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Semi-Autonomous Laparoscopic Robotic Electro-surgery with a Novel 3D Endoscope*

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Abstract

This paper reports a robotic laparoscopic surgery system performing electro-surgery on porcine cadaver kidney, and evaluates its accuracy in an open loop control scheme to conduct targeting and cutting tasks guided by a novel 3D endoscope. We describe the design and integration of the novel laparoscopic imaging system that is capable of reconstructing the surgical field using structured light. A targeting task is first performed to determine the average positioning error of the system as guided by the laparoscopic camera. The imaging system is then used to reconstruct the surface of a porcine cadaver kidney, and generate a cutting trajectory with consistent depth. The paper concludes by using the robotic system in open loop control to cut this trajectory using a multi degree of freedom electro-surgical tool. It is demonstrated that for a cutting depth of 3 mm, the robotic surgical system follows the trajectory with an average depth of 2.44 mm and standard deviation of 0.34 mm. The average positional accuracy of the system was 2.74 ± 0.99 mm.

I. INTRODUCTION

Kidney sparing partial nephrectomy (PN) surgery is now the accepted standard of care for most localized small renal masses. In the USA, the overall age-adjusted incidence rate of kidney cancer per 100,000 person-years increased from 7.1 cases in 1983 to 11.7 in 2006, primarily owing to the widespread use of diagnostic imaging [1]. The volume of preserved kidney after partial nephrectomy fundamentally determines postoperative renal function [2], [3]. Thus, it is important to minimize the amount of normal kidney tissue excised during nephron-sparing surgery, and also to minimize vascular compromise. Another key factor determining the renal function after surgery is the duration of induced ischaemia caused by clamping of the renal arteries during the surgery. Reported positive surgical margins range between 1.0 % of specimens from patients who underwent the open procedure to 1.6 % of laparoscopic partial nephrectomy specimens [4] and 0–5.5 % in robot assisted surgery (RAS) [5].

Compared to open PN, laparoscopic PN improves the recovery rate and satisfaction of patients [6]. However, laparoscopic PN requires highly skilled surgeons to perform precise procedures in a confined space compounded by the limited maneuverability of instruments [7]. Limitations of maneuverability result in imperfect intra-corporeal suturing, excision, and consequent postoperative complications. Recent advances in robotic technology has enabled surgeons to perform PN more frequently [8]. This is achieved because the articulated robotic surgical instruments enable movements with higher degrees of freedom (DOF) and precision compared to conventional laparoscopic PN [9]. Furthermore, the higher precision and shorter learning curves for the surgeons via robotic PN reduces excisional volume loss (EVL) of healthy tissues compared to open and conventional PN [10], [7]. In some cases, prolonged robotic PN surgeries can lead to postoperative complications [11] due to prolonged induced ischemia. Although advances in minimally invasive surgery (MIS) and tele-operated RAS have reduced collateral tissue injury associated with an open surgical approach, complication rates have not improved [12], [13]. Semi-autonomous and autonomous RAS may improve patient outcomes and decrease complications rates by giving all patients access to best surgical techniques independent of surgeons experience and training. Autonomous or semi-autonomous RAS could potentially improve surgical margins, reduce EVL, and decrease surgery times and therefore improve postsurgical kidney function and reduce complications. In this paper we describe a laparoscopic robotic system capable of autonomous tissue electro-surgery utilizing a novel structured light 3D endoscope, shown in Fig. 1. To our knowledge this is the first time a robotic system performs semi-autonomous laparoscopic electro-cautery.

II. PRIOR WORK

A. Robotic Surgery

The advantage of autonomous robotics has been well-established in the medical field [14]. Limited autonomous RAS has been introduced in bony orthopedic procedures [15], radiotherapy [16], and cochlear implantation [17]. When considering soft tissue surgeries, challenges from blood occlusion, char, and tissue deformation make autonomous surgical tasks more challenging. For autonomous cutting tasks, one surgical robot is the RAVEN surgical system. Using stereo imaging and low level motion planning, the robot can detect the pose of surgical graspers to perform multilateral debridement of phantom tissues [18]. This work was expanded to include the da Vinci Research Kit (dVRK) [19]. Using learning by observation and ultrasound guidance, the dVRK performed surgical debridement, detection of circular targets, completed circular cutting tasks [18] as well as completed autonomous dissection of cryogel phantoms [20]. When outfitted with custom surgical tools and automated tool changing, the dVRK autonomously detected, mapped, and removed pseudo tumors from phantom tissue [18].

Alternative technologies include the use of the RAVEN II surgical system to detect and ablated pseudo neuroblastoma with stereo tracking [21], and the robotic platform Probot which performed repetitive cutting motions to aid in the removal of prostate tissue [22]. While each solution demonstrates proof of concept for autonomous cutting, they assume cutting tasks can be completed in a single pass, and are limited to phantom tissues that do

not adequately address motion, deformation, textural similarity, or tissue char observed in cadaveric tissue.

The first robot to address the challenges of tissue motion, and deformation to complete a surgical task in soft tissues was the Smart Tissue Autonomous Robot (STAR). By incorporating a 7-DOF lightweight robot (KUKA LWR 4+), a plenoptic and NIR imaging system, motion planning software, and robotic tool sets, STAR tracked a deformable surgical environment and performed semi-autonomous intestinal anastomosis in soft tissue with more consistency and quality than surgeons using open, laparoscopic, or RAS techniques in phantom and ex vivo tissue [23], [24]. STAR was used successfully in a pre-clinical study to perform intestinal anastomosis in four porcine models [24]. Most recently, STAR was paired with NIR imaging to detect pseudotumors in porcine cadaver tissue, performed line cutting tasks with more accuracy and less deviation than expert surgeons, and autonomously resected pseudo tumors from porcine cadaver tissues with clinically acceptable margins using visual servoing [25].

B. Three-dimensional Surgical Imaging

The 3D detection for STAR was previously implemented in an open surgical setting using a commercial plenoptic camera. In this work, we extend our navigating scheme to support minimally invasive setting, therefore, an imaging system with endoscopic view is desired. Efforts in 3D imaging for minimally invasive surgical applications are studied using approaches such as stereoscopy, time of flight, structured illumination, and recently plenoptic imaging [26], [27].

The stereoscopic technique is based on searching of stereo correspondences between two distinct but overlapped views of the object. Using the triangulation principle between the cameras, the correspondences are found to reconstruct a disparity estimation of the studied site. The development of stereoscopic is extended to various feature-based reconstruction optimization to compensate for the high computation cost of correspondence searching. Some examples are the hybrid recursive matching [28], [29], the seed propagation method [30], and the cost-volume method [31], which lead to depth resolution of up to 50 μm .

The time-of-flight (ToF) method traces the time difference between the emitted light and the reflected light from the object, thus calculating the surface structure of the object. Therefore, ToF system can be compact, however, the technique depends on the quality of the return rays, which is highly impacted by biological properties of the sample and the sensor performance. Depth resolution using ToF in surgical endoscope ranges from 0.89 mm [32] to about 4 mm [33].

Plenoptic imaging is an extended method of stereoscopy using multiple collected views of the site based on focused images through a microlens array, instead of using two cameras as in the conventional stereoscopy. Hence, the disparity search is faster with the defined geometrical configuration of the microlenses. Moreover, the system is compact by using only a single camera, a microlens array, and a focusing main lens. The development of plenoptic imaging system in endoscopic settings is rather new with current depth and axial resolution in the order of more than 1 mm [34], [35].

Structured illumination (SI) also provides depth quantification using triangulation from the parallax angle between two devices. However, instead of using multiple cameras as in stereoscopy, SI searches for structural information based on phase variations of the object. Efforts using SI in endoscopic settings resulted in augmented reality visualization [26], tubular tissue scanning using a modified ring projection producing an average dimensional error of 92 μm [27], and multi spectral spot projection for motion detection that can achieve up to 50 μm [28]. The low light condition is compensated by using a high power light source such as super-continuum as in [29] or laser in [28]. In this study, we develop and test a 3D endoscopic system with lowpower LED-based illumination and optimal parallax angle of approximately 10 degree shown in Fig. 2. The parallax angle is determined experimentally to achieve the most overlapped field of view between the projecting and imaging scopes, while keeping the distance between the two scopes reasonable for MIS. A previous study had demonstrated this angle relation to the system accuracy error [36]. The depth of field of the system is reported as 20 mm with depth accuracy of static objects of about 200 μm [30]. The system is one of the first to demonstrate its capability to reconstruct 3D surfaces of dark biological samples such as porcine kidney and, together with robotic registration, provide surgical navigation.

III. METHODS

A. Scope and Contributions of the Paper

In this work, STAR performed semi-autonomous targeting and cutting tasks on porcine cadaver kidney tissue. The design and integration of a novel laparoscopic imaging system to reconstruct three dimensional point clouds of cadaveric tissue, and to control the location of an electrosurgical tool is first discussed. The positional accuracy of the system is evaluated via targeting tasks, where STAR is used to locate and mark six fiducials on porcine kidney models using the imaging system. The accuracy is reported as the average xy norm distance between the centroids of each fiducial and their corresponding mark.

After performing targeting tasks, the laparoscopic imaging system reconstructs the surface of a porcine kidney, and generates a linear surgical cutting plan for STAR to follow. STAR then cuts along the trajectory with a predetermined depth in a simulated minimally invasive procedure. The depth accuracy of the system is evaluated by measuring the average minimal distance between the cut depth and kidney surface.

B. Multi Degree of Freedom Electrosurgical Tool

A custom two degree of freedom electro-surgical tool is coupled to the mounting flange of the KUKA IIWA [25] to perform semi-autonomous cutting tasks. As illustrated in Fig. 3, the electro-surgical tool incorporates a multi-axis motor pack, universal tool base, and quick release electrode holder so the specific cutting electrode for a surgical task is used. Two motors control the pitch and roll of the cutting electrode. The pitch and roll motions provide two extra DOF within the patient, so that the tool can reach any target in the surgical field after it is placed through a laparoscopic trocar. Each motor includes an encoder for precise positioning using EPOS2 controllers (Maxon Motors, Sachseln, Switzerland).

Communication between the robot and tool is through a controller area network (CAN). The

tool is homed by energizing the pitch motor with current control until it reaches a hard stop. The cutting tool is powered by an electro-surgical unit (DRE Medical, Louisville, KY) capable of delivering 0–300W. In the following experiments, a 60W cutting waveform is used to mark and cut cadaveric tissue with an insulated needle electrode.

C. Endoscopic Imaging System

To record surgical scenes in three dimensions for robotic navigation, we developed the endoscopic imager based on structured illumination (SI) technique [30] illustrated in Fig. 2. Currently, SI endoscopy in medical intraoperative inspection has been rather limited due to several reasons. Because of the restricted working volume in the endoscope, many SI endoscope designs do not consider the parallax angle between the illumination and imaging ports, but using the same optical path between them by a beam-splitter, making the depth accuracy non-uniform and the field of view narrowed [37], or providing perceptual depth measurement [26], [38], making it incompatible for robotic application. In addition, due to the high optical attenuation in endoscopic setting on realistic tissue, current approach for SI utilizes high power light source to illuminate a coded pattern on the tissue and deduce height values from a hybrid technique similarly to stereoscopy, making the collected 3D point cloud distribution lossy for tissue surfaces with low features [38]. Here we design an endoscopic system with a unique vertical gradient pattern for 3D tissue reconstruction, where the discontinuity of the object gap is constructed from phase information calculated by multiple shifted patterns from low to high frequencies. To our knowledge, this demonstration is the first quantified 3D of a dark tissue surface such as kidney in an endoscopic setting. The system is composed of two zeroangle rigid boroscopes (5mm-diameter, Karl Storz GmbH & Co. KG, Tuttlingen, Germany) for pattern illumination and imaging. The pattern projection is generated by a digital light projector equipped with an internal LED-based optical chamber. The projected patterns sequentially create shadows on the studied object. The concavity and convexity of the collected shadows are consecutively relayed and focused to the imaging endoscope, an achromatic doublet lens (AC254060-A, Thorlabs, Newton, New Jersey, USA), and a chargecouple-device (GS3-U3-15S5M-C, Point Grey Research Inc., Richmond, Canada). The 3D endoscopic imager is mounted on an optical plate attached to a positioning arm. Prior to registration with the robot, the 3D imager system is adjusted so that the plane containing the two scopes is perpendicular to the defined zero plane for camera-projector calibration. The calibration process is adapted from [31] using multiple positions of a printed chessboard to deduce the extrinsic and intrinsic camera parameters in relations with the digital projector. After the camera-projector calibration, the position of the 3D system is adjusted at an optimal angle and distance from the sample's plane to allow robot access. The position of the scopes allows the most overlapped region between the fringe projection and sample imaging, and the entire imaging system is aligned at about 60 degrees and about 5cm way from the sample plane to give enough space for the robot to maneuver. The depth calculation is described in the following steps:

Step 1: Generating vertical fringes I_f digitally for the projection. The generated pattern follows Eq. (1) [32]:

$$I_i(u,v) = I_o(1 + \cos(2\pi ku/w + \delta)) \quad (1)$$

where I_o is amplitude of the modulated fringe, k is the fringe number, (u, v) is the spatial pixel indices, w is the pattern width and δ is the shifted phase.

For each 3D reconstruction, a collection of 16 I_f fringes (4 sets of fringe number with 4 sets of shifted phase) is generated. The speed of 3D reconstruction mainly depends on the camera sensitivity to image tissue surfaces. In our case of imaging kidneys, the camera exposure is maintained at around 50 ms, resulting in 800 ms (1.25 Hz) for the entire fringe set.

Step 2: Calculating phase. The captured fringed images are used to compute 4 sets of wrapped phase ϕ_n^w using the conventional four-step shift method [39]. The phase is then unwrapped from its principle range of 2π following Eq. (2)

$$\phi_n = \phi_n^w + 2\pi < (\phi_{n-1} k_n / k_{n-1} - \phi_n^w) / 2\pi > \quad 2$$

where n is the order of the selected frequencies ($n = 2, 3, 4$ and $\phi = \phi_1^w$) and $< . >$ denotes the rounding up to the nearest integer value.

Step 3: Calculating height. The sample height z at each pixel index is produced from the unwrapped phase with highest frequency as described in Eq. (3), where ϕ is the unwrapped phase distribution of the highest frequency, as calculated in Eq. (2) and $\phi = \phi_4$, c_{0-11} and d_{0-11} are resultant parameters from the camera calibration. The camera-projector calibration procedure is detailed in [32].

D. Calculation of Cutting Trajectory and Surface Normal

Reconstruction of the 3D surface based on point clouds of the kidney surface with the white markers provided by the imager and calculation of the cutting path along the surface is performed as follows:

Step 1: Post-processing of the 3D point cloud. The statistical outlier removal module in the Point Cloud Library (PCL) is used to remove noise in the point cloud. The module computes the mean μ and the standard deviation σ of the distance of K -nearest neighbor points, and removes all points outside $\mu + \alpha\sigma$. In our experiment, we selected $\alpha = 1$ and $k = 50$ [2]. In addition, a moving least square (MLS) algorithm was implemented to create a smooth point cloud distribution [40], [41]. The MLS calculates a fitting surface by higher order polynomial interpolation among points selected in a sphere radius r and then resamples missing points on the surface, creating a smoothed distribution of points. We selected a sphere radius of $r=4$, to create a smooth surface while maintaining details Fig. 4 (a).

Step 2: Reconstructing the surface. The surface of the point cloud was reconstructed by creating 2-D Delaunay triangulations in MATLAB (MathWorks, Natick, Massachusetts, USA). After completing Delaunay triangulation, a trisurf function was applied to create a triangular surface plot Fig. 4 (b).

Step 3: Generating a cutting path on the surface, i.e. the trajectory line. The trajectory, shown in red line, was created by using Dijkstras algorithm to calculate the shortest cutting path along the surface between two selected points Fig. 4 (c). The trajectory consists of node sets which contain position information given by the point clouds.

$$z = \frac{c_o + c_1\phi + (c_2 + c_3\phi)u + (c_4 + c_5\phi)v + (c_6 + c_7\phi)u^2 + (c_8 + c_9\phi)v^2 + (c_{10} + c_{11}\phi)uv}{d_o + d_1\phi + (d_2 + d_3\phi)u + (d_4 + d_5\phi)v + (d_6 + d_7\phi)u^2 + (d_8 + d_9\phi)v^2 + (d_{10} + d_{11}\phi)uv} \quad (3)$$

Moreover, in order to ensure the end-effector would always be normal to the surface while moving along the path, we calculated the normal vector along the cutting path (Fig. 4 (d)) as follows:

$$\vec{n}_{path} = \frac{\vec{P}_{nk} \times \vec{P}_{nn+1} + \vec{P}_{nn+1} \times \vec{P}_{nm}}{|\vec{P}_{nk} \times \vec{P}_{nn+1} + \vec{P}_{nn+1} \times \vec{P}_{nm}|} \quad (4)$$

where $\vec{n}_{path}$ is the unit normal vector along the cutting path $\vec{P}_{nn+1}$. The average between the cross products of vectors $\vec{P}_{nk}$ with $\vec{P}_{nn+1}$ and $\vec{P}_{nm}$ with $\vec{P}_{nn+1}$ form the normal vectors along the trajectory as shown in Fig 5. The shortest cutting path calculated by Dijkstras algorithm was compared with distances between start and end point on kidneys measured manually using a ruler, verifying the output of the algorithm.

The computation time for each step was 4.14s, 6.01s, and 6.41s, respectively. The results were obtained with an i76500U processor (Intel, Santa Clara, California, USA).

E. Control Method

The heterogenous hardware of the STAR includes a KUKA IIWA that is controlled by the ROS IIWA stack [33] and an actuated cutting tool that is controlled by an Orocos RTT component [42]. At a higher level, both devices are controlled from another Orocos component that agglomerates the telemetry, computes inverse kinematics and dispatches joint commands to the devices as illustrated in Fig. 6. Cartesian trajectories are computed inside that component by using the Reflexxes library [43]. Hand-eye calibration between the endoscope and the robot is achieved by mounting a pointing tool on the IIWA and touching a series of fiducial markers by slowly jogging the robot. These markers are then registered in

the 3D point cloud and the eye-to-hand transformation is computed by using a least squares solution [44]. In these experiments, we used four coplanar fiducials markers and the hand-eye calibration procedure takes approximately 10 minutes to execute.

IV. EXPERIMENTS AND RESULTS

A. Experimental Setup

For all experiments, the KUKA IIWA was mounted to the top of a steel cart, 60 cm from the edge, to enable simple kinematic configurations of the robot and tool. A plastic torso was used to mimic the human body and a porcine cadaveric kidney was placed on an electrosurgical unit grounding pad that was fixed to a pedestal located between the T12 and L3 vertebrae and 5 cm below the surface of the back. A retroperitoneal approach was chosen for the procedure as it is standard practice for PN [45], [46]. The torso was screwed onto two parallel T-shaped aluminum extrusions which were attached to each leg of the steel cart. The body was placed 30 cm from the surface of the cart and the aluminum extrusions provided height adjust-ability of ± 15 cm.

B. Targeting Test Description and Results

The kidney was marked with six dots white dots in a 3×2 matrix, where rows and columns were separated by 5mm and 10mm respectively as shown in Fig. 7. The imaging system captured the three dimensional coordinates of each mark, which were passed to the IIWA controller. These coordinates are calculated as the centroids of each mark and are detected by using intensity thresholding and k-means clustering. The KUKA LWR was outfitted with the electro-surgical cutting tool, and commanded to each target location within the surgical phantom. A cutting waveform of 60W was applied at each commanded position for 5s to mark the tissue.

The targeting accuracy of the system was defined as the xy norm between centroids for each target and corresponding burn. Centroids for the targets (black) and burns (yellow) were determined using the imaging software, ImageJ (National Institutes of Health, Bethesda, MD) and are shown in Fig.7. Centroid locations were calculated by converting an RGB image of the surgical scene to gray scale, applying the default thresholding operation, and using the built in center of mass measurement tool. The center of mass for each fiducial illustrated Table I details the x-y norm for each targeted position. The average targeting accuracy of the system was 2.74 ± 0.99 mm standard deviation. The standard of care for tumor margins in nephron sparing surgery is reported as 10 mm [47], with more recent studies demonstrating that 5 mm is just as effective [48]. Following a theoretical margin of 10 mm, STAR's resected tumor margin would range from 12.74 mm and 7.26 mm which is more than two standard deviations away from the minimum margin of 5mm, meaning STAR's accuracy is suitable for PN.

C. Cutting Test Description and Results

Cadaveric porcine kidney was prepared and placed within the torso phantom as described above. The kidney was randomly marked at two locations 2 cm apart from one another. The imaging system was used to reconstruct a three dimensional point cloud of the kidney

surface and to extract the centroids of each marker points in similar way as described in IV-B. Using MATLAB, the trajectory that followed the minimum distance between the two selected centroid points was determined, and a surface normal was calculated for each trajectory point as shown in Fig. 4. The trajectory and surface norms were then passed to the IIWA controller, and the KUKA LWR was used to cut along the trajectory at a depth of 3 mm using the electro-surgical tool to apply a 60 W cutting waveform.

The depth accuracy of the system was defined as the commanded depth minus the average minimum distance between the surface and cut profile in the kidney. To measure the depth accuracy, the kidney was dissected along the electro-surgical cut using a scalpel. Figure 8 illustrates a cross sectional RGB image of the cut edge along with the process of calculating the depth. The kidney slice image is manually marked to highlight the cut region (step b). Python code developed via OpenCV is used to crop the image, convert it to gray scale, apply gradient thresholding to detect the region edges (step c), and finally find the depth via calculating the distance between the top and bottom edges (step d). The depth accuracy of the system was found to be an average of 2.44 ± 0.34 mm standard deviation. Similarly to the x-y targeting accuracy, the depth accuracy of STAR is suitable for PN. When following a theoretical margin of 10mm, we would expect STAR's depth margin to range from 12.44mm to 7.56mm, a distance more than seven standard deviations away from the minimum clinical margin of 5mm.

V. DISCUSSION

STAR is the first robot capable of semi-autonomous surgical tasks in soft tissue. By incorporating a novel laparoscopic imaging system capable of three dimensional reconstruction, STAR can perform clinical tasks in a simulated surgical environment. Although initial targeting accuracy was not sub-millimeter, the system successfully performed a cutting task in porcine cadaver tissue within a simulated surgical phantom.

One limitation of the work is positional accuracy of the system. For these tasks, STAR is operated in an open loop control scheme where target locations and cutting trajectories are calculated from the imaging system and streamed to the LWR. The open loop system is more prone to positioning error, due to the propagation of errors in camera registration and calibration. Despite best efforts to minimize this error, the average positional accuracy of the STAR system is 0.50 ± 0.20 mm [23]. The additional error observed in this experiment is most likely attribute to backlash within the gearing of the distal tip of the tool, which is magnified by the length of the needle electrode. To minimize this error, a closed loop control scheme will be implemented as detailed in previous work using STAR to autonomously reset pseudotumors for porcine tissue.

Secondly, the system did not consider the effects of tissue motion or soft deformations. This limitation is the result of the integration between camera, surgical planner, and LWR. While the laparoscopic imaging system can reconstruct the surgical scene in real-time, the work flow to calculate surface normals for a cutting trajectory has not been optimized for time yet. In previous work, a plenoptic imaging system was used to generate surgical plans that were streamed directly to the LWR [24]. In future iterations, a similar strategy will be

implemented with the laparoscopic imaging system, so STAR can detect and react to deformations in real time.

Thirdly, the 3D depth quality depends strongly on the initial camera-projector calibration for the 3D imaging endoscope. The calibration is done by varying multiple positions of a printed chessboard within the endoscopic view, where the first position is regarded as the zero-reference plane and is positioned normal to the longitudinal axis of the endoscope. In that, the first position should be carefully adjusted and a high number of calibration position (>10) is desired to cover the field of view and correct for the radial distortion of the surgical scope. Moreover, the current 3D imaging system utilizes two separate rigid scopes, which limits its use in applications with restrained access. Therefore, a more compact design, which combines the two scopes into one when maintaining the triangulation quality is important and can be achieved by using flexible imaging fiber as the relay optics.

Finally, the excessive length of the distal joint of the cutting tool limits autonomous cutting routines to superficial tissue and organs. When a commercial needle electrode is used with the cutting tool, the length of the distal joint is 9.42 cm. When used with a remote center of motion, the excessive length limits access to deep regions such as the pelvis. To expand the use of the STAR to other minimally invasive procedures, the quick connect interface and electrode length should be shortened as far as possible. The resulting geometry would substantially improve the kinematics of the system, and allow for additional cutting motion within deep structures.

Despite study limitations, this work has the potential to transform clinical care for patients in need of partial nephrectomy. The integration of a laparoscopic structured light camera, multidegree of freedom electro-surgical tool, and open loop control scheme demonstrate the first successful cutting task on soft tissue within a simulated minimally invasive procedure. Additionally, impact to the clinical workflow is minimal as the setup and location of the kidney phantom is physiologically correct, and all interventions were performed with laparoscopic access. The current cutting strategy relies on an open loop control strategy to follow cutting trajectories. Prior to clinical implementation, a closed loop cutting strategy should be implemented to minimize targeting and cutting error.

VI. CONCLUSION

This paper reports the design and first evaluation of a laparoscopic robotic system for electro-surgery. Using a novel structured light endoscope, the system achieved an open-loop positioning accuracy of 2.74 mm and when commanded to cut at a consistent 3 mm deep trajectory long the kidney surface, achieved an average cutting depth of 2.44 mm. Successful completion of targeting and cutting tasks using laparoscopic constraints demonstrate feasibility of this technology to complete semi-autonomous electro-surgery in minimally invasive procedures.

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Fig. 1.
Picture of the laparoscopic robotic system with 3D endoscope

Fig. 2.
3D endoscopic camera mounted on a positional arm

Fig. 3.

A multidegree of freedom electrosurgical tool for semiautonomous electrosurgery. Pitch and roll actuations are controlled by a six axis motor pack, the the distal tip contains a quick connect interface that is compatible will standard electrodes.

Fig. 4. Surface reconstruction and path planning for cutting task: a) Postprocessing of point cloud, b) surface creation by Delaunay triangulation, c) cutting path calculation along the surface, and d) normal vector generation along the cutting path.

Fig. 5.
Average normal vector along the trajectory

Fig. 6.
Software architecture of the STAR

Fig. 7.

Pattern of target fiducials (T1-T6, white) and burned marks (B1-B6, black) that were placed on cadaver porcine kidney tissue during targeting accuracy tests. Targeting error was calculated as the average norm distance between centroids for fiducials (black) and their corresponding burned mark (yellow).

Fig. 8.

Steps of analyzing the depth of cut: a) slicing the kidney along the cut, b) marking and cropping the cut region, c) detecting the edges of cut, d) calculating the depth via the distance between top and bottom edges.

TABLE I**RESULTS OF THE TARGETING ACCURACY TEST**

#	Target Fiducial		Burned Mark		Error (mm)
	X (mm)	Y (mm)	X (mm)	Y (mm)	
1	8.94	26.43	10.38	27.45	1.76
2	20.02	27.41	22.68	30.73	4.25
3	10.13	20.65	11.86	20.21	1.78
4	21.88	20.73	24.6	22.79	3.40
5	11.42	14.83	13.18	13.43	2.25
6	23.17	13.94	26.18	13.78	3.01