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A robotics-based flat-panel Ultrasound Device for continuous intraoperative transcutaneous Imaging

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***Abstract*—Laparoscopic partial nephrectomy has become more and more popular in the last decade. Video laparoscopes remain the gold standard of intraoperative imaging during laparoscopic interventions. However, providing only superficial images of the target tissue. In contrast, ultrasound (US) imaging may offer crucial information of the interior of the target tissue that could improve surgical outcome.**

In this paper, we propose a new concept and prototype system to manipulate an US-probe during laparoscopic partial nephrectomies. Our primary goal was to provide the surgeon with US-images during the intervention in real-time. The prototype system consists of three components: a conventional US-machine, a manipulator to guide the US-probe, and a joystick console to control the manipulator. The results of our experiments show that the concept is feasible for US-imaging during laparoscopic partial nephrectomy.

I. Introduction

In Germany, there were approximately 16,500 renal tumor incidences in 2006 [1]. In the case of renal cell carcinomas, new imaging technologies have led to increased detection of small lesions and therefore facilitated a shift from radical to partial nephrectomy [2]. Specifically, laparoscopic partial nephrectomy has emerged as a viable intervention minimizing patient morbidity [3]. In the beginning, laparoscopic partial nephrectomy was limited to patients with small, superficial, and solitary tumors. However, increasing laparoscopic experience facilitated the resection of larger, central, and hilar tumors [2]. Nowadays, almost every urologic operation can be performed laparoscopically just as efficiently, and with fewer and less serious complications, than with conventional open surgery [4].

Intraoperative imaging during laparoscopic interventions is mainly performed by video laparoscopes allowing only for monitoring of the organ's surface. Therefore, it is impossible to detect the boundaries of a tumor within an organ as well as internal blood vessels intraoperatively. Valuable information on whether the surgeon works in a safe layer of dissection or whether a structure can be spared or needs to be removed is missing and could improve surgical outcome [5].

Ultrasonography can overcome this limitation intraoperatively by providing an internal view of the target tissue. The major advantages of ultrasound (US) are [6]: a) its minimal-invasiveness to the patient and the surgeon while providing images from within the patient, due to the absence of harmful radiation; b) its compactness, mobility, and relatively low costs resulting in a wide availability; c) its real-time capabilities that may enhance the surgeon's intraoperative decision-making capacity.

The de facto standard for intraoperative use of US during nephron sparing laparoscopic surgery has been described by Gill et al. [3]. They employ laparoscopic US-probes that are inserted through regular laparoscopic trocars and deposited on the target organ. Gill et al. use US primarily for intraoperative staging. The navigation of the probe is indirectly performed through the images of the video laparoscopes [7]. Accumulated experience has already confirmed the essential benefits of intraoperative US during difficult resections of renal carcinomas with minimally invasive surgery techniques [8], [9]. However laparoscopic ultrasonography has shortcomings, namely that: a) it provides little overview over the surgical field due to its close contact and the small size of the transducers; b) one hand of the surgeon is occupied with steering the probe, leaving only one hand to perform surgical tasks [7]; c) the eye-hand coordination is demanding while controlling the probe since the clinician sees the probe only indirectly through the video laparoscope [10].

An alternative US-imaging approach are robot-guided transcutaneous US-probes. These systems allow for telematic control of the transducer by the surgeon. These robots may overcome the following shortcomings of laparoscopic US-probes: a) Transcutaneous probes are applied from outside on the skin of the patient. Therefore, they provide a better overview over the surgical field since they are operated at a further distance leaving more room to spread for the US-waves. b) The robot guides the US-probe leaving two hands for the surgeon to operate. Multiple approaches have been presented to guide transcutaneous US-probes. Most prominent are the systems by Salcudean et al. [11], Pierrot et al. [12] and Delgorge et al. [13]. These robots have in common that they employ conventional US-probes. Salcudean et al. proposed a backdrivable kinematics consisting of parallel linkages that has seven degrees of freedom (DOF). The manipulator is mounted next to the table on which the patient is lying. It was designed for examinations of the cartoid arteries at the throat of the patient [11]. The "OTELO" robot described by Delgorge et al. has 6 DOF, a remote center of motion and is directly placed on the abdomen of the patient while in use. During

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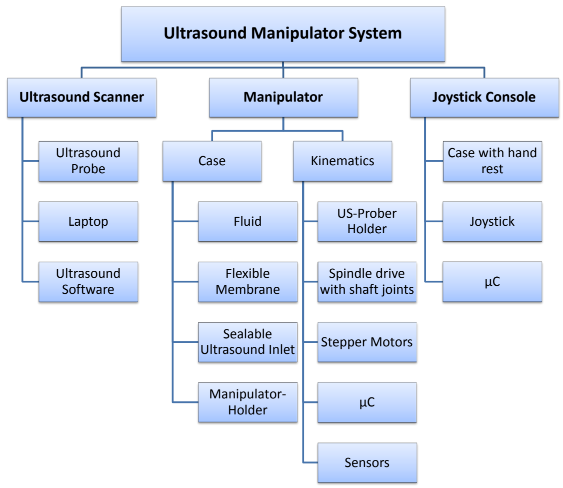

Fig. 1. Static system description of the ultrasound manipulator system

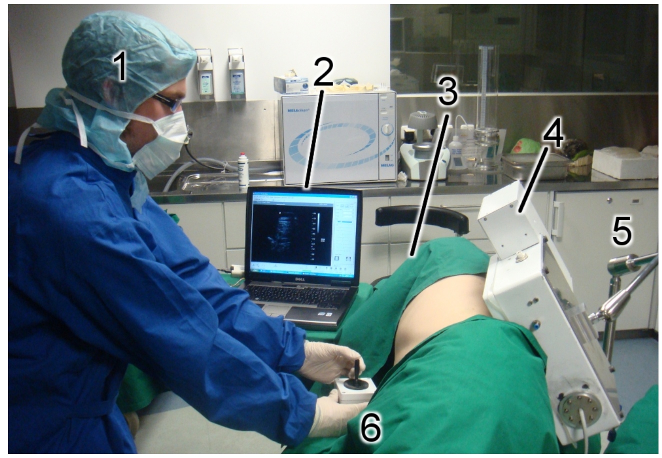

Fig. 2. Exemplary intraoperative setup: 1) surgeon; 2) laptop to visualize the ultrasound images; 3) patient; 4) ultrasound manipulator; 5) holding mechanism.

the examination the system must be balanced by the patient or kept hold of by a second person [13]. The "HIPPOCRATE" robot was presented by Pierrot et al. with the goal to quantify the volume of atheromatous plaque. It has six DOF and is mounted to a rigid base frame. However, none of the proposed systems is designed for intraoperative use and none is feasible to support laparoscopic partial nephrectomies. Additionally, neither have the manipulators a sterilization concept, nor can they be operated by the surgeon alone.

In this paper, we now report a new concept to manipulate an US-probe during laparoscopic partial nephrectomy. Based on the drawbacks of the state of the art we defined a specific set of goals that we want to achieve. First, we want to build a system that is easy to use, i.e. it shall: a) permit the surgeon to keep on operating while US-imaging is performed; b) provide the surgeon with an overview of the surgical field; and c) be directly and sterile controllable by the surgeon. Second, no further workforce shall be needed besides the surgeon to operate the manipulator. Third, the system shall be intraoperatively applicable for laparoscopic renal interventions.

II. Material and Methods

We designed a new concept of a system to fulfill the set requirements. The concept will be described in the following.

A. Static System Description

The system should have three components: 1) an US-machine, 2) a manipulator to actuate an US-probe, and 3) a joystick console for the surgeon to control the manipulator (Fig. 1). The system should use a conventional US-machine that displays its images on a regular laptop. The manipulator should consist of a case in which the US-probe resides. The case should be filled with a fluid that can be traversed by US-waves. The top of the case through which the US-waves enter the patient must be penetrable by the US-waves. The top must be some kind of flexible membrane that can be adapted to the shape of the patient. The frame should have a holding mechanism to be mounted at the operating room (OR)-table. The US-probe should be actuated by a spindle drive with shaft joints providing a self-locking mechanism. The kinematics should be actuated by two stepper motor, one for each axis. Both stepper motors should be controlled by a microcontroller (μC) and powered by separate driver units. A μC should process the control commands articulated by the surgeon through the joystick. In order to be operated in sterile the console and the manipulator should be covered with sterile foils.

B. Dynamic System Description

The working position of the system is on the back of the patient who is lying on the side, as shown in Fig. 2. The surgeon controls the manipulator through the joystick console. The joystick of the console has two DOF to control the two DOF of the US-manipulator. During operation, the US-machine provides the surgeon with US-images from within the patient. This is possible since the US-waves traverse only areas they are able penetrated like the fluid in the manipulator or tissue. Preliminary experiments revealed that none of gas, inflating the abdomen cavity, resides between the kidney and the skin of the patient during laparoscopic partial nephrectomies.

C. Implementation

The implementation of this new system was based on the experiences we had made with the system proposed by Gumprecht et al. [14]. The employed US-machine (Teratech, Burlington, MA, USA) has an operating frequency of 7.5 MHz and a maximum penetration depth of 90 mm. The case is made of aluminium and filled with water. The top of the case is covered by a silicone membrane with a thickness of 1.5 mm and a hardness of shore 40A (Siltec GmbH & Co. KG, Weiler-Simmerberg, Germany). The membrane is flexible enough to adapted to different shapes of the human body. More information on the selection process of the membrane,

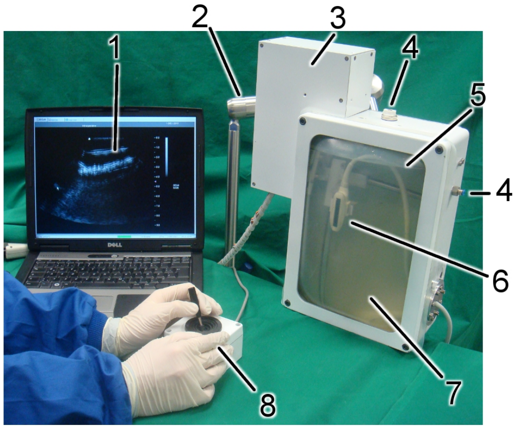

Fig. 3. Implementation of the ultrasound manipulator system: 1) Laptop to display the ultrasound images; 2) holding arm for the manipulator; 3) cover for the stepper motors; 4) valves to insert and release the water inside the manipulator; 5) water inside the manipulator; 6) ultrasound-probe; 7) flexible silicone membrane; 8) joystick to control the manipulator.

based on the imaging quality, is presented by Gumprecht et al. in a second publication at this conference. The holding mechanism for the manipulator consists of two ball joints and one revolute joint providing enough degrees of freedom to adjust the manipulator in any direction. Both DOF of the kinematics consist of a rail with a sledge that is driven by a spindle. The spindles have a pitch of 3 mm. Attached to each sledge is a rod to carry the US-probe. The stepper motors have a holding torque of 49 Ncm at a voltage of 24V and a respective current of 1 A. The angular step size of the stepper motors is 1.8°. The transmission ratio between the motors and the spindle is the same for both axis, 1:2.4. This leads to a step size of the whole system in each direction of:

$$3\,mm * 2.4 * \frac{1.8°}{360°} = 0.04\,mm \tag{1}$$

Both μC are 8-bit RISC-based μCs with an operating frequency of 16 MHz (ATMega2560, Atmel Corp., San Jose, CA, USA). Hall sensors are used as end stops to minimize the number of drills through the case. The hall sensors are placed outside the aluminium case while the triggering magnets are inside in the water. The joystick console is based on the system proposed by Maier et al. [15].

D. Experiments

We performed two experiments to assess the properties of the system. In the first experiment, we tried to verify that the system can be used for US-imaging. Therefore, we placed the US-probe in water and recorded the images of an US-phantom proposed by Seidl et al. [16]. The US-images were first recorded directly and then through the employed silicone membrane. During the experiment we used identical settings for the US-scanner, i.e. 7.5 MHz operating frequency, 9 cm penetration depth. We evaluated two areas of 100×100 pixels in 10 US-images for both setups. The first area was located

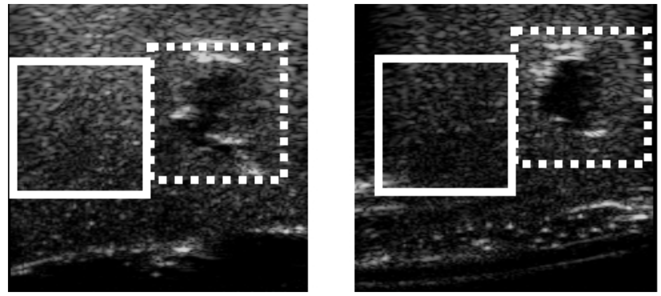

Fig. 4. Comparison of the ultrasound B-Mode images: Left US-image without membrane; Right US-image with clear visible membrane at the bottom. The square marked with a solid frame defines the homogeneous area for the measurements. The area of interest with the tumor is defined by the square with the dotted frame.

at a homogenous area within the images. The distance from the surface of the phantom was the same for all images. The second area was located at the same tumor for all recordings. For both areas we determined three parameters:

a) average grey-value: Lower grey values indicate fewer reflection. Since the areas of interest are located at the same position within the phantom lower average values indicate that less energy of the US-waves pass the membrane.

b) signal-to-noise ratio: One of the shortcoming of US is its poor signal-to-noise ratio (SNR). Therefore, we calculated the SNR to assess how it is affected by the membrane. We calculated the SNR using the following formula:

$$RMS\ noise = \sqrt{\frac{\sum_{i=1}^{n}\left(X_i - \frac{\sum_{i=1}^{n} X_i}{n}\right)^2}{n}} \tag{2}$$

$$SNR = 20\,log_{10}\frac{signal}{RMS\ noise} \tag{3}$$

X are the pixels within the area of interest. $signal$ is the average grey value within the area of interest.

c) range of contrast: is an important factor for quality of the US-images. We calculated the range of contrast using the following formula:

$$C_m = \frac{gv_{max} - gv_{min}}{gv_{max} + gv_{min}} \tag{4}$$

gv_{max} and gv_{min} are the highest and the lowest values respectively within the area of interest. In the second experiment, we measured the real step size to confirm our calculations of (1). For the measurements we employed a length gauge with a system accuracy of $\pm$0,2 μm (MT12, Heidenhain, Traunreut, Germany).

III. Results

The results of the first experiment are summarized in Fig. 4 and in Table I. The images with the membrane (*M*) are slightly darker than those without the membrane (*NM*). This can be directly seen when comparing the averages in the area of interest in both images: 95.92 (*M-Tumor*) vs. 97.80 (*NM-Tumor*), and 113.31 (*M-Homogenous*) vs. 123.65 (*NM-Homogeneous*)).

TABLE I
COMPARISON OF THE GREY VALUE OF THE US-IMAGES WITH AND WITHOUT THE SILICONE MEMBRANE AT DEFINED AREAS OF INTEREST

	Tumor area			**Homogeneous area**		
	Average (SD)	*SNR (SD)*	*Range of Contrast (SD)*	*Average (SD)*	*SNR (SD)*	*Range of Contrast (SD)*
Without Membrane	97.80 (±2.73)	22.68 (±0.36)	0.85 (±0.02)	123,65 (±4.94)	31.21 (±0.56)	0.24 (±0.04)
With Membrane	95.92 (±2.61)	22.86 (±0.68)	0.84 (±0.02)	113.31 (±5.9)	30.35 (±1.97)	0.31 (±0.08)
Change in %	-2	-0	-2.2	-8.4	-2.8	+27

The change in the values from *M* to *NM* is -2% for the tumor area and -8.4% for the homogeneous area. The SNR is the same in both tumor areas 22.86 (*M-Tumor*) vs. 22.68 (*NM-Tumor*) (±0%) but changes slightly in the homogenous area 30.35 (*M-Tumor*) vs. 31.21 (*NM-Tumor*) (-2.8%) showing that the membrane worsens the SNR slightly. The range of contrast increases in the homogenous area in the images with the membrane 0.31 (*M-Tumor*) vs. 0.24 (*NM-Tumor*) (+27%) but stays almost the same for the tumor area (0.85 (*M-Tumor*) vs. 0.84 (*NM-Tumor*) (-2.2%). The results of the second experiment are as follows. The measured real step size is 0,04 mm verifying our calculations.

IV. CONCLUSION

In this paper, we present a new concept to manipulate an US-probe during laparoscopic partial nephrectomies. Our foremost goal was to provide the surgeon with US-images during the operation in real time. Therefore, we realized a specialized system that consists of a conventional US-machine, a manipulator for an US-probe, and a joystick console to control the manipulator. The manipulator actuates the US-probe in two DOFs. The probe resides in a fluid that can be traversed by the US-wave. The US-waves are guided through a flexible membrane into the body of the patient. During operation the US-machine provides images from within the patient. In the first experiment we compared the US-images with and without the membrane with the following results. The grey-values are lowered in the images with the membrane. This effect can easily be compensated by amplifying the gain at the US-machine. The SNR as well as the range of contrast changes in the images with the membrane only slightly. Based on the results of the experiments we conclude that our system is feasible for US-imaging. The second experiment verified that the calculated step size of the system is correct. Future work will include a clinical evaluation of the system.

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