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Ultrasound Standoff Optimization for Maximum Image Quality of a Robot-Assisted Flat-Panel Ultrasound Device

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***Abstract*—Ultrasonography is a widespread intraoperative imaging modality. However, it suffers from several shortcomings e.g. its dependance on the skills of the operator for the image quality. To overcome this shortcoming, Gumprecht et al. [1] recently proposed a new robot-assisted flat-panel ultrasound device for continuous intraoperative imaging during laparoscopic tumor resection in urology. This device is integrated in the OR-table and performs its imaging through the back of a supine patient. The ultrasound probe resides in a tank, filled with a fluid that is traversable by the ultrasound waves. A flexible membrane is stretched over the tank and is in contact with the fluid and the patient. Through is flexibility, the membrane can adapt to the shape of the patient. Therefore, the membrane assures for sufficient coupling of ultrasound waves into the patient. We based the selection of the membrane and the fluid upon the quality of the ultrasound images that can be recorded with this combination. In this paper, we present the results of the experiment that lead to the standoff used in the robotic device of Gumprecht et al. [1].**

I. INTRODUCTION

Ultrasonography (USG) is a widespread intraoperative imaging modality due to the following advantages [2]: a) USG provides real-time imaging that may enhance the surgeons intraoperative decision-making capacity; b) USG employs no radiation that may be harmful for the patient or surgeon while providing images from within the patient; c) USG machines are compact, mobile, and available at relatively low cost.

On the other hand USG suffers from some disadvantages [3], namely: a) its signal-to-noise ratio is often low; b) the presence of different kinds of artefacts, in particular enhancement, shadowing, and reverberations; c) its dependance on the operator for high quality images. Trained skills and experience are required for the acquisition of good-quality images, for making accurate diagnoses, and for the completion of precise interventions.

Several robotic approaches have been proposed to reduce the dependance on the operator for the quality of the ultrasound (US)-images while. These approaches can be divided into two groups. The first group was designed for tele-echographic remote examinations in real-time to overcome the need of an on-site expert. More precisely, the expert can remain at some distant office with a master manipulator and perform distant examinations on arbitrary places all over the world through a slave manipulator. The second group of robots was designed for on-site support of US users to improve the accuracy of the examinations or to avoid work related injuries.

The most cited concepts for remote echographic diagnoses were proposed by Delgorge et al. [4], Vilchis et al. [5], Masuda et al. [6], and Misuishi et al. [7]. The “OTELO” robot described by Delgorge et al. has six degrees of freedom (DOF), a remote center of motion and is directly placed on the abdomen of the patient while in use [4]. During the examination the system must be balanced by the patient or held of by a second person. The robotic concept “TER” presented by Vilchis et al. [5] is aligned on the patient by four belts through which it is able to adapt to the shape of the body of the patient. In combination with its three rotational DOF the system has five DOF. The slave robot in Masuda’s system is also lying on the patient’s body. It has three DOF [6]. Mitsuishi’s system has seven DOF to manipulate the US-probe. During the examination the slave robot is aligned at a sitting patient [7]. The presented tele-echographic approaches are similar because they were not designed for a specific medical examination but were rather focused on technical challenges.

Most prominent on-site US-manipulators have been proposed by Salcudean et al. [8] and Pierrot et al. [9]. Salcudean et al. proposed a backdrivable kinematics consisting of parallel linkages that has seven 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 [8]. 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.

All referenced robots employ a conventional US-probe. They guide the probe directly on the skin of the patient. To ensure continuous direct contact of the probe to the skin all the time is a demanding challenge. Therefore, complex kinematics with at least three translational DOF are necessary. In some cases simple contact with the skin is not enough. To improve the coupling of the US-waves into the patient several rotational DOF are necessary. During the application of the

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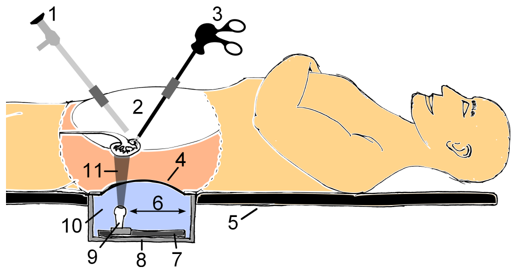

Fig. 1. Standoff concept for intraoperative ultrasound imaging through the back of a supine patient during laparoscopic tumor resection in urology. The US-probe (9) resides in a fluid (10). US-waves (11) are coupled into the patient through a single membrane (4). 1: Laparoscope, 2: Pneumoperitoneum, 3: Laparoscopic instrument, 4: Flexible membrane, 5: OR-Table, 6: Directions of movement of the ultrasound transducer, 7: Customized cinematics to move the transducer, 8: Tank, 9: Ultrasound transducer, 10: Transmission medium to guide the ultrasound waves from the transducer to the membrane, 11: Fan of ultrasound waves. Source Gumprecht et al. [1].

robots US-transmission gel must be applied repeatedly on the contact surface. Furthermore, the presented systems have no sterilization concept prohibiting intraoperative use.

In Gumprecht et al. [1] we proposed an alternative concept for an robot-guided US-probe (Fig. 1). Its purpose is continuous intraoperative real-time examinations during laparoscopic tumor resections in urology. The robot is integrated with the operating room (OR)-table underneath the patient. The DOF are limited to two. Each axis is independently electronically actuated. The manipulator is directly controlled by the surgeon through a joystick console. Sterilization is guaranteed through the covering of the manipulator and the console by sterile foils. US-images are provided by a conventional US-machine.

In order to be able to perform automatic US-imaging through the back of the patient with two DOF a custom standoff had to be designed [10]. Our list of requirements was as follows: a) the standoff must be flexible to adapt to the individual shape of the patient to allow for sufficient transmission of US-waves; b) the standoff should not require repeated application of US-transmission gel; c) the standoff should not impair the US-images, more precisely it should attenuate the US-waves as little as possible, it should not change the signal-to-noise ratio (SNR) and the range of contrast of the US-images.

This paper proposes the design as well as the experiments that lead to the standoff of the system of Gumprecht et al. [1].

II. Material and Methods

In the presented standoff, the US-probe resides within a tank filled with a fluid that transmits the US-waves (Fig. 1). The top of the tank is covered with a flexible membrane. On one side the membrane is in contact with the fluid an on the other side with the patient. Through its flexibility the membrande can be adapted to the shape of the patient. Therefore, the standoff permits sufficient coupling of US into the patient all the time. Greasing of the patient with US-gel is only necessary once before application since no direct movements are performed on the skin of the patient that could remove the gel.

A. Description of the experiment

In an experiment, we tried to determine which combination of a fluid with a single membrane provides the best US-images. In our case, that means little attenuation of the US-waves (lowering of the grey-values in the US-images), little degradation in the signal-to-noise ratio of the images and little lowering of the contrast of US-images.

Three fluids (F1: tap water, and F2: silicone oil (silicone oil 350 CST, engineering firm Claas Meinecke, Nordstemmen, Germany), F3: purified water) and four silicone membranes (M1: thickness (THK): 0.5 mm, shore hardness (HS): 60 °A; M2: THK: 1.1 mm, HS: 60 °A; M3: THK: 1.1 mm, HS: 40 °A; M4: THK: 1.5 mm, HS: 40 °A;) were used for the experiment. Load tests showed that all four membranes can handle even the weight of heavy patients (>300 kg) [1]. The fluid is pressurized to facilitate an optimum membrane shape to maximize the contact area with the patient. We employed a conventional US-machine (Terason, Burlington, MA, USA; Modell: 2000; 8IOL4 Intraoperative Smart Probe; center frequency: 7.5 MHz). The US-images were visualized and stored on a laptop using the *Terason Ultrasound System Software* (Version 3.6.5). Aquasonic 100 (Parker Laboratories Inc., Fairfiled, NJ, USA) was applied as US transmission gel. An US-phantom proposed by Seidl et al. [11] was used to record the images. Within the phantom there were artificial tumors and blood vessel.

The size of the US-images was 527×224 (H×W) at a penetration depth of 90 mm, e. g. 14.87 ppi. We evaluated two areas of 75×75 pixel (12.81 mm×12.81 mm) in the US-image for every setup. The first area was located on a homogenous position within the image. The distance from the surface of the phantom was the same for all image. The second area was located above the same tumor for all recordings. For both areas we determined three parameters:

a) average grey-value: Lower grey values in a B-mode image visualize lower amplitudes of the returning ultrasound waves, i.e., a loss of energy. If all system parameters are constant for two US-images except the membrane, lower grey values at identical areas of interest indicate a loss of energy caused by the membrane.

b) signal-to-noise ratio: One of the shortcomings of US compared to other medical imaging techniques like magnetic resonance imaging (MRI) or computed tomography (CT) is its poor SNR. Therefore, we calculated the SNR to assess how it is affected by the membrane. We calculated the SNR using the following formula

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

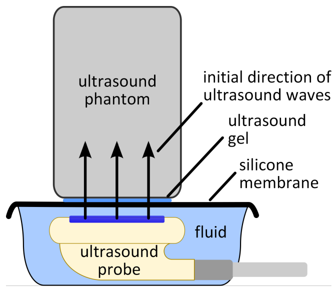

Fig. 2. Setup of the experiment to evaluate of the quality of the US-images. The distance between probe and the membrane in the experiment was set to the distance between the probe and membrane in the manipulator.

with X_i grey value of pixel #i and n total number of pixels in the area of interest.

$$SNR = 20\,log_{10}\,\frac{signal}{RMS\ noise} \tag{2}$$

The average grey value of the area of interest was chosen as the *signal*-value.

c) range of contrast: is an important factor for the quality of the US-images. We calculated the range-of-contrast (RoC) according the formula of Michelson [12]:

$$C_m = \frac{gv_{max} - gv_{min}}{gv_{max} + gv_{min}} \tag{3}$$

gv_{max} and gv_{min} are within the area of interest the highest and the lowest values above the noise level respectively.

An additional series of measurements without a membrane were performed as reference. Water was used as fluid. For each series of measurements we performed a t-test against this reference series. A probability of 5% was set as limit of significance.

During the experiment the US-probe resided in the fluid (Temperature: 20°C) and recorded US-images of the US-phantom through the membrane (see Fig. 2 a)). The US-phantom was directly applied to the membrane with a thin layer of the US transmission gel in between.

US-images were recorded for each combination of the three fluids with the four membranes at 21°C. The same US-phantom at the same position and orientation was used for all US-images. The settings for the US-machine were constant for all recorded images (Frequency: 7.5 MHz; maximal penetration depth: 9 cm).

III. Results

The results of the first experiment are summarized in the diagrams in Fig. 4. Two sample images are shown in Fig. 3. There was no difference measurable between the images recorded with purified water and tap water. Therefore, we decided to skip the tests with purified water.

A. Average grey values

As expected, the grey-values in images taken with any of the four membranes were lower than the grey-values in images taken without a membrane. Multiple reasons for this effect

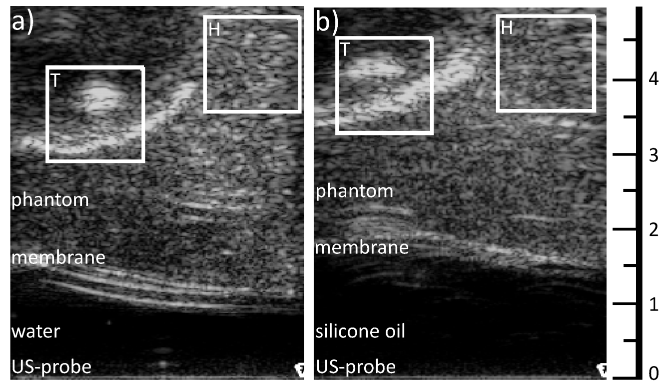

Fig. 3. US-images from the experiment: Initial direction of the ultrasound waves is from bottom to top. A metric scale with distances in cm is provided on the right. The squares highlight the areas that were evaluated in the experiment. The *T* marks the square covering the artificial tumor and the *H* marks the homogenous area. a) shows an image recorded with water and M3 (silicone membrane: thickness: 1.1 mm, shore hardness: 40° A); b) shows an image recorded with the same membrane but with silicone oil as fluid. The dark area on the bottom of the images is the fluid between the US-probe and the membrane. Above the fluid is the membrane. The lower and the upper boundary layers are clearly visible in a). Above the the membrane is the phantom with a clearly visible artificial tumor on the left side. Shadow effects are visible above the both tumors. Noteworthy is that on image b) the contours of the the membrane as well as the tumor are more blurred than in image a)

could be considered: Attenuation of the waves within in the membrane, reflections of the waves at the boundary layers of the membrane, scattering of the waves at the boundary layers etc. . Further investigations in this direction were not part of this publication. Most important results were: a) thicker membranes lower the amplitude (grey values in the B-mode image) of the glsus-waves more than thinner membranes b) lower hardness of the membrane results in higher amplitudes (grey- values in the B-mode image). On average there was no measurable different in the grey-value if either water or silicone oil is employed. However, the standard deviation of the grey-value averages taken with silicone oil were approximately twice as much as with water (7.3 vs. 3.4). For the homogenous area the t-test showed that all series of measurements were taken from different basic sets than the reference series of measurements, i.e. the membranes change the US-images significantly. Within the tumor area the series of measurements were too little for the t-test to provide meaningful results.

B. Signal-to-Noise ratio

The changes in the SNR were below 10 percent if a any of the four membranes were employed independent of the fluid (Homogenous Area: Water vs. Water-M vs. SilOil-M: 31,2 vs. 30.3 vs. 29.7). Therefore, the authors concluded that neither the kind of fluid nor the kind of membrane had an effect on the SNR. The t-test revealed no useful information due to a too little number of measurements. It was notable that the SNR in the tumor area were around 2/3 of the SNR in the homogenous area.

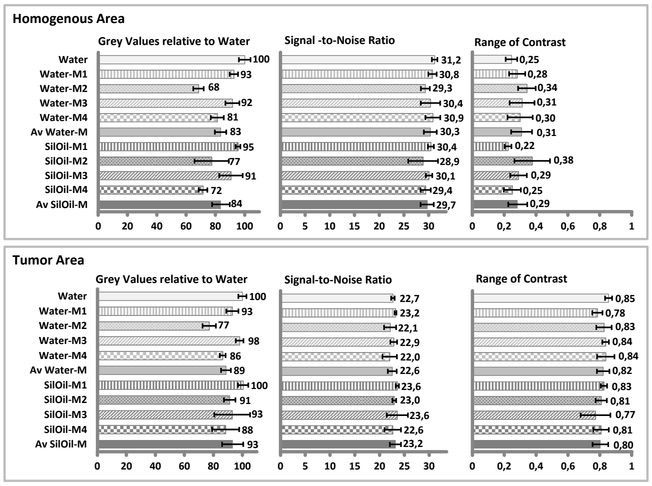

Fig. 4. Results of the experiment: The value to the right of every bar shows the measured value. The standard deviation is visualized with a "double T-mark" at the end of every bar. Measurements belonging to the same membrane are visualized with the same texture. Solid bars (Av Water-M; Av SilOil-M) show the average of the measurements using water or silicone oil respectively. The left diagrams show the average grey values for each combination of a fluid and a membrane. Values are relative to water with water set to 100. The top three diagrams show the measured values within the homogenous area and the lower three diagrams show the values measured within the tumor area.

C. Range of contrast

For the RoC no significant change between the different setups was measurable (Homogenous Area: Water vs. Water-M vs. SilOil: 0,25 vs. 0,31 vs. 0,29). Therefore, the authors concluded that neither the kind of fluid nor the kind of membrane had an effect on the RoC. On average, the values of the range of contrast within the homogenous area werwe a fourth of the values measured in the tumor region. The difference was caused by the white area of the tumor. By looking at the formula to calculate the RoC the reason for this result is obvious. In the tumor area the difference between the highest value (white) and the darkest value (black) were large compared to the sum of both values, resulting in a high RoC. While the difference is low for the homogenous area compared to the sum. The t-test revealed no useful information due to a too little number of measurements.

IV. Conclusion

In this paper, we present the results of our experiment for the optimization of a standoff design for a robotics-based flat-panel ultrasound device for intraoperative continuous transcutaneous imaging. In the proposed standoff concept, the US-probe resides in a tank filled with water. The US-waves traverse the water from the probe to the top of the tank. There is a flexible membrane stretched above the tank. The membrane is on one side in contact with the water and on the other side with the patient. The flexible membrane automatically adapts to the shape of the patient and therefore provide a sufficient coupling of US waves into the patient.

In an experiment, we investigated the quality of the US-images for the combination of the three fluids (purified water, tap water and silicone) in combination with four different (thickness, hardness) silicone membranes. Two different areas of interest in the US-images were evaluated for their average grey-value, signal-to-noise ratio (SNR), and range-of-contrast (RoC). The results were as follows: The thicker and / or the harder the membrane the lower the grey-values. Neither the membranes nor the fluid have an effect on either the SNR or the RoC. There was no measurable difference if purified water, tap water or silicone oil was employed. However, the visual impression of the US-images recorded with water is better than those recorded with silicone oil. In conclusion, the ideal standoff for the robot proposed by Gumprecht et al. [1] uses water as fluid and employs a thin and soft membrane that is still strong enough to carry the load of a patient.

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