are connected to the MCU circuit through the hole in the instrument’s shaft. It is equipped with the CDC chip and two electrodes (2 × 2 mm$^2$). The FPCB is adhered to a V-shaped insulator composed of polyether ether ketone, which was selected due to its high mechanical strength and electrical insulation.

A fully assembled sensorized grasper is shown in Fig. 8(b). The prototype of the sensorized instrument has wrist, rolling, and grasping joints to allow four axis motions and was manufactured with the proposed sensor, as presented in Fig. 8(c). Table I shows that the sensorized instrument’s Denavit–Hartenberg (D–H) parameters are the same as those of the Raven-II instrument [32]. The shared configuration provides compatibility between the developed instrument and the Raven-II surgical robotic platform.

### III. VERIFICATION OF THE SENSORIZED INSTRUMENT

#### A. Experimental Setup

As presented in Fig. 9, the experimental setup was built to test the developed sensor system. In Fig. 9(a), the setup consists of the sensorized instrument, a motorized stage assembled with a push–pull gauge (RX-2, AIKOH) to measure applied force, a manual stage for arbitrary input of external force, and jigs to fix the instrument in place. Force data measured by a push–pull gauge were used to calibrate the capacitive data measured by the sensor, inputting external forces by using the stages as shown in the right inset in Fig 9(a).

Fig. 9(b) shows the experimental setup used to verify the sensor performance of the sensorized instrument, a force/torque (F/T) sensor (Nano17, ATI), a simulated tissue, and a motorized stage that can move automatically.

For communication between the computer and sensor, a microcontroller unit (MCU) board was designed and included in the sensorized instrument. In the board, a chip (ARM cortex M3, 70 MHz) was selected for the MCU in order to provide sensor information in real time. The board reads capacitance data digitized by the CDC chip in the developed sensor and sends the data to the computer by controller area network communication.

One side of the tissue was fixed to the reference sensor assembled on the stage. The tissue, a sponge strip (5 × 3 × 40 mm$^3$), was selected because its mechanical property (Young’s modulus: 160 kPa) was similar to the liver [5]. On the surface of the reference sensor, a jig was assembled, consisting of two bolts of which tips were covered with a plastic cover as demonstrated in the right inset of Fig 9(b). Using the jig, the tissue was fixed to the reference sensor. And, the sensorized forceps grasped the other side of the tissue by rotating and locking the pulley in the joint actuation unit of the instrument by the jig under the unit. The inset in Fig. 9(b) shows the difference between the x, y, and z coordinates of two sensors. For the comparison with data...

<table>
<thead>
<tr>
<th>$i$</th>
<th>$a_{i-1}$</th>
<th>$d_i$</th>
<th>$\theta_i$</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>0</td>
<td>0</td>
<td>458.7 mm</td>
</tr>
<tr>
<td>5</td>
<td>π/2</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>6</td>
<td>π/2</td>
<td>11 mm</td>
<td>0</td>
</tr>
</tbody>
</table>
Fig. 9. Experimental setting for calibration and evaluation of the developed sensorized forceps. (a) Experimental setting for calibration of each grasper-integrated force sensor. (b) Experimental setting for evaluation of the sensorized forceps.

of the sensors, the coordinates were matched by multiplying a rotation matrix. When the stage moved, equal force was applied to the forceps and the reference sensor. The linear stage was controlled by a Lab-View program.

B. Calibration

Using the geometric relation of the sensor structure, all four forces were transformed into normal force, shear force, three-axial pulling forces, and a single-axial grasping force. In other words, the normal, shear, pulling, and grasping forces can be calculated using the aforementioned matrixes [see (3)–(6)] with simple calibration of the four sensor cell data. As shown in Fig. 9(a), experiments were conducted to calibrate data measured by the four capacitive-type sensor cells. To create a calibration matrix, the grasper-integrated force sensor was mounted on the jig. Using a motorized stage with a push–pull gauge, the sensor was then loaded to 3 N with the resolution (0.01 N) of the push–pull gauge. At this moment, the push–pull gauge measures the normal force \( F_{\text{nor}} \) applied on the grasper's surface, and force at the sensor's cell is treated as \( F_{\text{nor}}/2\sin\theta \) according to the transformation matrixes mentioned above. To check the tendency of capacitance change by the external force, data at intervals of 0.5 N are selected, which are linearly fit as shown in Fig. 10. Each point is determined as the average of ten data measured repeatedly. The distribution of the ten data is expressed by using error bar. Fig. 11(a) represents capacitance data measured by a sensing cell at 0.2 Hz repeatedly. The hysteresis was observed by increasing and decreasing the applied force three times repeatedly as shown in Fig. 11(b). The same experiment was conducted for the other grasper-integrated force sensor. Fig. 10 shows the linear characteristics of the loading forces and sensor output data. Using this linear characteristic, the relation between loading forces and sensor data can be expressed as [33]

\[
\begin{bmatrix} F_{\text{cell}1} \\ F_{\text{cell}2} \\ F_{\text{cell}3} \\ F_{\text{cell}4} \end{bmatrix} = A \cdot \begin{bmatrix} C_{\text{cell}1} \\ C_{\text{cell}2} \\ C_{\text{cell}3} \\ C_{\text{cell}4} \end{bmatrix}
\]

where \( \begin{bmatrix} F_{\text{cell}1} \\ F_{\text{cell}2} \\ F_{\text{cell}3} \\ F_{\text{cell}4} \end{bmatrix} \) indicates the force vector, and \( \begin{bmatrix} C_{\text{cell}1} \\ C_{\text{cell}2} \\ C_{\text{cell}3} \\ C_{\text{cell}4} \end{bmatrix} \) is the vector of the sensor's capacitance data at each capacitive-type sensor cell. \( A \) is the calibration matrix that determines the relation between capacitance measured by the sensors and the four output forces. Because this relation is linear with one-to-one correspondence, the matrix is a diagonal matrix, and the diagonal elements were calculated as 0.02697, 0.01901, 0.01839, and 0.03763, depending on Fig. 10. Thus, two normal and shear forces, three-axial pulling force, and a single-axial grasping force were calculated as

\[
\begin{bmatrix} F_{\text{she}} \\ F_{\text{nor}} \\ F'_{\text{nor}} \\ F'_{\text{she}} \end{bmatrix} = T_{\text{NS}} \cdot A \cdot \begin{bmatrix} C_{\text{cell}1} \\ C_{\text{cell}2} \\ C_{\text{cell}3} \\ C_{\text{cell}4} \end{bmatrix}
\]
\[
\begin{bmatrix}
F_{P_x} & F_{P_y} & F_{P_z} & F_G
\end{bmatrix}^T = 
T_{PG} \cdot T_{NS} \cdot A \cdot [C_{cell1} \ C_{cell2} \ C_{cell3} \ C_{cell4}]^T. 
\tag{9}
\]

C. Testing With a Sensorized Surgical Instrument

For the performance verification, pulling and grasping forces measured by the proposed sensorized instrument were compared with those measured by a reference sensor in Cartesian coordinates using the experimental setup as shown in Fig. 9(b). The measured forces were estimated based on the measured cell forces and the transformations in (8) and (9). Data acquisition of two sensors was conducted with the same sampling rate and under a common clock. The four capacitance values of the four sensing cells were simultaneously read, while the forceps grasped the tissue fixed to the reference part. The data were calibrated using the method mentioned above. Fig. 12(a) shows the force data obtained by grasping and pulling with the forceps. The first increase in data values indicates that the tissue was grasped by the forceps. The motorized stage assembled to the reference sensor automatically pulled and released the tissue fixed between the forceps and the reference sensor. In this experiment, the four pulling motions were carried out in 48 s, with a 12-s cycle. Fluctuations in force data are explained as repeated pull and release of tissue.

Data transformed by the transformation matrix in (8) are presented in Fig. 12(b). The transformation process was conducted in real time. Fig. 12(b) shows the normal and shear forces calculated by the transformation matrix. In these graphs, the upper two data points indicate the shear and normal forces measured by the upper grasper. The other points are the shear and normal forces measured by the lower grasper. During the grasping and pulling motions, the two normal forces \( (F_{nor}, F'_{nor}) \) increased by about 4 N each, and the two shear forces \( (F_{she}, F'_{she}) \) changed by about 0.72 and \(-0.15\) N, respectively. Since the pulling motion was inclined toward the \( x \)-axis, the measured \( F_{she} \) was larger than the \( F'_{she} \). The pulling and grasping forces of the developed sensor and reference sensor responses are presented in Fig. 12(c). These forces were calculated using the transformation matrix in (9). The pulling force elements \( (F_{P_x}, F_{P_y}, F_{P_z}) \) were measured up to 0.72, \(-0.15\), and \(-0.33\) N, respectively. The graphs indicate similarities between responses of the developed sensor and reference in time-domain responses. Through the experiments, the RMS errors were calculated as 0.08, 0.07, and 0.11 N. The grasping force was measured up to about 4 N. Because there are no reference data comparing grasping forces in this experiment, additional experiment was conducted. Grasping force \( (F_G) \) is the average of two normal forces, and the two normal forces mean grasping forces at upper and lower graspers, respectively. Therefore, two normal forces were compared in the experimental setup as presented in Fig. 9(a). The grasper placed on a manual stage contacts the probe of the push–pull gauge with no load. And, in this stage, normal force was applied arbitrarily, with the measurement of the instrument and the push–pull gauge as references during 60 s. Fig. 13 shows similarities between responses of the developed sensor and reference. Through the experiments, the RMS errors of the pulling and grasping forces were calculated as 0.0837, 0.0732, 0.114, and 0.0957 N. The sensor’s force ranges are determined as \( \pm 2.5, \pm 5, \pm 2.5, \) and 5 N, respectively. It is possible to detect the force applied to tissue up to the safety threshold of 200 kPa for cell apoptosis in abdominal organs [5]. The repeatable errors for four force values were found to be 1.23%, 1.58%, 1.34%, and 1.56% of the entire force range.
Fig. 13. Time-domain responses of two normal forces measured by two grasper-integrated force sensors and a push–pull gauge (red).

Table II
SPECIFICATIONS OF THE PROPOSED SENSORIZED INSTRUMENT

<table>
<thead>
<tr>
<th></th>
<th>$F_{Px}$</th>
<th>$F_{Py}$</th>
<th>$F_{Pz}$</th>
<th>$F_G$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force range</td>
<td>±2.5 N</td>
<td>±5 N</td>
<td>±2.5 N</td>
<td>5 N</td>
</tr>
<tr>
<td>Resolution</td>
<td>42 mN</td>
<td>58 mN</td>
<td>72 mN</td>
<td>46 mN</td>
</tr>
<tr>
<td>RMS Error</td>
<td>0.0837 N</td>
<td>0.0732 N</td>
<td>0.114 N</td>
<td>0.0957</td>
</tr>
<tr>
<td>Repeatability</td>
<td>1.23%</td>
<td>1.58%</td>
<td>1.34%</td>
<td>1.56%</td>
</tr>
<tr>
<td>Hysteresis</td>
<td>1.96%</td>
<td>2.16%</td>
<td>2.03%</td>
<td>1.75%</td>
</tr>
</tbody>
</table>

after performing the experiments four times repeatedly. And, hysteresis errors were found to be 1.96%, 2.16%, 2.03%, and 1.75% of the force range [34]. The results are summarized in Table II. The transformation occurred in real time (650 Hz). The change of $\alpha$ affects force data of $F_{Px}$ and $F_{Py}$ as presented in Fig 5. However, $\alpha$ was set as a constant value on account of the angle between the graspers during this experiment.

IV. VERIFICATION OF THE INTEGRATED SYSTEM

A. Experimental Setup With the Raven-II Surgical Robot System

The sensorized instrument was integrated with the open-source surgical robot platform Raven-II to verify the entire surgical robotic system, as shown in Fig. 14. Because the instrument was designed with the same mechanical parameters and transmission ratio as the Raven-II, there were no control issues [32]. The Raven-II uses the ROS interface [35], and thus, an additional ROS interface node was designed to communicate sensing data measured from the instrument. In this process, force data calibration and transformation were conducted by determining the included angle ($\alpha$). The ROS interface provided the bidirectional communication between the Raven-II and the developed instrument, enabling force feedback control using a haptic device as the master. In the experiment, position control of the robot was conducted by using a remote master (Phantom Omni, Sensable Co.). In this setup, for the verification of the developed sensor, the reference sensor was fixed on the ground with the jig mentioned above. During experiments, information of $\alpha$ and the robot’s coordinate from its controller were reflected to match and compare the data measured by two sensors.

B. Experiments With the Raven-II Surgical Robot System

Experiments were conducted to verify the developed surgical instrument in a simulated surgical environment. The verification was performed by comparing force data measured by the sensorized forceps and the reference sensor. In addition, the proposed force transformation method was completely proved by reflecting the change of the angle ($\alpha$) as a variable to the matrix. As shown in Fig. 14, the forceps of the integrated instrument grasped and pulled the simulated tissue secured to the jig assembled to the reference sensor. As mentioned above, the forceps measured pulling force in Cartesian coordinates from the forceps position. As is shown, the forceps measured pulling force in Cartesian coordinates from the forceps position. As is shown, the coordinates of the forceps and reference sensor points differ. The forceps coordinates were rotated to match the reference sensor coordinates using joint position information from the Raven-II manipulator. During the
During force measurements with the sensorized forceps and reference sensor, the robot delicately manipulated the tissue as in the actual robotic surgery situation. Tissue grasped by the forceps was pulled repeatedly in random directions for 2 min. The sensing data sampling rate was 650 Hz in the experiment and the $\alpha$ was read simultaneously as shown in Fig. 16. Fig. 17(a) shows the measured normal and shear forces. The upper grasper measured the upper normal and shear forces, while the lower grasper measured the lower forces. The maximum normal and shear forces applied were 2.65, 2.1.55, and 0.15 N, respectively. Fig. 17(b) represents the pulling and grasping forces measured by the forceps and reference sensor. The $\alpha$ in matrix [see (9)] used to calculate pulling and grasping forces was reflected as the grasper joint angle of the robot. In addition, according to the data shown in Figs. 16 and 17, it is noted that larger grasping force is needed when the forceps pull tissue. The larger force changes the angle $\alpha$ by deforming the tissue. The applied maximum pulling force was measured up to 1.45, 0.21, and 0.15 N in Cartesian coordinates. The maximum grasping force was also measured as 2.1 N. The pulling force of the sensorized forceps matched that of the reference sensor data, and the RMS errors were 0.14, 0.07, and 0.05 N. In the experimental setup using the Raven surgical robot, the grasping force cannot be compared because we do not have reference data. Instead, we verified the grasping force in the instrument level. Because the elements used to calculate a grasping force are related to the pulling force, verification of the pulling force can be used to validate the measured grasping force.

V. DISCUSSIONS AND CONCLUSION

In this paper, a novel surgical instrument integrated with a four-DOF force sensor was proposed. The instrument could measure three-DOF pulling forces and single-DOF grasping force directly at the tips. For this purpose, two grasper-integrated capacitive force transducers were embedded in the forceps of the instrument. The compact design of the sensor was realized with a triangular prism structure and two capacitance sensing cells for each tip while preserving the grasper’s shape. The readings of each cell were transformed into two normal and two shear forces. Three-DOF pulling forces and single grasping forces were computed by using the transformation matrix based on the sensor’s geometry relationship.

The instrument was tested focusing on the factors such as resolution, RMS error, repeatability, and hysteresis by using reference sensors. As the results, the forceps had the RMS errors around 0.1 N with good repeatability and low hysteresis. However, the force transformation method could not be rigorously proved, since the included angle ($\alpha$) of the forceps was fixed and considered as a constant in the experiments. For that reason, the experiments using Raven-II surgical platform were conducted. The instrument was applied to the platform and the force sensing test was performed according to the transformation reflecting the change in the included angle ($\alpha$). Through the experiments, it was confirmed that the proposed method worked well.

The four-DOF force information at the tip of the forceps is useful for preventing tissue damage caused by surgeon’s grasping and incipient slips. It allows the force feedback control of robotic systems with improved surgeon’s operation skill. Furthermore, the sensorized forceps is easily fabricated, inexpensive, and even disposable.

In this study, we would like to note some challenges of the sensorized forceps to be investigated in the future. In the first, the proposed design is focusing on handling the tissue with the measurement of applied force only at the front portion of the
inner surface of the forceps, since tissues are typically manipulated and grasped with this part. However, in the case of surgical operations such as palpation, etc., the force sensing at the other regions of the forceps such as the end of the grasper’s tip is necessary because pushing or dragging on the tissues with the forceps is needed. In the second, at least three-DOF force sensing at each tip of the grasper is required in case of contacting tissues with only single side of the forceps. In this case, the proposed sensor just provides two-DOF force information, and thus, proper force feedback control cannot be achieved. In addition, miniaturization, embedding, and packaging of the sensor system are significant issues to clear the concern of sanitization.

As the extension of this study, we are developing a sensorized forceps integrating three-DOF force sensor to the inner distal region of the grasper’s tip. The instrument with the ability of sensing the external forces applied to all the surface of the grasper is under investigation, and three-DOF force sensing will be possible at each tip.

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Abstract—This paper presents a novel surgical instrument integrated with a four-degree-of-freedom (DOF) force sensor. By adopting the capacitive transduction principle, the sensor enables the direct sensing of normal and shear forces at surgical instrument tips. Thus, three-DOF pulling forces and a single-DOF grasping force can be measured for haptic feedback control of robotic minimally invasive surgery systems. The sensor consists of four capacitive transducers, and all the transducers including analog signal processing units are embedded in small surgical instrument tips. The four-DOF force sensing is enabled thanks to the four capacitive transducers by using the force transformation method. In this study, the instrument is designed and manufactured to be adaptable to the open-source surgical robot platform, called Raven-II. In addition, the sensing system is experimentally validated through its application to the Raven-II by using a reference force sensor.

Index Terms—Capacitive sensor, force sensor, haptic feedback, minimally invasive surgery (MIS).

I. INTRODUCTION

The application of robotic systems to minimally invasive surgery (MIS) has introduced a number of advantages in terms of postoperative pain, recovery time, and overall costs of medical treatment [1]–[4]. Although minimally invasive robotic surgery (MIRS) is advantageous for patients, it also involves some limitations. MIRS mainly relies on a surgeon’s skill to operate a master device to manipulate the surgical robot [5]–[7], and there is always the possibility of surgeon’s damage to the tissue due to excessive grasping force or possible slips during the operations [8]–[10]. To overcome this problem, the force-sensing system designed for MIRS has been studied by many research groups [11]–[18]. However, the difficulty in integrating the sensing system still hampers its introduction to surgical instruments in practical applications.

Fig. 1 shows some of the locations on an instrument, which are deemed to bear possibility at the onset of our experiment, which are the gripper, articulated joint, instrument shaft, and joint actuation unit. However, all but the gripper are not an appropriate candidate due to the following limitations. The sensor at the shaft and joint actuation unit may provide inaccurate force information due to variables such as inertia, friction, backlash, and gravity [9]. The articulated joint is also not an appropriate location because there are numerous mechanical disturbances such as driving mechanisms, friction, and backlash. On the contrary, force sensors placed at the tip of the instrument, which are colocated ones, can measure accurate force information without the aforementioned problems. The sensor can read multidimensional contact forces accurately because there is direct contact with tissue; thus, we can prevent excessive grasping forces and slipping during operations.

Recently, a number of studies have been performed to integrate force sensors at the tips of surgical instruments. King et al. [19] designed a piezoresistive tactile sensor integrated into the surgical instrument tip. Although this sensor can measure direct contact forces, it does not provide the essential functionality of a surgical gripper. Qasaimeh et al. [20] developed a PVDF-based microfabricated tactile sensor mounted on the
surgical grasper with preserved grasper shape. However, the approach provided only grasping force information, which is insufficient for deducing contact force information. Hong and Jo [21] fabricated a compliant forceps using two flexure hinges with the capability to sense two axial forces. Sensorized forceps can measure single-axis pulling and grasping forces. However, measurements of single-axis pulling force do not provide sufficient information to allow force feedback control for surgical robotic systems. Lee et al. [22] developed a multiaxis contact force sensor integrated into the surgical forceps with multiaxis force-sensing capability. However, it is difficult to measure accurate contact force information due to nonlinearity and complicated calibration of the sensor.

In our previous research [1], the concept of sensorized forceps consisting of two grasper-integrated force sensors was presented. On each side of the grasper, capacitive transducers with triangular prism structure are attached. From the readings of the transducers, the tissue handling forces, which are a three-degree-of-freedom (DOF) pulling force and a single-DOF grasping force, are extracted in real time. The sensing system is simple in its structure and, thus, appropriate for disposable instruments with low cost. Although the paper deals with the conceptual design of the sensorized surgical forceps, however, thorough analyses of the sensing principle and the force transformation method are not addressed. In addition, since the forceps were not integrated with an actual surgical robot, the performance verification had not been performed.

This study extends the previous research including the complete evaluation of the sensorized forceps and further performance analysis of the proposed sensing system. The forceps is validated via experiments at the instrument level. In addition, the instrument is designed to be adaptable to the open-source surgical robot platform, called Raven-II, for verification in the actual robotic system. In the actual surgical robot operating environment, the manufactured prototype instrument is evaluated through experiments using a reference force sensor. The force transformation method also is completely verified by reflecting the change of the included angle between two graspers during the operation. In addition, additional experiments about grasping force sensing are conducted for verifying the sensing’s feasibility. And, data analyses of the sensor are conducted to estimate the sensor’s performances such as force range, resolution, RMS error, repeatability, and hysteresis through additional experiments.

This paper is organized as follows: The development of sensorized forceps is introduced in Section II. Verification of sensorized forceps is explained in Section III. In Section IV, verification of the sensorized integrated system is presented in an experimental setup. Finally, discussion and the conclusion are given in Section V.

II. DEVELOPMENT OF THE SENSORIZED INSTRUMENT

A. Overview of Sensing System

Measuring contact and shear forces at surgical instrument tips is important to improve surgeon’s skill for preventing failures in robotic surgery [1], [23], [24]. Thus, the surgical graspers should be able to measure contact and shear forces with enough sensitivity to detect changes of force in handling soft tissue. And, the sensor’s force range should be capable of the maximum force applied during various surgical tasks using a surgical robot. In [5], the maximum stress applied to tissues such as liver was measured as about 200 kPa before those failures. The stress can be calculated as 5 N, assuming that the grasper’s surface is 25 mm². Forces applied to sutures using a surgical robot were detected to be less than 4 N [25]. To satisfy these requirements, a sensorized grasper composed of two grasper-integrated force sensors is designed as shown in Fig. 2. By specifically arranging capacitive-type sensor cells, contact and shear forces can be measured on the tips of the sensorized grasper while preserving the surgical grasper’s shape. Capacitive-type sensors consisting of a dielectric elastomer and two electrodes allow reduction in size, enhanced sensitivity, and proper force range [9], [26], [27]. The grasper provides information regarding all directional contact and shear forces applied to the tip. The force elements, i.e., two normal forces and two shear forces, can be transformed into a three-DOF Cartesian manipulating force and one-DOF grasping force by their geometric relation to the triangular prism.

B. Sensing Principles of Capacitive Transducer Cell

In general, the capacitance between two parallel conductors is obtained by

$$C = \frac{\varepsilon_0 \varepsilon_r A}{t}$$

where $C$ is the capacitance between the upper and lower electrodes, $\varepsilon_0$ represents the dielectric constant of air, and $\varepsilon_r$ is the relative permittivity of the dielectric elastomer. $A$ is the overlapping area, and $t$ is the distance between the two electrodes or the thickness of the polymer layer. Thus, it is noted that the capacitance depends on the distance between the upper and lower electrode layers and the overlapping area of the electrode layers.

As depicted in Fig. 3, the proposed sensor is composed of a dielectric elastomer substrate and two electrode layers. The sensor can measure two directional forces as changes in the
capacitance value between electrodes occur. Fig. 3(a) illustrates the movement of the proposed sensor when a vertical force is applied to the sensor surface. Vertical displacement causes changes in the dielectric elastomer’s thickness $t$, which alters the capacitance value. Since the applied force is directly related to displacement, changes in the capacitance enable the applied force to be computed. When the horizontal force is applied to the sensor surface, changes in the overlapping area between two electrodes occur along with a change in capacitance as presented in Fig. 3(b). In this case, the change of capacitance is very small, and accurate measurement cannot be guaranteed [22]. In this study, we have removed the influence of horizontal force by designing the lower electrode smaller than the upper electrode as shown in Fig. 3(c). Thus, any influence of capacitance changes caused by changes in the overlapping area is eliminated, and only pure vertical force sensing can be observed.

According to Hooke’s law, the relationship between the thickness and the applied force is estimated to be in line with each other; if the change of thickness is small, change in capacitance is also related to that of thickness. Thus, the relationship between capacitance and applied force with the thickness ($t$) is represented as

$$C = \varepsilon_0 \varepsilon_r \frac{A}{t(1 - \frac{F}{AE})}$$  \hspace{1cm} (2)

where $C$ is the capacitance, and $F$ is the force applied on the sensor’s surface. $A'$ is the sensor area exposed to an external force, and $E$ is Young’s modulus of the dielectric elastomer substrate. Fig. 4 illustrates the sensing principles of a grasper-integrated sensor. The sensor is composed of two capacitive-type sensor cells, a triangular prism, and a bottom structure. The two cells are adhered to the gap between the triangular prism and the bottom structure as shown in Fig. 4(a). The positive electrode is placed at the bottom surface, and the triangular prism structure is used as a common ground. As shown in Fig. 4(b), a normal force applied on the sensor’s surface deforms the two cells with symmetric capacitance responses. On the contrary, a shear force deforms the two cells with asymmetric capacitance changes as shown in Fig. 4(c), and thus, normal and shear force can be discerned. The combination of the force information measured by two capacitive-type sensor cells allows the derivation of the normal and shear forces applied to the surface of the grasper including the magnitude and direction of the external force.

### C. Force Transformation of Sensorized Forceps

As mentioned above, the four capacitance values measured by the sensorized forceps give normal and shear forces at the contact point and they need to be transformed into a three-DOF manipulating force and one-DOF grasping force. The three-DOF manipulating force indicates the 3-D Cartesian force, which can be detected while handling tissues. The one-DOF grasping force represents the squeezing force which grasps the tissue. As displayed in Fig. 5, the upper grasper-integrated force sensor measures normal and shear forces, and the lower force sensor measures normal and shear forces in a direction orthogonal to the upper ones. The geometry of the triangular prism structure
is described by a transformation matrix that relates force data from capacitive sensor cells to two normal forces and two shear forces as

\[
[ F_{\text{she}} F_{\text{nor}} F'_{\text{nor}} F'_{\text{she}} ]^T = T_{\text{NS}} \cdot [ F_{\text{cell1}} F_{\text{cell2}} F_{\text{cell3}} F_{\text{cell4}} ]^T
\]  

(3)

where \([ F_{\text{she}} F_{\text{nor}} F'_{\text{nor}} F'_{\text{she}} ]^T\) represents the vector of each normal and shear force, and \([ F_{\text{cell1}} F_{\text{cell2}} F_{\text{cell3}} F_{\text{cell4}} ]^T\) denotes the vector of four output forces measured by four cells as shown in Fig. 5. \(T_{\text{NS}} \in \mathbb{R}^{4 \times 4}\) transforms the four output forces into two normal and two shear forces. Based on the geometric configuration of the triangular prism, \(T_{\text{NS}}\) is obtained by

\[
T_{\text{NS}} = \begin{bmatrix}
\cos \theta & -\cos \theta & 0 & 0 \\
\sin \theta & \sin \theta & 0 & 0 \\
0 & 0 & \sin \theta & \sin \theta \\
0 & 0 & \cos \theta & -\cos \theta
\end{bmatrix}
\]  

(4)

where \(\theta\) is the angle of the four output forces on the coordinate based on normal and shear force vectors. Since the cross section of the triangular prism is a right-angled isosceles triangle, \(\theta\) is 45°. With Cartesian coordinates, the direction vectors of the normal and shear forces are determined using the included angle \((\alpha)\) between two graspers. These values are transformed into three-DOF pulling and one-DOF grasping forces such as

\[
[ F_{P_x} F_{P_y} F_{P_z} F_{G} ]^T = T_{\text{PG}} \cdot [ F_{\text{she}} F_{\text{nor}} F'_{\text{nor}} F'_{\text{she}} ]^T
\]  

(5)

where \([ F_{P_x} F_{P_y} F_{P_z} F_{G} ]^T\) indicates the vector of pulling and grasping forces. \(T_{\text{PG}} \in \mathbb{R}^{4 \times 4}\) transforms the normal and shear forces into manipulating and grasping forces. \(T_{\text{PG}}\) is formulated as

\[
T_{\text{PG}} = \begin{bmatrix}
\cos \alpha & -\sin \alpha & -\sin \alpha & 0 \\
\sin \alpha & \cos \alpha & -\cos \alpha & 0 \\
0 & 0 & 0 & 1 \\
0 & 0.5 & 0.5 & 0
\end{bmatrix}
\]  

(6)

where grasping force is defined as the average of two normal forces at the sensorized grasper, and pulling force is defined as the three-DOF pulling force with Cartesian coordinates. \(\alpha\) is the angle between the graspers. Depending on \(\alpha\), the manipulating and grasping forces are determined by the two normal and two shear forces of the graspers. The calculated transformation matrices can be used to calibrate the force values of the four sensing cells.

D. Fabrication of Grasper-Integrated Force Sensor

As shown in Fig. 6(a), the grasper-integrated force sensor consists of a triangular prism structure, a flexible printed circuit board (FPCB), an insulator layer, a base structure, and a chip case. Two electrodes are printed, and a capacitance-digit-converter (CDC) chip is soldered on the FPCB. The FPCB is placed on the V-shaped insulator, which isolates capacitance from the base structure. The triangular prism, base structure, and chip case are connected to the ground to minimize outer electromagnetic interference, since the electrodes and the CDC chip need to be shielded due to the capacitive-type sensor’s sensitivity to changes in outer electrical energy (noise) [30], [31].

The polymer layer is sandwiched between the triangular prism and two electrodes. The measured capacitance changes with the layer’s displacement caused by contact forces. Fig. 6(b) shows the other force sensor, which is rotated 90° along the y-axis and can measure the z-directional shear force. The assembled sensorized pair of forceps is shown in Fig. 6(c).

To fabricate the grasper-integrated force sensor, the insulator part of the FPCB is attached to the base structure. Then, the triangular prism structure is placed on the FPCB with two spacers that create a gap (0.1 mm) between the two electrodes in the FPCB and the triangular prism structure as shown in Fig. 6(a) and (b). The gap formed by the spacers is filled with the elastomer (PDMS). Finally, after the elastomer is cured, the spacers are removed.

The proposed instrument consists of a four-axial joint, tool shaft, joint actuation unit, and sensorized forceps, as shown in Fig. 7. The proposed instrument is compatible with an open-source surgical robot platform, called Raven-II, and its tool attachment interface exactly fits into that of Raven-II. The Raven-II is a seven-DOF robot composed of a four-DOF surgical instrument and a three-DOF manipulator [28]. In the joint actuation unit, four pulleys use a tendon-driven mechanism to create motion with four-DOF: a yaw motion by a working wrist joint, a rolling motion by a rolling joint, and pitching and grasping motions by two grasping joints. The proposed sensorized forceps measures applied forces on the tips of the surgical instrument. To verify the performance of the sensing system, a force gauge device and commercial force/torque sensor are used as...
The sensorized instrument is also tested via integration into the Raven-II. The Raven II has been proven as a reliable robotic surgery platform through various experiments [29]. The proposed sensor is fabricated as presented in Fig. 8. Fig. 8(a) shows a picture of the semiassembled grasper-integrated force sensor. The triangular prism, base structure, and chip case are made of aluminum alloy that possesses good electrical conductivity for passive shielding to minimize outer electromagnetic interference with capacitance. The AD7147A (2.1 × 2.3 mm²) produced by Analog Devices was selected as a small CDC chip. The AD7147 chip can measure capacitance with 650-Hz speed and 16-bit resolution. Using the I²C communication interface, the digitized capacitance is sent to the outer signal processing device without any additional device. The FPCB is 4 mm wide with four lines (two lines for I²C communication and two sensor power lines) that are connected to the MCU circuit through the hole in the instrument’s shaft. It is equipped with the CDC chip and two electrodes (2 × 2 mm²). The FPCB is adhered to a V-shaped insulator composed of polyether ether ketone, which was selected due to its high mechanical strength and electrical insulation.

A fully assembled sensorized grasper is shown in Fig. 8(b). The prototype of the sensorized instrument has wrist, rolling, and grasping joints to allow four axis motions and was manufactured with the proposed sensor, as presented in Fig. 8(c). Table I shows that the sensorized instrument’s Denavit–Hartenberg (D–H) parameters are the same as those of the Raven-II instrument [32]. The shared configuration provides compatibility between the developed instrument and the Raven-II surgical robotic platform.

### III. Verification of the Sensorized Instrument

#### A. Experimental Setup

As presented in Fig. 9, the experimental setup was built to test the developed sensor system. In Fig. 9(a), the setup consists of the sensorized instrument, a motorized stage assembled with a push–pull gauge (RX-2, AIKOH) to measure applied force, a manual stage for arbitrary input of external force, and jigs to fix the instrument in place. Force data measured by a push–pull gauge were used to calibrate the capacitive data measured by the sensor, inputting external forces by using the stages as shown in the right inset in Fig 9(a).

Fig. 9(b) shows the experimental setup used to verify the sensor performance of the sensorized instrument, a force/torque (F/T) sensor (Nano17, ATI), a simulated tissue, and a motorized stage that can move automatically.

For communication between the computer and sensor, a microcontroller unit (MCU) board was designed and included in the sensorized instrument. In the board, a chip (ARM cortex M3, 70 MHz) was selected for the MCU in order to provide sensor information in real time. The board reads capacitance data digitized by the CDC chip in the developed sensor and sends the data to the computer by controller area network communication.

One side of the tissue was fixed to the reference sensor assembled on the stage. The tissue, a sponge strip (5 × 3 × 40 mm³), was selected because its mechanical property (Young’s modulus: 160 kPa) was similar to the liver [5]. On the surface of the reference sensor, a jig was assembled, consisting of two bolts of which tips were covered with a plastic cover as demonstrated in the right inset of Fig 9(b). Using the jig, the tissue was fixed to the reference sensor. And, the sensorized forceps grasped the other side of the tissue by rotating and locking the pulley in the joint actuation unit of the instrument by the jig under the unit. The inset in Fig. 9(b) shows the difference between the x, y, and z coordinates of two sensors. For the comparison with data...
of the sensors, the coordinates were matched by multiplying a rotation matrix. When the stage moved, equal force was applied to the forceps and the reference sensor. The linear stage was controlled by a Lab-View program.

B. Calibration

Using the geometric relation of the sensor structure, all four forces were transformed into normal force, shear force, three-axial pulling forces, and a single-axial grasping force. In other words, the normal, shear, pulling, and grasping forces can be calculated using the aforementioned matrixes (see (3)–(6)) with simple calibration of the four sensor cell data. As shown in Fig. 9(a), experiments were conducted to calibrate data measured by the four capacitive-type sensor cells. To create a calibration matrix, the grasper-integrated force sensor was mounted on the jig. Using a motorized stage with a push–pull gauge, the sensor was then loaded to 3 N with the resolution (0.01 N) of the push–pull gauge. At this moment, the push–pull gauge measures the normal force \( F_{\text{nor}} \) applied on the grasper’s surface, and force at the sensor’s cell is treated as \( F_{\text{nor}}/2\sin \theta \) according to the transformation matrices mentioned above. To check the tendency of capacitance change by the external force, data at intervals of 0.5 N are selected, which are linearly fit as shown in Fig. 10. Each point is determined as the average of ten data was measured repeatedly. The distribution of the ten data is expressed by using error bar. Fig. 11(a) represents capacitance data measured by a sensing cell at 0.2 Hz repeatedly. The hysteresis was observed by increasing and decreasing the applied force three times repeatedly as shown in Fig. 11(b). The same experiment was conducted for the other grasper-integrated force sensor. Fig. 10 shows the linear characteristics of the loading forces and sensor output data. Using this linear characteristic, the relation between loading forces and sensor data can be expressed as [33]

\[
\begin{bmatrix}
F_{\text{cell}1} \\
F_{\text{cell}2} \\
F_{\text{cell}3} \\
F_{\text{cell}4}
\end{bmatrix} = \mathbf{A} \cdot \begin{bmatrix}
C_{\text{cell}1} \\
C_{\text{cell}2} \\
C_{\text{cell}3} \\
C_{\text{cell}4}
\end{bmatrix}
\]

(7)

where \([F_{\text{cell}1} F_{\text{cell}2} F_{\text{cell}3} F_{\text{cell}4}]^{T}\) indicates the force vector, and \([C_{\text{cell}1} C_{\text{cell}2} C_{\text{cell}3} C_{\text{cell}4}]^{T}\) is the vector of the sensor’s capacitance data at each capacitive-type sensor cell. \(\mathbf{A}\) is the calibration matrix that determines the relation between capacitance measured by the sensors and the four output forces. Because this relation is linear with one-to-one correspondence, the matrix is a diagonal matrix, and the diagonal elements were calculated as 0.02697, 0.01901, 0.01839, and 0.03763, depending on Fig. 10. Thus, two normal and shear forces, three-axial pulling force, and a single-axial grasping force were calculated as

\[
\begin{bmatrix}
F_{\text{she}} \\
F_{\text{nor}} \\
F'_{\text{nor}} \\
F'_{\text{she}}
\end{bmatrix}^{T} = \mathbf{T}_{\text{NS}} \cdot \mathbf{A} \cdot \begin{bmatrix}
C_{\text{cell}1} \\
C_{\text{cell}2} \\
C_{\text{cell}3} \\
C_{\text{cell}4}
\end{bmatrix}
\]

(8)
\[
[F_{P_x}, F_{P_y}, F_{P_z}, F_G]^T = T_{PG} \cdot T_{NS} \cdot A \cdot [C_{cell1}, C_{cell2}, C_{cell3}, C_{cell4}]^T.
\] (9)

C. Testing With a Sensorized Surgical Instrument

For the performance verification, pulling and grasping forces measured by the proposed sensorized instrument were compared with those measured by a reference sensor in Cartesian coordinates using the experimental setup as shown in Fig. 9(b). The measured forces were estimated based on the measured cell forces and the transformations in (8) and (9). Data acquisition of two sensors was conducted with the same sampling rate and under a common clock. The four capacitance values of the four sensing cells were simultaneously read, while the forceps grasped the tissue fixed to the reference part. The data were calibrated using the method mentioned above. Fig. 12(a) shows the force data obtained by grasping and pulling with the forceps. The first increase in data values indicates that the tissue was grasped by the forceps. The motorized stage assembled to the reference sensor automatically pulled and released the tissue fixed between the forceps and the reference sensor. In this experiment, the four pulling motions were carried out in 48 s, with a 12-s cycle. Fluctuations in force data are explained as repeated pull and release of tissue.

Data transformed by the transformation matrix in (8) are presented in Fig. 12(b). The transformation process was conducted in real time. Fig. 12(b) shows the normal and shear forces calculated by the transformation matrix. In these graphs, the upper two data points indicate the shear and normal forces measured by the upper grasper. The other points are the shear and normal forces measured by the lower grasper. During the grasping and pulling motions, the two normal forces \((F_{nor}, F'_{nor})\) increased by about 4 N each, and the two shear forces \((F_{she}, F'_{she})\) changed by about 0.72 and −0.15 N, respectively. Since the pulling motion was inclined toward the \(x\)-axis, the measured \(F_{she}\) was larger than the \(F'_{she}\). The pulling and grasping forces of the developed sensor and reference sensor responses are presented in Fig. 12(c). These forces were calculated using the transformation matrix in (9). The pulling force elements \((F_{P_x}, F_{P_y}, F_{P_z})\) were measured up to 0.72, −0.15, and −0.33 N, respectively. The graphs indicate similarities between responses of the developed sensor and reference in time-domain responses. Through the experiments, the RMS errors were calculated as 0.08, 0.07, and 0.11 N. The grasping force was measured up to about 4 N. Because there are no reference data comparing grasping forces in this experiment, additional experiment was conducted. Grasping force \((F_{Gr})\) is the average of two normal forces, and the two normal forces mean grasping forces at upper and lower graspers, respectively. Therefore, two normal forces were compared in the experimental setup as presented in Fig. 9(a). The grasper placed on a manual stage contacts the probe of the push–pull gauge with no load. And, in this stage, normal force was applied arbitrarily, with the measurement of the instrument and the push–pull gauge as references during 60 s. Fig. 13 shows similarities between responses of the developed sensor and reference. Through the experiments, the RMS errors of the pulling and grasping forces were calculated as 0.0837, 0.0732, 0.114, and 0.0957 N. The sensor’s force ranges are determined as \(\pm 2.5, \pm 5, \pm 2.5,\) and 5 N, respectively. It is possible to detect the force applied to tissue up to the safety threshold of 200 kPa for cell apoptosis in abdominal organs [5]. The repeatable errors for four force values were found to be 1.23%, 1.58%, 1.34%, and 1.56% of the entire force range.
Fig. 13. Time-domain responses of two normal forces measured by two grasper-integrated force sensors and a push–pull gauge (red).

Fig. 14. Picture of the developed sensorized surgical instrument integrated with a Raven-II surgical robot platform.

Fig. 15. Experimental image during grasping and pulling of simulated tissue. The sensorized forceps and reference sensor simultaneously measured four force elements for verification. (a) Experimental set-up composed of the instrument, reference sensor, and the tissue. (b) Scene that the forceps grasp and pull the tissue.

<table>
<thead>
<tr>
<th>Specifications of the Proposed Sensorized Instrument</th>
<th>$F_{P_x}$</th>
<th>$F_{P_y}$</th>
<th>$F_{P_z}$</th>
<th>$F_G$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force range</td>
<td>±2.5 N</td>
<td>±5 N</td>
<td>±2.5 N</td>
<td>5 N</td>
</tr>
<tr>
<td>Resolution</td>
<td>42 mN</td>
<td>58 mN</td>
<td>72 mN</td>
<td>46 mN</td>
</tr>
<tr>
<td>RMS Error</td>
<td>0.0837 N</td>
<td>0.0732 N</td>
<td>0.114 N</td>
<td>0.0957</td>
</tr>
<tr>
<td>Repeatability</td>
<td>1.23%</td>
<td>1.58%</td>
<td>1.34%</td>
<td>1.56%</td>
</tr>
<tr>
<td>Hysteresis</td>
<td>1.96%</td>
<td>2.16%</td>
<td>2.03%</td>
<td>1.75%</td>
</tr>
</tbody>
</table>

Table II

after performing the experiments four times repeatedly. And, hysteresis errors were found to be 1.96%, 2.16%, 2.03%, and 1.75% of the force range [34]. The results are summarized in Table II. The transformation occurred in real time (650 Hz). The change of $\alpha$ affects force data of $F_{P_x}$ and $F_{P_y}$ as presented in Fig 5. However, $\alpha$ was set as a constant value on account of the angle between the graspers during this experiment.

IV. Verification of the Integrated System

A. Experimental Setup With the Raven-II Surgical Robot System

The sensorized instrument was integrated with the open-source surgical robot platform Raven-II to verify the entire surgical robotic system, as shown in Fig. 14. Because the instrument was designed with the same mechanical parameters and transmission ratio as the Raven-II, there were no control issues [32]. The Raven-II uses the ROS interface [35], and thus, an additional ROS interface node was designed to communicate sensing data measured from the instrument. In this process, force data calibration and transformation were conducted by determining the included angle ($\alpha$). The ROS interface provided the bidirectional communication between the Raven-II and the developed instrument, enabling force feedback control using a haptic device as the master. In the experiment, position control of the robot was conducted by using a remote master (Phantom Omni, Sensable Co.). In this setup, for the verification of the developed sensor, the reference sensor was fixed on the ground with the jig mentioned above. During experiments, information of $\alpha$ and the robot’s coordinate from its controller were reflected to match and compare the data measured by two sensors.

B. Experiments With the Raven-II Surgical Robot System

Experiments were conducted to verify the developed surgical instrument in a simulated surgical environment. The verification was performed by comparing force data measured by the sensorized forceps and the reference sensor. In addition, the proposed force transformation method was completely proved by reflecting the change of the angle ($\alpha$) as a variable to the matrix. As shown in Fig. 14, the forceps of the integrated instrument grasped and pulled the simulated tissue secured to the jig assembled to the reference sensor. As mentioned above, the forceps measured pulling force in Cartesian coordinates from the forceps position. As is shown, the coordinates of the forceps and reference sensor points differ. The forceps coordinates were rotated to match the reference sensor coordinates using joint position information from the Raven-II manipulator. During the
C. Experimental Results

During force measurements with the sensorized forceps and reference sensor, the robot delicately manipulated the tissue as in the actual robotic surgery situation. Tissue grasped by the forceps was pulled repeatedly in random directions for 2 min. The sensing data sampling rate was 650 Hz in the experiment and the $\alpha$ was read simultaneously as shown in Fig. 16. Fig. 17(a) shows the measured normal and shear forces. The upper grasper measured the upper normal and shear forces, while the lower grasper measured the lower forces. The maximum normal and shear forces applied were 2.65, 2, 1.55, and 0.15 N, respectively. Fig. 17(b) represents the pulling and grasping forces measured by the forceps and reference sensor. The $\alpha$ in matrix [see (9)] used to calculate pulling and grasping forces was reflected as the grasper joint angle of the robot. In addition, according to the data shown in Figs. 16 and 17, it is noted that larger grasping force is needed when the forceps pull tissue. The larger force changes the angle $\alpha$ by deforming the tissue. The applied maximum pulling force was measured up to 1.45, 0.21, and 0.15 N in Cartesian coordinates. The maximum grasping force was also measured as 2.1 N. The pulling force of the sensorized forceps matched that of the reference sensor data, and the RMS errors were 0.14, 0.07, and 0.05 N. In the experimental setup using the Raven surgical robot, the grasping force cannot be compared because we do not have reference data. Instead, we verified the grasping force in the instrument level. Because the elements used to calculate a grasping force are related to the pulling force, verification of the pulling force can be used to validate the measured grasping force.

V. DISCUSSIONS AND CONCLUSION

In this paper, a novel surgical instrument integrated with a four-DOF force sensor was proposed. The instrument could measure three-DOF pulling forces and single-DOF grasping force directly at the tips. For this purpose, two grasper-integrated capacitive force transducers were embedded in the forceps of the instrument. The compact design of the sensor was realized with a triangular prism structure and two capacitance sensing cells for each tip while preserving the grasper’s shape. The readings of each cell were transformed into two normal and two shear forces. Three-DOF pulling forces and single grasping forces were computed by using the transformation matrix based on the sensor’s geometry relationship.

The instrument was tested focusing on the factors such as resolution, RMS error, repeatability, and hysteresis by using reference sensors. As the results, the forceps had the RMS errors around 0.1 N with good repeatability and low hysteresis. However, the force transformation method could not be rigorously proved, since the included angle ($\alpha$) of the forceps was fixed and considered as a constant in the experiments. For that reason, the experiments using Raven-II surgical platform were conducted. The instrument was applied to the platform and the force sensing test was performed according to the transformation reflecting the change in the included angle ($\alpha$). Through the experiments, it was confirmed that the proposed method worked well.

The four-DOF force information at the tip of the forceps is useful for preventing tissue damage caused by surgeon’s grasping and incipient slips. It allows the force feedback control of robotic systems with improved surgeon’s operation skill. Furthermore, the sensorized forceps is easily fabricated, inexpensive, and even disposable.

In this study, we would like to note some challenges of the sensorized forceps to be investigated in the future. In the first, the proposed design is focusing on handling the tissue with the measurement of applied force only at the front portion of the
inner surface of the forceps, since tissues are typically manipulated and grasped with this part. However, in the case of surgical operations such as palpation, etc., the force sensing at the other regions of the forceps such as the end of the grasper’s tip is necessary because pushing or dragging on the tissues with the forceps is needed. In the second, at least three-DOF force sensing at each tip of the grasper is required in case of contacting tissues with only single side of the forceps. In this case, the proposed sensor just provides two-DOF force information, and thus, proper force feedback control cannot be achieved. In addition, miniaturization, embedding, and packaging of the sensor system are significant issues to clear the concern of sanitization.

As the extension of this study, we are developing a sensorized forceps integrating three-DOF force sensor to the inner distal region of the grasper’s tip. The instrument with the ability of sensing the external forces applied to all the surface of the grasper is under investigation, and three-DOF force sensing will be possible at each tip.

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