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Modular Instrument for a Haptically-Enabled Robotic Surgical System (HeroSurg)

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ABSTRACT To restore the sense of touch in robotic surgical systems, a modular force feedback-enabled laparoscopic instrument is developed and employed in a robotic-assisted minimally invasive surgical system (HeroSurg). Strain gauge technology is incorporated into the instrument to measure tip/tissue lateral interaction forces. The modularity feature of the proposed instrument makes it interchangeable between various tip types of different functionalities, e.g., cutter, grasper, and dissector, without losing force sensing capability. Series of experiments are conducted and results are reported to evaluate force sensing capability of the instrument. The results reveal mean errors of 1.32 g and 1.98° in the measurements of tip/tissue load magnitude and direction across all experiments, respectively.

INDEX TERMS Surgical instruments, force feedback, calibration, strain measurement, surgical robotics modularity.

I. INTRODUCTION

Current robotic assisted minimally invasive surgical systems (RAMIS) suffer from a lack of haptic feedback. While high-definition and 3D displays of modern endoscopic systems can be used to estimate the interaction forces throughout an operating procedure, providing both visual and force feedback leads to more efficient teleoperation and less tissue damage by reducing the interaction forces and number of operational errors [1]–[4]. Studies have revealed 2-6% reduction in applied force, 30-60% in RMS force, 60% in error, and 30% in surgical completion time with haptic feedback [5]–[8]. A recent study has also discovered that the addition of haptic feedback results in an average of 3.1-fold less applied force on the tissue, better tissue discrimination, and higher confidence in decision-making [9].

Enabling a force feedback capability has been the main motivation for the development of specially designed stand-alone actuated laparoscopic instruments with force measurement capability [10], [11]. These instruments are similar to conventional laparoscopic tools; however, they use novel designs to incorporate force and tactile sensing by using strain gauges, capacitive-type sensor cells, Bragg grating sensors, and complex six degrees of freedom (DOF) force-torque sensor based on a Stewart platform [12]–[17]. They also incorporate the advantages of actuation mechanisms for utilization in robotic surgical systems [18]–[21]. These methods have been significant advancements towards incorporating force feedback in RAMIS; however, they do not promise both force measurement features and modularity to exchange tool type during operation without losing force sensing capability. Methods such as sliding perturbation observer (SPO) and torque transfer system (TTS) have also been proposed to estimate the grasping forces in surgical robotic instruments [22], [23]. These methods are more straightforward to implement compared to hardware-based solutions; however, they tend to compromise accuracy and efficiency due to assumptions exploited in their models [24].

The main contribution of this study is the development of a modular force feedback-enabled instrument capable of measuring tip/tissue interaction forces without any sensor at the tip. The modularity feature of the proposed instrument enables quick replacement between a variety of laparoscopic tool types such as grasping, cutting, and dissecting without losing the force sensing feature. A calibration and actuation module is also specially designed and developed to facilitate fast calibration of the instrument.

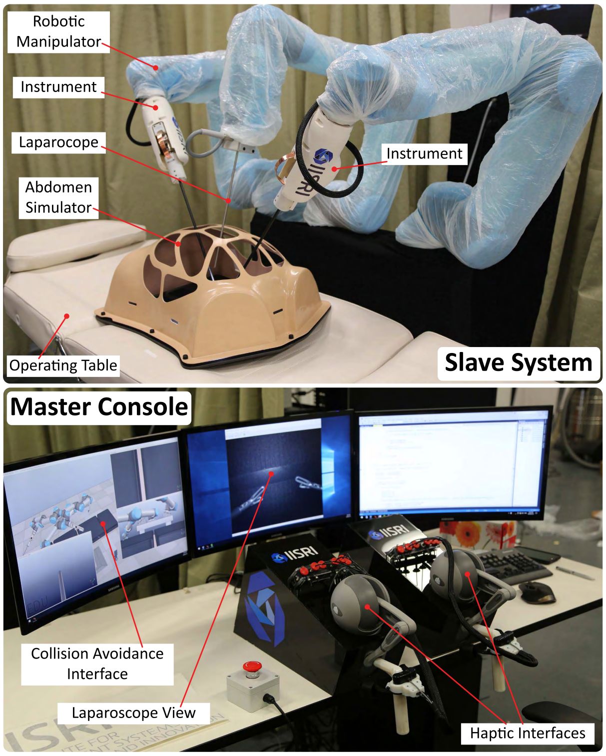


FIGURE 1. Master and slave console of HeroSurg system - a robotic assisted surgical system with haptic feedback, collision avoidance and automatic bed/patient/tissue motion compensation capabilities.

The proposed instrument has been directly utilized in a haptically-enabled robotic surgical system (HeroSurg) (Figure 1). HeroSurg is a robotic surgical system with key features of haptic feedback, collision avoidance and automatic bed/patient/tissue motion compensation. The developed instrument is the next significantly enhanced version of the previously developed instrument for a parallel robot assisted minimally invasive surgery/microsurgery system (PRAMiSS) [19]. The modularity feature, force sensing capability, and actuation mechanism of the proposed instrument have been considerably improved and are proposed in this paper.

In the following section, the features and capabilities of the developed instrument are described. Section III explains the developed calibration and actuation module as well as the calibration process and results. The experimental setup and procedure are explained and the results are reported in section IV. Section VI provides concluding remarks.

II. FEATURES AND CAPABILITIES

The key features and capabilities of the proposed instrument shown in Figure 2 are force sensing, modularity and actuation mechanism. This mechanism was especially designed to allow quick replacement of various tip types without losing force sensing capability. These features are explained in details in the following sections.

A. FORCE SENSING CAPABILITY

To facilitate sensing of sideways tip/tissue interaction forces, a 7 mm in diameter stainless steel tube (Figure 3a) was developed as a sleeve to fit concentrically over 5 mm laparoscopic inserts [19]. Translational displacements of the insert rod along the sleeve operate the tip jaws with different functionalities. The lateral tissue interaction forces applied to the tip produce bending in the sleeve. Thus, the tip/tissue lateral interaction forces can be estimated by evaluating surface strains along the sleeve. Two half-bridge strain gauges (four strain gauges with 7.4 mm length, 3.3 mm width and 350 Ω resistance) are applied to the opposite sides of the outer surface of the sleeve (Figure 3b). These half-bridge strain gauges are referred to as *bending bridges* in the rest of the paper. The magnitude of the sideways loads applied to the tip can be estimated as the vector sum of the outputs of these half-bridge configurations.

In order to avoid capturing the strains created in the sleeve at the trocar port and provide intra-abdominal force measurement capability, the strain gauges are applied to the sleeve at points close to the tip (Figure 3a). This ensures that the strain gauges are lied inside the abdominal wall during laparoscopic procedures. To increase the measurement sensitivity, a portion of the sleeve, where the strain gauges are applied, is machined to contain a smaller cross-sectional area. This increases the surface strains captured by the strain gauges that are produced by the lateral forces applied to the tip. On the portion of the tube with a larger cross-sectional area, four grooves are made along the tube axis to accommodate wires of the strain gauges for sterilization purposes (Figure 3b). The strain gauges and wires are covered with a protective coating and also a heat-shrink tube for protection purposes during experiments (Figure 3a).

In the previous instrument [19], [25], the strain gage signal leadwires were directly attached to the soldering tabs of the four gages. This means that there are two leadwires per strain gage; 8 in total. Since these are the main signal leadwires coming out of the force sensing sleeve and may be in contact with other mechanical components, they need to be durable with multi-layer coating for reliability and sterilizability purposes which make them relatively think. Soldering thick wires directly to the tiny and very close soldering tabs of the strain gages are difficult and prone to potential noise due to installation errors. Furthermore, tension forces transmitted along the main leadwires may damage the strain gages or degrade their performance. As shown in Figure 3b, in the new design presented in this paper, two sets of three-tab bondable printed-circuit soldering terminals are employed that are bonded to the force-sensing sleeve (Figure 3b-Step1). Using these terminals makes the strain gages application more compact, easier, and more durable. They also decrease the number of main leadwires to 3 per two gages; 6 in total for four gages. In this configuration,

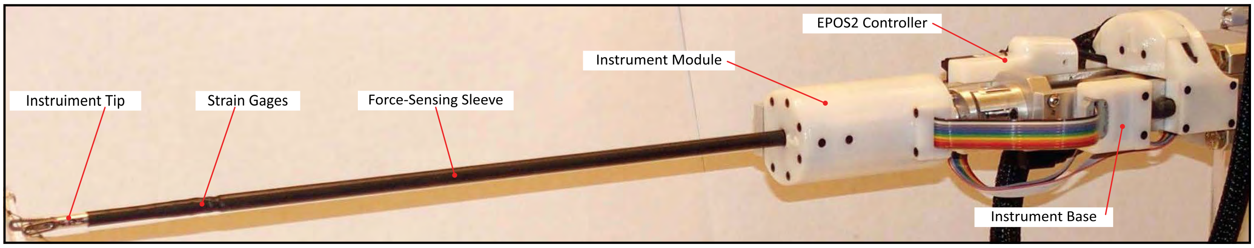
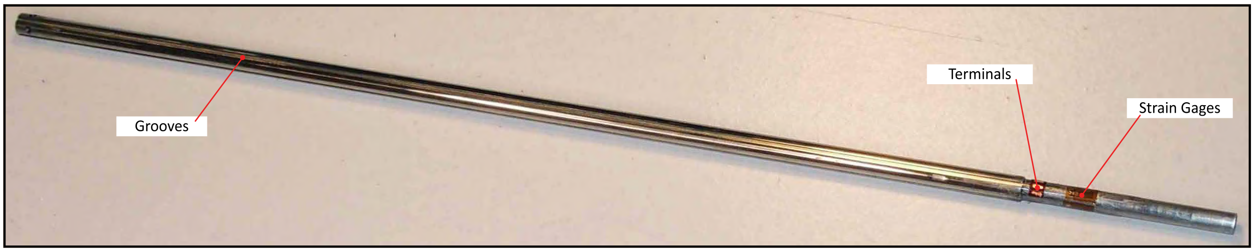
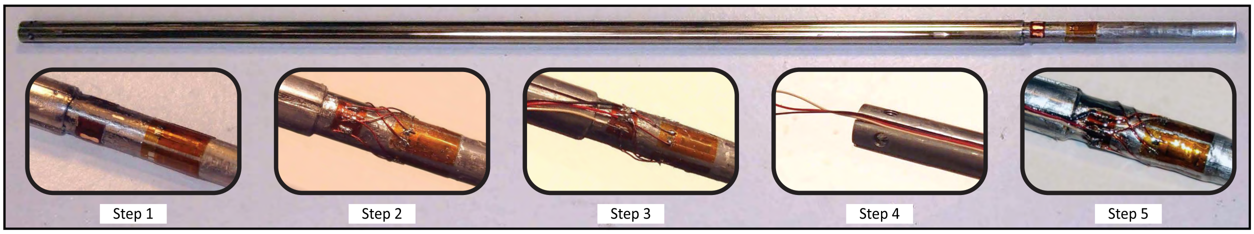


FIGURE 2. Haptically-enabled modular laparoscopic instrument.



(a)



(b)

FIGURE 3. (a) Force sensing sleeve (b) and developmental steps for installation of strain gages, bonding terminals, and signal wires.

very thin, delicate, and easier to install jumper wires can be used for connectivity between the strain gages and terminals (Figure 3b-Step2).

Terminals are applied to the sleeve close to the strain gages to shorten the jumper wires (Figure 3b-Step2). On the other side of the terminals, coated multi-stranded copper leadwires are soldered to provide reliable and less noisy signal connectivity (Figure 3b-Step3). Use of terminals also provides an anchor for both sets of leadwires to prevent forces transmitted along the main leadwire system from damaging the strain gages or degrading their performance (Figure 3b-Step4). Another improvement of the current design of the instrument over the previous one in terms of force sensing capability is the use of amplifiers and high performance data acquisition (DAQ) system the leads to higher quality signals.

It is worth noting that the high voltage occurring during radio-frequency ablation power delivery may lead to corrupted measurements of bending strains installed on the force-sensing sleeve. Solutions have been proposed to cope with this limitation by using a radio-frequency ablation equipment embedded at the tip of the instrument that allows for synchronization of power delivery and data acquisition [26]. Similar approach can be integrated into the proposed instrument, however a separate instrument is required to provide this capability for ablation as well as force measurement for haptic feedback.

B. MODULARITY AND ACTUATION

Modularity is a key feature of the proposed instrument that allows for a fast interchange of a variety of laparoscopic insert types, e.g. grasping, cutting and dissecting without loss of the force sensing capability (Figure 4). Figure 5 presents the actuation and attachment mechanism of the instrument module to the instrument base. The instrument module can be attached to the base module using a rear rectangular pin that, once inside the hole, is fixed in position using a thumbscrew. When the instrument is mounted and fixed in position, a cable is connected to the female connector of the instrument base to transfer the strain gauge signals (Figure 4).

When attaching the instrument module to the base, the round component attached to the instrument module slides into a groove built into the lead part of a lead-screw mechanism on the base. This lead-screw mechanism transmits the rotational motions of the actuator shaft to the linear motion required to operate the tip jaws (Figure 4).

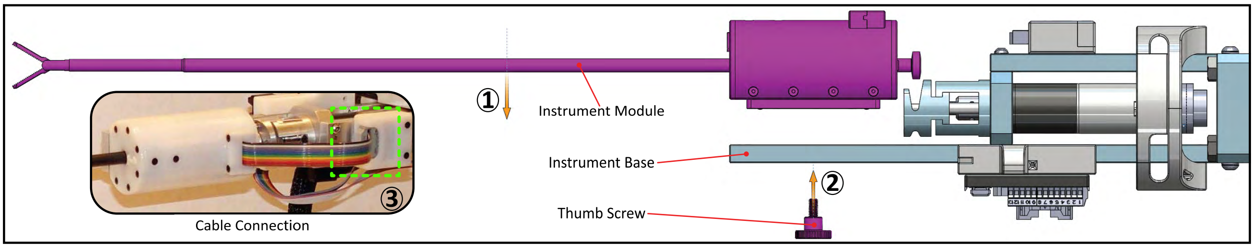


FIGURE 4. Components and assembly order of the instrument for easy and quick attachment to the base module.

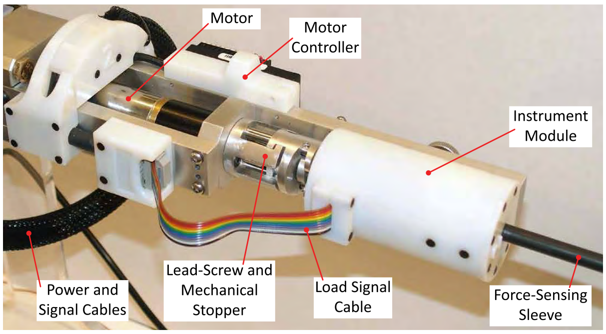


FIGURE 5. Instrument actuation and module/base mounting mechanisms for easy and quick attachment of instrument and base modules.

A mechanical stopper is embedded into the lead-screw mechanism to limit the linear displacement of the lead part to 2 mm and avoid any excessive tension or pressure in the insert.

In terms of modularity and actuation, although in both versions of the instruments, it is possible to switch between the instrument types, the exchange process in the new design is significantly quicker and easier and also results in higher force measurement accuracy. In the previous version, some of the mechanical components need to be disassembled to access and replace the insert type which takes more than 15 minutes. This requires the re-calibration of the strain gages. Another method of changing the instrument type in the previous version of the instrument is by exchanging the entire instrument module including base and the two actuators which may not be practical and cost effective. As shown in Figure 4, the new instrument presented in this paper has a modular design in a way that the instrument and base modules are completely separated and can be attached/detached with no re-calibration requirement in less than 5 seconds.

A DC geared motor (Maxon Motors Inc. Model EC-max 22, 6072 Sachseln, Switzerland) is utilized for the actuation of the tip jaws (Figure 5). To avoid any electromagnetic interference on the strain gauge signals caused by the actuator or its controller, shielded cables are utilized for transferring strain gauge signals, controller data and power. This, compared to the previously developed instrument [25], significantly improves the quality of signals. It also enables the motor controllers to be positioned on the instrument base which further decreases the noise, and also allows a longer functional distance of the instrument from the data acquisition (DAQ) unit.

III. CALIBRATION

Each instrument has a sleeve that is integrated with strain gages and assembled with insert and mounting components to form one single instrument. The insert can have any tip type. e.g. grasping or cutting. Once they are assembled, the instrument and not the force sensing sleeve will be calibrated. Therefore, once engaged and assembled, the sleeve and insert will never be disengaged unless for maintenance purposes in which case the calibration process will be repeated. Therefore, to change the tip type during the surgical operation, the entire instrument part (and not the insert or sleeve components) will be replaced with another one that has the desired tip functionality (Figure 4). This can be achieved by help of a thumb screw designed for this purpose without any disassembly/assembly of the instrument components including sleeve or insert. Two independent half bridges of strain gauges were incorporated into the force-sensing sleeve of the proposed instrument. Theses were used to measure the bending forces in the instrument sleeve. To obtain the relationships between the bridge signals and the applied loads, a calibration process was performed independently for each bridge configuration. A calibration and actuation module was specifically developed in order to facilitate the calibration process.

A. CALIBRATION SETUP

The calibration module presented in Figure 6 is developed to rotate the instrument around the insert axis while keeping it horizontally. A horizontal configuration of the instrument is chosen to enable the use of hanging masses for applying lateral interaction forces at the instrument tip. A DC geared motor (Model 328065, Maxon Motors Inc.) is utilized in the calibration module for the angular displacements of the instrument. The height of the module is chosen to provide enough space for one turn rotation of the signal and power cables of the instruments around it. It should be noted that the calibration module is merely used for calibration process and in real operation the instrument base will be mounted and manipulated by a robot manipulator.

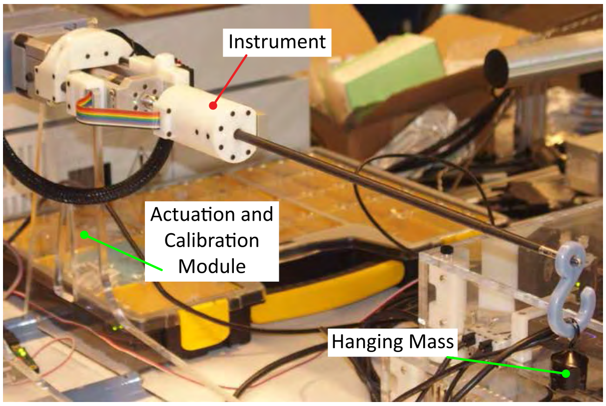


FIGURE 6. Calibration module to keep and rotate the instrument around its axis while a hanging mass is applying constant lateral interaction force at the instrument tip. This module is used for calibration of the strain gauges applied to the force-sensing sleeve for the measurement of the sideways interaction loads at the tip.

B. CALIBRATION PROCEDURE AND RESULTS

The calibration module (Figure 6) is used in the calibration process to apply known masses to the instrument tip and generate bending in the sleeve. In light of the inherent linearity between the applied loads and bridge signals, the least-squares method was employed to develop a linear system for each bending bridge represented as $v = mf + b_y$, where v and f are the bridge output voltage and applied loads and m and b_y are the slope and y-intercept of the linear model, respectively. The known loads of 20, 40, 50, 70, 90, and 100 grams were used in this process. The hanging hook (Figure 6) was also measured as 6 grams. The bending bridge signals were captured using a data acquisition system (LabJack U6 Pro, LabJack Corp., Lakewood, CO, United States). The excitation voltages of 5 V were applied. The calibration processed was repeated three times for each hanging mass.

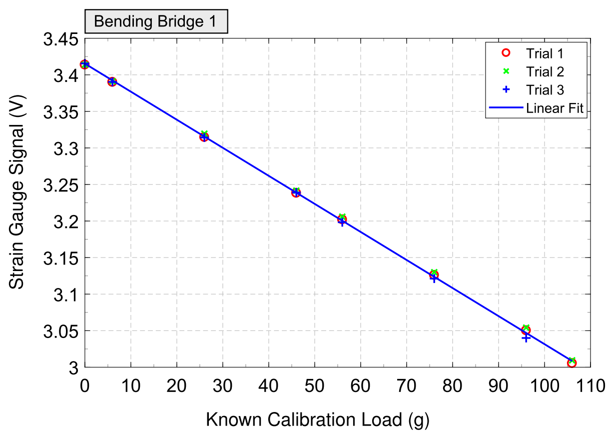


FIGURE 7. Data samples for three repetitive calibration experiments and the least-squares linear fit for one of the strain gauge bridges applied to the force-sensing sleeve for the measurement of the lateral interaction forces.

The raw calibration data captured from three repetitive cycles as well as the best fitted linear curve for one of the bending bridges are presented in Figure 7. Due to the existence of the initial loads, there is a no-load voltage. Theses initial loads may be due to the protective covers, temperature, or even errors in the application of strain gauges. As implied from this figure, the results verify the monotonic responses of the strain gauge bridge that is a desirable characteristic for considering a linear relationship between the applied loads and the bridge signals. The step response time is smaller than 4.6 millisecond. This is well above the required bandwidth for force reflecting teleoperated systems [27]. In the typical moderately controlled lab environment, drift observed across one day of experiments was less than 3 grams. This is certainly more than the length of the time of typical surgical operations which involves the human body as a regulated thermal environment.

The instrument is calibrated in horizontal position. Using instrument in any other orientation may cause slight error due to the gravitational effects. This potential error is expected to be less than 5 grams because of the lightweight force-sensing sleeve and the fact that the strain gages are applied close to the tip of the instrument. Moreover, the manipulations in surgical operations are usually slow with low acceleration that further trivialize the gravitational effects on the force measurement accuracy.

IV. EXPERIMENTS

Experiments were conducted to evaluate the accuracy and performance of the proposed instrument in measuring sideways tip/tissue interaction forces. During the experiment presented in Figure 8, the instrument was held horizontally and known masses were applied to the tip using a ring-shaped part hanging at the tip. The calibration and actuation module (Figure 6) was used to hold and rotate instruments in this experiment. While the mass was applying a constant lateral force at the tip, the instrument was rotated once through 360 degrees around its axis. Figure 9 presents different stages of the rotation in the experiment. The strain gauge signals were captured and served as inputs for the calibration models developed in the calibration process to estimate the magnitude and direction of lateral loads applied at the tip of the instruments.

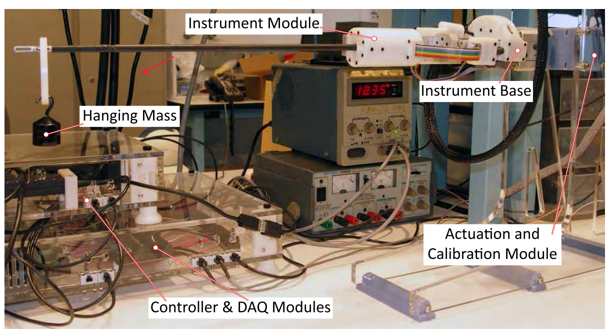


FIGURE 8. Experimental setup used to hold the instrument horizontally and rotate it while a constant lateral force is applying to the instrument tip by using a hanging ringed-shape mass at the tip.

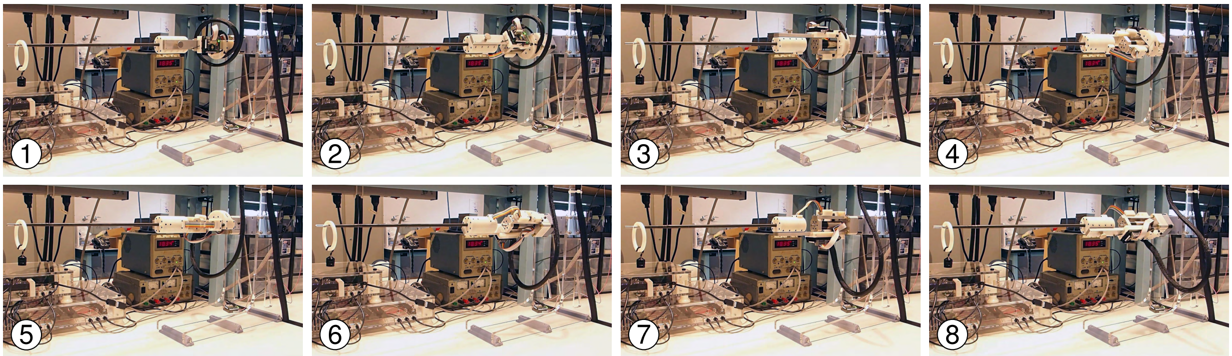


FIGURE 9. Different stages of the experiment to evaluate the accuracy and performance of the proposed instruments in measuring sideways tip/tissue interaction forces. The instrument was rotated once around its axis while it was held horizontally and known masses were applying to the tip using a ring-shaped part hanging at the tip.

The force magnitudes of the applied load were estimated as the vector sum of the two orthogonal load components as measured from the bending bridges. This data was then compared with the applied known hanging loads. Three known calibration weights of 20, 50, and 100 grams were chosen for this experiment. The ring-shaped part was also measured as 4.8 grams. These loads were deliberately selected from the lower range of the typical forces applied in laparoscopic surgical operations in order to evaluate the sensitivity of the force measurement capability of the proposed instrument [28]–[30]. The tip jaws were closed during the experiments. Each experiment was repeated three times for each known calibration load.

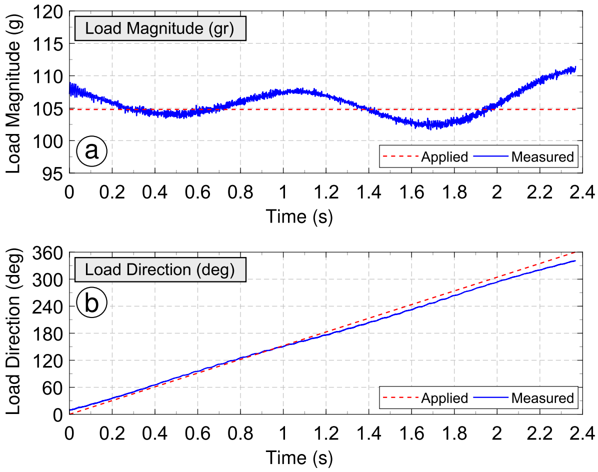


FIGURE 10. Raw experimental data for the magnitude (a) and direction (b) of the applied and measured loads for the known load of 104.8 grams during one turn rotation of the instrument (the maximum load applied in the experiment is chosen for this plot to present the worst tracking errors in the experiments.)

Figure 10 presents a sample raw data for the magnitude (Figure 10a) and direction (Figure 10b) of the applied and measured loads for the known load of 104.8 grams during one turn rotation of the instrument. The maximum load applied in the experiment is chosen for this plot to present the worst

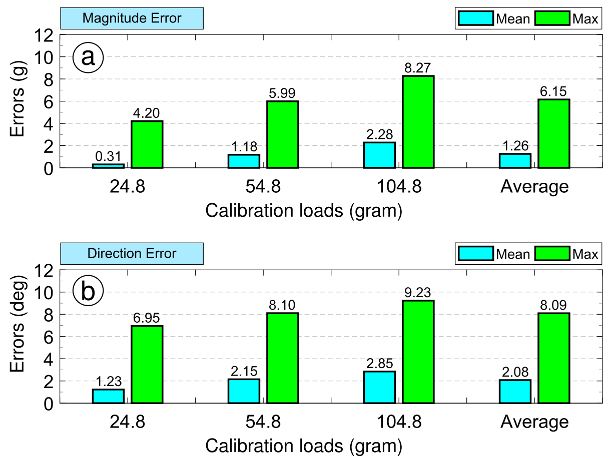


FIGURE 11. Mean and maximum errors in the magnitude (a) and direction (b) of the measured loads across three repetitive trials for the three known loads of 24.8, 54.8, and 104.8 grams as well as errors across all trials and loads during one turn rotation experiment.

tracking errors observed in the experiments. These sinusoidal errors may be because the strain gages are not evenly separated by angles of 90 degree which requires further investigation and study to compensate for this application error. Figure 11a presents the mean and maximum errors in the magnitude of the measured loads across three repetitive trials as well as errors across all trials and loads. The results reveal an increase in both the mean and maximum values of the errors as the load increases. The reason of this increase may be due to small vibrations caused in the long sleeve during the continuous rotation of the instrument. The mean error for the largest load of 104.8 grams is as low as 2.80 grams and the maximum error is 8.27 grams. The last column in the plot of the Figure 11a shows the average values for the mean and maximum magnitude errors across all loads and trials which are 1.26 and 6.15 grams, respectively.

Besides the magnitude of the tip/tissue sideways interaction forces, the direction of these forces can also be measured as the angular relationship between the two measured load components from the bending bridges. The accuracy and effectiveness of the force direction measurement capability of the proposed instrument was also evaluated. This was achieved by comparing the estimated load directions with the input position profile commanded to the actuator of the calibration and actuation module.

The mean and maximum errors in the measured direction of the applied loads for the three repetitive trials, as well as the errors across all trials and loads are presented in Figure 11b. Similar to the errors in the load magnitude measurements, the results show an increases in both the mean and maximum values of the errors in the measured direction for larger loads. Since the errors in the magnitudes of the loads directly affects the estimated direction, the behavior in the results of the estimated direction may be also related to the vibrations generated in the sleeve. The mean errors in the measured direction for the largest load is as low as 2.85 deg for the mean error and 9.23 deg for the maximum error. The last column of the plot (Figure 11b) shows the average values of the mean and maximum direction errors across all loads which are 2.08 and 8.09 deg, respectively.

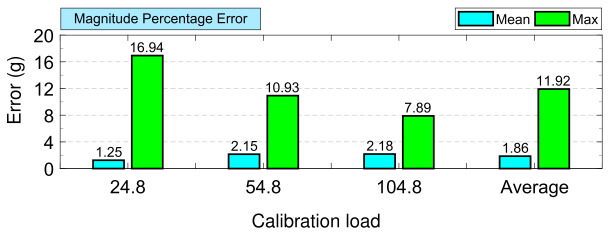


FIGURE 12. Mean and maximum percentage errors in the magnitude of the measured loads across three repetitive trials for the three known loads of 24.8, 54.8, and 104.8 grams as well as errors across all trials and loads.

Figure 12 presents the mean and maximum percentage errors in the magnitude of the measured loads across three repetitive trials as well as the percentage errors across all trials and loads. Although the magnitude of the errors increased as the applied loads increased, the percentage of the maximum errors decreased as the load increased. The reason can be attributed to the errors caused by pre-loading or temperature effects in the strain gauges. These potential effects have the same kind of effects for all loads and cause similar magnitude of error for all loads. This results in a smaller error percentage for larger loads. The results do not present a significant difference between the mean percentage errors for different loads for the magnitude.

The experimental results presented the mean errors of 1.26 grams and 2.08 deg for the magnitude and direction of the applied lateral tip/tissue loads, respectively. This demonstrates the capability of the proposed instrument in measuring sideways interaction forces with accuracy far below the level of accuracy expected in delicate surgical operations [28]–[30]. It should be noted that the results listed in this paper are the preliminary results from the developed prototype and further experiments and analyses are required to study dynamic behavior and transient effects and precisely characterize the force sensing capability of the instrument. Given the advances in the field of strain analysis including strain gage installation and measurement, the load measurement capability of the proposed instrument can be significantly improved in the final product ready for the operating room.

V. STERILIZATION

The material of the force sending sleeve used in the developed instrument is chosen to be stainless steel. Grooves are made along the sleeve axis to accommodate wires of the strain gauges for protection purposes. Given the material and design of the sleeve, the matter of sterilizability comes down to choosing the right material used in application, connection, and protection of the strain gages. Studies have shown that autoclave sterilization is possible for strain-gauge instrumented devices providing the right choice of cables and connectors, as well as an optimal combination of strain gauge adhesives and coatings to withstand 121° Celsius at 207 kPa and 100% humidity for 30 minutes during the autoclave sterilization [16], [31]. The results from these studies showed that Henkel Loctite M-3981 adhesive in combination with Henkel Loctite M-11FL coating provides sufficient protection to allow the strain gauges to survive at least 5 autoclave sterilization cycles with excellent performance. For cables, teflon coated, multi-stranded bare copper conductors, with a gold-plated copper braided shield cables from Cooner Wire, model number CZ-1223-4 with a nominal outer diameter of 0.82 mm rated to 200° Celsius are good example of commercially available cables capable of withstanding autoclave serialization process. There are also commercially available autoclave-compatible connectors rated to 200° Celsius, e.g. from Fischer Core Series, model number 1031A019-130 for the connector and K1031A019-130 for the corresponding receptacle. The current design of the developed instrument can accommodate the autoclave-compatible adhesive, coating, cable, and connectors. However, since these specifications are not considered in the exact instrument presented in this paper, it is not considered sterilizable in its current state.

VI. CONCLUSION

A modular laparoscopic instrument was developed that is capable of measuring tip/tissue lateral interaction forces. The instrument is able to operate and sense the normal loads at the tip jaws without using any actuator or sensor at the tip. The modularity feature of the instrument enables the operator to quickly change the tip functionality, e.g. cutter, grasper, and dissector, without loss of control and force measurement capabilities. The specially designed calibration module and procedures provide easy and quick calibration of the strain gauges. Experiments were conducted to evaluate the force sensing feature of the instrument for lateral tip/tissue interaction forces. The results presented good accuracy and performance and verified the ability of the instrument to measure both the magnitude and direction of the applied lateral forces at the tip. The modularity and force sensing features of the proposed instrument enables the device to be employed in the haptically enabled robotic surgical system (HeroSurg) to restore the sense of touch in robotic surgical systems.

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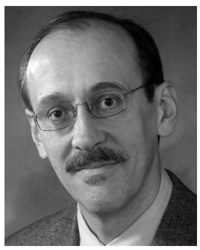

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