Laser-scan endoscope system for intraoperative geometry acquisition and surgical robot safety management

Mitsuhiro Hayashibe a,*, Naoki Suzuki a, Yoshihiko Nakamura b

a Institute for High Dimensional Medical Imaging, The Jikei University School of Medicine, 4-11-1, Izumihoncho, Komae-shi, Tokyo 201-8601, Japan
b Department of Mechano-Informatics, University of Tokyo, 7-3-1, Hongo, Bunkyoku, Tokyo 113-8656, Japan

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Abstract

In laparoscopic surgery, surgeons find particular difficulties related to the operation technique. Due to restricted view, lack of depth information from the monocular endoscope and limited degree of freedom, surgeons find their movements impeded. A support system that provides improved laparoscopic vision would help to overcome the difficulties. If real-time visualization of abdominal structures were feasible, more accurate procedures and improved quantitative evaluations in laparoscopic surgery might be possible. In this study, a laser-scan endoscope system was developed to acquire and visualize the shape and texture of the area of interest instantaneously. The intraoperative geometric information of deformable organ could be applied for robotic safety management via geometric computation of robot position and organ shape. Results of in vivo experiments on a pig liver verified effectiveness of the proposed system.

Keywords: Robotic surgery; Laser-scanning; 3D shape recovery; Robotic safety management

1. Introduction

Minimally invasive surgery provides significant advantages for patients as smaller incisions are made and post-operative recovery is shortened. However, laparoscopic surgery forces surgeons to operate under conditions of greater mental stress and within mechanical and visual constraints. Looking at the monitor, the surgeon has to handle instruments inserted through trocars in the abdominal cavity via small holes in order to perform desired surgical tasks. The visual difficulties are due to narrow endoscopic field of view and lack of depth perception. Surgeons are required to develop a sense of laparoscopic orientation from memorized anatomical structures. Operating in this environment that differs from typical open surgery settings may lead to awkward manipulations and increase possibility of misperceptions (Committee of malpractice counter measure of Japan Gastroenterological Endoscopy Society, 2000).

A surgical assist system that would overcome the above-mentioned limitations has been specified and implemented. The initial idea is to have a robot holding the scope and responding to positioning commands of an operator (Taylor et al., 1994, Wang, 1993). Recently, teleoperated surgical robot systems have been developed and well-known commercial systems have been widely applied, e.g., ZEUS (Ghodoussi et al., 2002) (Computer Motion Inc.) and da Vinci (Guthart, 2000) (Intuitive Surgical Inc.). In these systems, the surgeon teleoperates the slave robot through master arms while utilizing the visual feedback from the laparoscopic images. This reduces surgeon’s tiredness and enhances his dexterity.

Robotic laparoscopic surgery would be technologically improved if surgeons were provided with a 3D representation of an internal geometry in an intuitive manner. As earlier work in that direction, a touch screen interface for non-master–slave operation of surgical robots was presented in Hayashibe and Nakamura (2001), where the 3D...
coordinates of a touched point on the screen were measured and used in surgical robot guidance. In the present study, we aimed to develop a surgical system for high-speed measurements of the 3D surgical field under laparoscopy and utilize it in surgical robot navigation. ZEUS and da Vinci systems, for instance, choose the master–slave configuration, but human interface is still limited to direct manipulation. We think that intraoperative organ shape data would be useful to provide 3D geometrical calculations of the abdominal space structures in order to avoid collision of surgical robot forceps. If a wrong master command is accidentally input by the operator, the system can warn the operator and thus inadvertent injury may be avoided.

Studies on visual servoing techniques and 3D position extraction in robotic laparoscopic surgery have been reported in the literature. Laparoscopic image analysis has been implemented as well as segmentation, localization and tracking of surgical instruments (Wang et al., 1998, Wei et al., 1997). Autonomous 3D positioning of a laparoscopic instrument has been achieved by means of a laser pointer and single standard monocular endoscope (Krupa et al., 2003). The surgeon could guide a surgical instrument that was out of the field of view. Some studies on systems measuring 3D shapes through an endoscope for medical use have been also reported. In the passive method, shading (Okatani and Deguchi, 1997) or parallel stereo pairs (Stoyanov et al., 2004) have been used for 3D shape extraction. In the active light projection, a laser beam has been applied for the triangulation through endoscopic optics (Haneishi et al., 1994).

In the present study, a laser-scan endoscope was developed to accomplish laparoscopic real-time 3D visualization of surgical objects with video-texture mapping. In the developed system, not only 3D point measurements but also 3D shape intraoperative measurements of soft tissue structures were performed during laparoscopy. Preliminary results of in vivo experiments verified functionality of this system. Robotic navigation and safety management was demonstrated by taking advantage of intraoperatively obtained surface shape of a pig liver as a clinical application.

2. Method
2.1. System configuration

Measurement accuracy is a fundamental factor, when intraoperative geometric information is obtained for quantitative evaluation of organs in the operative field and in surgery navigation. If the measurement is taken with a stereoscopic endoscope, the accuracy would be guaranteed for the area only around the tip of the endoscope. However, the distance between the viewpoints of the stereo pairs in one scope is insufficient for accurate measurement of the whole abdominal space. Our developed system enabled simultaneous acquisition and visualization of the shape and texture of a soft tissue organ instantaneously by using two scopes of an optical galvano scanner and a high-speed camera. Fig. 1 depicts the configuration of the prototyped system. Two endoscopes are inserted into the abdominal cavity, and inside, a laser-beam strip is actively controlled by an optical galvano scanner. A high-speed camera detects the laser beam line to obtain the 3D geometry of a given organ.

A closed-loop galvano scanner (General Scanning Inc.), which responds up to a frequency of 1 kHz is used. The laser line pattern is captured by a 262 fps high-speed camera (528 × 512 pixels, 256 gray scale, DALSA Inc. CA-D6 512 w) and a high-speed image processing board (CORECO Inc. Viper-Digital). Since the image obtained from the high-speed camera is in gray scale, and as such does not provide sufficient information to the surgeon, a beam-splitting prism is utilized to split the optical path. Color images are also captured by a 3 CCD digital camera through an IEEE1394 interface and presented to the surgeon. The information from a high-speed camera is processed only for the 3D geometric information of organs. The 3D coordinates of the reference points are reconstructed from the detected 2D data by the image capture/processing board and the input angle signals to the 2D laser scanner. The laser and camera coordinate systems are identified using an OPTOTRAK (Northern Digital Inc.). Fig. 2 shows the prototyped laser-scan endoscope. The upper image is the inside of the laser box and below is the dual-head of the high-speed camera and NTSC color CCD. All components, including galvano meter, laser diode and laparoscope are assembled in a compact laser box.
The computed surface of an operated organ was automatically visualized by real-time reconstruction of the scanned intraoperative range data. The on-line visualization of an organ shape could be executed by parallel processing with 2 PCs. The tasks were divided into measurement and visualization. The scanned 3D point data was stacked into the shared memory board of one PC and the other PC retrieved the point information and created polygons for surface reconstruction. The scene was updated by OpenGL to visualize the organ deformation with the frame rate 5–6 frames per second, if 20 lines were sampled for each frame (Pentium3 1 GHz).

2.2. Camera calibration

In order to calculate the numerical depth from the camera images, two types of parameters should be calibrated: internal camera parameters such as lens distortion and focal length, and external parameters such as camera position and its direction. In this study, the infra-red markers attached to camera and laser device as in Fig. 4 were used to obtain external parameters. The measuring precision of these markers has a RMS error of 0.1 mm and resolution of 0.01 mm. Only the external parameters were measured in situ during the procedure. The internal parameters were calibrated in advance as below.

The laparoscope optical system is composed of relayed, axially aligned spherical lenses in a stainless steel tube. In result, laparoscopes allow observation over a wide field of view, but bring large amount of geometric distortion. Here, for distortion correction, a radial distortion model (Haneishi et al., 1995, Stefansic et al., 2000), which assumes that the distortion occurs radially from the center of the lens, was adopted. The radial distortion model can be represented by

\[
 r = \sqrt{(x - x_0)^2 + (y - y_0)^2},
\]

\[
 r' = r + a_1 r^2 + a_2 r^3 + \cdots + a_{n+1} r^{n+1},
\]

\[
 x' = x_0 + r' \cos \theta,
\]

\[
 y' = y_0 + r' \sin \theta,
\]

\[
 \theta = \arctan\left\{\frac{y - y_0}{x - x_0}\right\},
\]

where \((x,y)\), pixel coordinates in the pre-corrected image; \((x_0,y_0)\), pixel coordinates of image center point; \((x',y')\), pixel coordinates in the corrected image; \(r\), radial distance of referent pre-corrected pixel point; \(r'\), radial distance of referent corrected pixel point; \(a_n\), coefficient of distortion correction.

The known lattice points were captured by the high-speed camera and the distance \(r\) in the image plane was measured and compared with the ratio of the actual distance. The distribution of the ratio \(r'/r\) as to the referent points was approximated as a third order polynomial expression as given below

\[
 r' = r + a_1 r^2 + a_2 r^3,
\]

\[
 a_1 = -2.0 \times 10^{-4} [1/\text{pixel}],
\]

\[
 a_2 = 6.0 \times 10^{-6} [1/\text{pixel}^2].
\]

In the case of 3D reconstruction and internal camera calibration, the corrected point was used. Fig. 3 depicts the referent points in the pre-corrected and corrected image coordinates.

The point \(P\) in the camera frame was expressed as

\[
 cP = [x_c\ y_c\ z_c]^T.
\]

![Fig. 2. The prototyped laser scan system. (a) Inside of the laser box containing laser source and galvanometer. (b) Dual-head camera of high-speed camera and NTSC color CCD.](image)

![Fig. 3. Calibrated points for optical distortion.](image)
Regarding the known point \( P \), the 2D pixel value were measured to obtain the relationship between the pixel coordinates and the camera coordinates. The redundant numbers of the reference points were determined by the camera calibration (Zuang et al., 1995, Tsai, 1987). The redundant numbers of the reference points were measured to obtain the relationship between the pixel coordinates and the camera coordinates. The calibrated parameters were used for the triangulation between the camera frame and the laser frame, as shown in Fig. 4.

Actually, the calibration was conducted using 50-point correspondences. The calibrated parameters were used for the triangulation between the camera frame and the laser frame, as shown in Fig. 4.

2.3. Laser calibration and triangulation

A laser beam was projected on an internal organ through the endoscope. The end of the endoscope was inserted into the abdominal cavity and the laser beam was scanned by the mirror of the galvano scanner at the other end, as illustrated in Fig. 5. The direction of laser beam was changed through a laparoscope. We performed calibration of the endoscopic optics and determined the relationship between input angle and output angle of the laser beam direction against the laparoscope.

\( p \) was transformed to the 2D coordinates in the image plane by perspective transformation as follows:

\[
\begin{pmatrix}
    x_c \\
    y_c \\
    z_c
\end{pmatrix}
= \begin{pmatrix}
    f \\
    f \\
    1
\end{pmatrix}
\begin{pmatrix}
    x_p \\
    y_p \\
    1
\end{pmatrix},
\]

(10)

where \([x_p y_c]^{\top}\) denotes coordinates point \( P \) in the image plane, and \( f \) is the focal distance of the camera. The transformation from the pixel unit to the unit of the camera coordinates is

\[
s_x x_c = H_0 - H, \quad s_y y_c = V_0 - V,
\]

(11)

where \([HV]^{\top}\) denotes the pixel coordinates of point \( P \) on the image. \([H_0 V_0]^{\top}\) implies that of the image centroid, and \( s_x, s_y \) are the camera scale factors. From Eqs. (10) and (11), we obtain

\[
C x = h,
\]

(12)

\[
C \equiv \begin{bmatrix}
    -x_c & 0 & z_c \\
    0 & -y_c & 0 \\
    z_c & 0 & 0
\end{bmatrix},
\]

(13)

\[
\alpha \equiv \begin{bmatrix}
    s_x f \\
    s_y f \\
    H_0 & V_0
\end{bmatrix}^{\top},
\]

(14)

\[
h \equiv \begin{bmatrix}
    z_c H \\
    z_c V
\end{bmatrix}^{\top}.
\]

(15)

\( \alpha \) contains the unknown transformation parameters from the camera coordinates to the pixel units. These are determined by the camera calibration (Zuang et al., 1995, Tsai, 1987). The redundant numbers of the reference points were measured to obtain the relationship between the pixel coordinates in the image and the camera coordinates. The least square computation was calculated using the pseudo-inverse matrix as follows:

\[
\alpha = C^{\dagger} h.
\]

(19)

The least square estimation of the transformation matrix was solved using pseudo-inverse matrix \( \alpha^\dagger \) of \( \alpha \) as below.

\[
\phi = \begin{bmatrix}
    \phi_{\text{yaw}} & \phi_{\text{pitch}}
\end{bmatrix}^{\top},
\]

(20)

\[
\psi = \begin{bmatrix}
    \psi_{\text{yaw}} & \psi_{\text{pitch}}
\end{bmatrix}^{\top},
\]

(21)

Assuming that the relation between \( \phi \) and \( \psi \) is linear

\[
\psi = T \phi^*,
\]

(22)

where \( \phi^* \) is defined as \([\phi_{\text{yaw}} \phi_{\text{pitch}} 1]^{\top}\) and \( T \) is \( 3 \times 2 \) transformation matrix. As with the camera calibration, the 3D known point indicated by laser beam [\( p = [x_l \ y_l \ z_l]^{\top} \)] and the corresponding input angle \([\phi_{\text{yaw}} \phi_{\text{pitch}}]^{\top}\) was measured \( N \) times. The \( i \)th measured value is expressed as \( \psi_i, \phi_i \). These parameter relationship was defined as follows:

\[
\psi = T \phi^*.
\]

(23)

\[
\psi \equiv \begin{bmatrix}
    \psi_1 & \cdots & \psi_N
\end{bmatrix},
\]

(24)

\[
\phi \equiv \begin{bmatrix}
    \phi_1 & \cdots & \phi_N
\end{bmatrix}.
\]

(25)

The least square estimation of the transformation matrix \( T \) was solved using pseudo-inverse matrix \( \phi^\dagger \) of \( \phi \) as below.

\[
T = \bar{\psi} \phi^\dagger.
\]

(26)
The direction vector of laser beam $\mathbf{l}_{\text{las}}$ would be expressed as

$$
\mathbf{l}_{\text{las}} = \begin{bmatrix} 1 & \tan \psi_{\text{yaw}} & \tan \psi_{\text{pitch}} \end{bmatrix}^T.
$$

(27)

The 3D coordinates of the laser mark were obtained as those of the intersection between the laser beam $\mathbf{L}_{\text{laser}}$ and the perspective line $\mathbf{L}_{\text{camera}}$. Eq. (12) can be rewritten as follows:

$$
\mathbf{D}'\mathbf{p} = \begin{bmatrix} 0 \\ 0 \end{bmatrix},
$$

(28)

$$
\mathbf{D} = \begin{bmatrix} s_x f & 0 & H - H_0 \\ 0 & s_y f & V - V_0 \end{bmatrix}.
$$

(29)

$\mathbf{p}$ that satisfies this equation is the laser marked point in camera coordinate as in Fig. 4. Thus, the laser marked point in the laser coordinate can be determined by Eq. (27) as below when its $x$ value is $u$

$$
\mathbf{p} = u\mathbf{l}_{\text{las}}.
$$

(30)

The equation relating $\mathbf{p}$ to $\mathbf{p}'$ was written as below with the homogeneous transformation matrix from camera to laser coordinate system

$$
\begin{bmatrix} \mathbf{p}' \\ 1 \end{bmatrix} = \begin{bmatrix} \mathbf{R}_i & \mathbf{t}_{\text{lc}} \\ \mathbf{0}_{3 \times 3} & 1 \end{bmatrix} \begin{bmatrix} \mathbf{p} \\ 1 \end{bmatrix},
$$

(31)

where $\mathbf{R}_i$ is the rotation matrix from camera to laser coordinate, $\mathbf{t}_{\text{lc}}$ is the origin vector of laser frame in the camera coordinate system. Eqs. (28), (31) were summarised using Eq. (30) as follows:

$$
\mathbf{M}_\eta = \mathbf{m},
$$

(32)

$$
\mathbf{M} = \begin{bmatrix} \mathbf{D} & \mathbf{O}_{2 \times 1} \\ \mathbf{E}_{3 \times 3} & \mathbf{R}_i \mathbf{l}_{\text{las}} \end{bmatrix},
$$

(33)

$$
\eta = \begin{bmatrix} \mathbf{p}' \\ u \end{bmatrix}^T,
$$

(34)

$$
\mathbf{m} = \begin{bmatrix} \mathbf{O}_{2 \times 1} & \mathbf{t}_{\text{lc}} \end{bmatrix}^T.
$$

(35)

$\mathbf{M}$ and $\mathbf{m}$ can be computed from the calibrated parameter, the scanning angle and camera image. Finally, the unknown parameter $\eta$ including the position of laser mark in camera coordinate $\mathbf{p}'$ was solved using least square estimation as

$$
\eta = \mathbf{M}'\mathbf{m}.
$$

(36)

### 2.4. Automatic polygon and texture generation

First, the laser-scan endoscope outputs the range data set of the scanned organ. If effective visualization of the organ shape is to be obtained using range data, the surface should be composed of numerous triangle patches. The combination of points that form these triangles can be used for geometric computation by checking collision between the given organ and surgical instruments, such as forceps. We developed a program, automatically generating triangular patches from range data for each frame. From the point of view of a laser scanner, the range data is a collection of points mostly located in a lattice, as shown in Fig. 6(a). When three neighboring points exist in a set of this order; the surface is considered as a collection of these triangles. If the number of sampled points in a line is $N_p$, and the number of scanned lines is $N_l$, then the combination of vertices of the upper triangle is

$$
(iN_p + j, (i + 1)N_p + j, iN_p + j + 1)
$$

(37)

and of the lower triangle is

$$
((i + 1)N_p + j, (i + 1)N_p + j + 1, iN_p + j + 1)
$$

(38)

where columns range from $i = 0, 1, 2, \ldots, N_l - 1$ and rows range from $j = 1, 2, \ldots, N_p$. The combination of vertices of the triangle patch is decided repeatedly. Fig. 6(b) depicts the surface model transformed by this algorithm. The condition of unevenness is easily perceived from lighting and sequentially scanned surfaces.

As described above, the range data was calculated from the triangulation of laser and camera system. When the range data was re-projected into the camera image as in Fig. 7, the correspondent coordinates of the measured
points could be determined for the laparoscopic image. Texture coordinates could be calculated as the origin is the bottom left corner of the image. Using OpenGL, the updated scanned surface was redrawn with newly captured texture from the laparoscopic video image in real-time. The scanned 3D data was automatically described in VRML 2.0 by the program that we developed. The shape of a soft tissue organ could be easily perceived utilizing a www browser with VRML plug-in software.

3. Experiments and results

3.1. Evaluation of measurement accuracy

In order to evaluate accuracy of the measurements, we scanned an object of a known shape with the laser-scan endoscope. A plane was located at a distance of approximately 160 mm and inclined at 20° angle against the camera coordinate. Fig. 8 shows the reconstructed surface from 8000 measured points in the area of 8 cm². Simultaneously, the position of the plane was measured by an OPTOTRAK, which has an accuracy of 0.1 mm RMS error. The position and attitude assumed as the actual value was calculated from the positional data at several points obtained with the OPTOTRAK digitizer. The measured data obtained from the laser-scan endoscope was subtracted from the actual plane. The average difference values were plotted in each series of x coordinate as in Fig. 9. The average error was 0.16 mm and the maximum error was 1.2% against the distance z from the tip of endoscope to the object.

Next, the sphere having 45 mm in diameter made of silicon rubber was scanned. It was located at a distance of 130 mm from the tip of endoscope. Fig. 10 shows the reconstructed surface from 8000 measured points in the area of 8 cm². The system detected the bright line of the laser beam on a gray-scale image from the high speed camera. It was feasible to measure an object that had the same red color as the laser light source itself. In addition, due to high refresh rate for CMOS elements of high speed camera, the glare caused by reflection on the object surface did not influence the measurement of surfaces in contrast to the case with color CCD. The maximum error was 1% against the distance z from the tip of endoscope to the object.

In laparoscopic surgery, the operation is conducted under the environment in which the distance between the laparoscope and the object is 10–15 cm. In this range, the maximum error is estimated as 1.0–1.5 mm. There would be a variation in the error margin depending on the angle between the optical axis of the camera system and the laser system. In addition, limitation of camera resolution and illumination conditions at the time of scanning would also influence the measurement accuracy. A large parallax might decrease the factor of margin of error. However, in the laparoscopic environment, the angle between the scope of the camera system and the laser system would be assumed to be approximately 40°. When this experiment was done, the object was scanned in the same positional condition.

3.2. Measurement of a deformable organ

Next, we tested measurement and visualization of the surface of an isolated pig’s liver. The laser scope and the camera scope were positioned at a distance of 15 cm from the organ. Even with the shining and wet surface, the image processing of the high-speed camera was successful and the deformation of the organ surface could be observed with rendered computer graphics in real-time. The frame rate for surface visualization actually depends on scanning resolution. In the rough scan mode, the number of sampled lines decreases leading to acceleration in the cycle updates. If a fine surface measurement is required, it is possible to change the scanning settings from the GUI of the program. In the rough scanning mode, we usually set the sampling number of scan lines to 20. In such case, the frame rate...
was 5–6 fps. Approximately 100 lines per second could be captured and measured. Even in the rough mode, it was possible to sufficiently perceive the entire shape of the object.

In this experiment, an incision was made on the liver surface as in Fig. 11. The length of the incision was 44 mm and the depth was 7 mm. The graphics at the bottom of Fig. 11 depicts the reconstructed surface of the liver. The scanned area is 8 cm × 6.5 cm. The lines show normal vectors of each vertex. Even on this uneven surface, the visualization of liver deformation was successfully carried out. Fig. 12 shows 3D visualization of crossed incisions with video-texture mapping. Surgeons would be able to perceive the depth on the laparoscopic image intuitively with this visualization system.

3.3. In vivo measurement experiment

In order to investigate optical conditions under laparoscopy and feasibility of the device settings, we made an in vivo experiment to obtain the intraoperative 3D geometry and execute a safety management protocol of the surgical robot with the geometric computation. Actual settings of the experiment are shown in Fig. 13.

The laparoscopic procedure consisted of the following steps:

1. The abdominal cavity was filled with CO₂ gas and the abdominal wall was lifted up by the static pressure. The surgical space and the space for observation were kept clear inside the abdomen.
2. The trocar sites of approximately 1 cm in diameter were opened at several points on the abdominal wall. The laparoscope was inserted through the trocar.
3. Under the laparoscopic view, the scope of the laser was guided to indicate an appropriate point on the organ.
4. Optical markers for the camera and laser devices were successfully detected by OPTOTRAK.
5. The area of interest on the abdominal organ was scanned and its geometry was measured and registered into the virtual space. The positional information of the robotic devices was also reflected in the geometry management system.

Even under in vivo conditions, it was feasible to obtain the 3D surface data using a 100 mW semiconductor laser. The eventual laser output through the endoscope was estimated as 20% of the laser light source. This was caused by the loss from inner characteristics of the endoscopic optics.
The intraoperative 3D geometry of the liver surface could be obtained in one second in fine scanning mode. Fig. 14 provides a 3D reconstructed surface of the liver during the in vivo measurement. We scanned the area of 20 cm² and obtained the 4000 points data. The obtained information could be used, for example, to quantitatively measure location, length, and/or area of interest inside the abdomen.

### 3.4. Robotic navigation with safety management

Spatial perception remains one of the main difficulties in laparoscopic surgery. Due to the narrow field of view, the area observable through the endoscope is limited. This carries a possibility of unexpected collisions between the organ and surgical robot forceps in the area that is out of view. Our system allowed representation of 3D geometry of an organ and surgical robot in the virtual space as shown in Fig. 15. Surgeons were able to ascertain how far the organ is from the tip of forceps with this intraoperative monitoring technique.

Using intraoperative geometric information collected by the laser-scan endoscope, we developed the geometry-based surgical robot navigation system. Laser-scan endoscope and surgical robot controller were connected to the computers that were linked with a shared memory board to facilitate real-time data transmission. The shared memory board (Memolink) enabled high-speed data transmission at the rate of 2 MB/s. The information was registered into the virtual space. The coordinates of device were measured by the OPTOTRAK. Motion of the surgical robot was also monitored in the virtual space. The geometric model of surgical robot and the scanned organ model were combined and integrated in the geometry management system. These
models allowed prediction of collision and proximity between organs and surgical instruments. Computation of geometric collision could inform surgeons of unexpected collision in advance, thus improving safety of the operation procedure.

In the study, AESOP (Computer Motion Inc.) was adopted as the surgical robot to hold forceps. AESOP was originally designed for laparoscope positioning and has 6 joints including 2 passive joints. The link parameters of these 2 joints are passively decided by the constraint of the abdominal wall hole. We redeveloped the robot controller to obtain the count of joint encoders. The AESOP modeling was done based on link shape and link parameters. Fig. 16 shows the flow chart for AESOP control.

Using synchronous flag, joint angles were transmitted into that of the virtual model at every 10 ms. The controller always observed the emergency stop flag, which was put on if proximity, within a certain tolerance, between the robot and the organ was detected. Fig. 17 shows the virtual model activated by the motion data from the surgical robot.

To prevent forceps collision with the organ or endoscopes, the system was constantly checking the distances between objects. For computation of the distances, we adopted the Proximity Query Package (Department of Computer Science, 1999) that is a geometric computation library for proximity queries performed on polyhedrons. The package provided distance computation and tolerance check. It calculated the distance between two models and two points giving the minimum distance for the models. Tolerance check could detect whether two models are closer or farther than a tolerance value; tolerance check computation was faster than distance computation.

Fig. 18 shows an example warning window for the surgeon to avoid the collision. If the distance between two objects was closer than the critical distance, the window warned surgeons by changing background color and...
popping up a warning message. For better navigation without collisions, the system could also show the shortest path between the closest points. It enabled emergency stop of the slave robot in case of emergency, even in the situation when a surgeon did not notice the closeness. In the experiment, the surgical robot could be successfully stopped when the forceps approached the organ as close as 10 mm as shown in Fig. 19. This was the minimum distance transition between the forceps and the organ. If the intraoperative information of the abdominal geometry were provided for the surgeon in a sub-display of the laparoscopic image, the doctor who operates the surgical robot could perceive the navigated entire surgical geometry as well as the usual laparoscopic view at the master console.

4. Conclusion

A laser-scan endoscope was developed to achieve real-time 3D visualization of surgical field with texture information in laparoscopy. The dual-head of the high-speed camera and the 3CCD digital color camera with a beam-splitting prism were integrated into one system for a high-frame rate 3D surface shape acquisition. Preliminary results of in vivo experiments verified functionality and showed results of the system performance. Safety management in laparoscopic robotic surgery could also improved with geometric computation of organ-instrument collisions within operative field using intraoperative surface shape representation of a liver.

The laser-scan endoscope system may have several applications. In tele-surgery or tele-medicine, an expert doctor in remote location may wish to know the 3D shape and size of a particular organ in order to consult effectively on clinical decision making; such information would be valuable also for local surgeons. Surgeons would interactively activate the laser-scan endoscope for scanning the area of interest. Being combined with ultra-violet light-emitting technology, the laser-scan endoscope might be used to locally illuminate and find the dye absorbing areas.

The precise anatomic relations and structure of organs are usually accessible with CT or MRI. The radiological data is evaluated during pre-operative conferences, when surgical procedures are discussed and planned. One of the current issues is how to utilize such data in the operating room. The CT or MRI data is geometrically inconsistent due to deformations related to changes in patient’s body postures. To perceive deformations, a surgeon has to develop a sense of imagination to map the pre-surgical data onto the deformed one. In the future, we hope that the intraoperative geometric data will be utilized for registration of preoperative 3D CT/MRI data to assess position of blood vessels. Such system has been developed and evaluated in our institute (Hayashibe et al., 2002). However, estimation of a deformed internal structure only with the organ surface shape computation still poses some difficulties. It would be necessary to integrate the organ shape data with information from other multimodal sensors such as ultra-sound and mobile C-arm CT in the operating room.

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