

Available online at www.sciencedirect.com



Medical Image Analysis 11 (2007) 429-442

MEDICAL IMAGE ANALYSIS

www.elsevier.com/locate/media

Recovery of respiratory motion and deformation of the liver using laparoscopic freehand 3D ultrasound system

Masahiko Nakamoto^{a,*}, Hiroaki Hirayama^a, Yoshinobu Sato^a, Kozo Konishi^b, Yoshihiro Kakeji^b, Makoto Hashizume^b, Shinichi Tamura^a

> ^a Division of Image Analysis, Osaka University Graduate School of Medicine, Japan ^b Kyushu University Graduate School of Medical Sciences, Japan

Received 20 April 2007; received in revised form 16 July 2007; accepted 17 July 2007 Available online 19 August 2007

Abstract

The present paper describes a method for intraoperative recovery of respiratory motion and deformation of the liver by using a laparoscopic freehand 3D ultrasound (US) system. The proposed method can extend 3D US data of the liver to 4D by acquiring additional several sequences of time-varying 2D US images during a couple of respiration cycles. 2D US images are acquired on several sagittal image planes and their time-varying 3D positions and orientations are measured using a miniature magnetic 3D position sensor attached to a laparoscopic US (LUS) probe. During the acquisition, the LUS probe is assumed to move together with hepatic surface. Respiratory phases and in-plane 2D deformation fields are estimated from time-varying 2D US images, and then time-varying 3D deformation fields on sagittal image planes are obtained by combining 3D positions and orientations of the image planes. Time-varying 3D deformation field of the volume, that is, 4D deformation field, is obtained by interpolating the 3D deformation fields estimated on several planes. *In vivo* experiments using a pig liver showed that the proposed method could perform accurate estimation of respiratory cycle and in-plane 2D deformation fields. Furthermore, evaluation for the effects of sagittal plane interval indicated that 4D deformation fields could be stably recovered.

© 2007 Elsevier B.V. All rights reserved.

Keywords: Respiratory motion compensation; Laparoscopic surgery; Non-rigid registration; Intraoperative image analysis; 4D deformation field

1. Introduction

In liver surgery, ultrasound (US) is a useful modality for intraoperative imaging of internal structures of the liver such as vessels and tumors due to its realtime and non-invasive nature. Therefore, surgical navigation systems using US have been developed (Sjølie et al., 2003; Boctor et al., 2004; Hong et al., 2004; Boctor et al., 2006). As laparoscopic surgery is becoming common, laparoscopic surgical navigation using US has also been developed, including systems based on 2D US images (Ellsmere et al., 2004; Kleemann et al., 2006), and those based on 3D US (Harms et al., 2001; Nakamoto et al., 2002; Leven et al., 2005; Bao et al., 2007). Aiming at safe and accurate surgery, the internal structures can be virtually seen through by superimposing 3D US data onto real laparoscope view.

Motion and deformation of the liver due to respiration is a major problem in maintaining accurate superimposition. 3D US is acquired during breath holding at expiration phase, and thus misalignment between 3D US and laparoscopic view occurs at other phases. Although breath holding during subsequent superimposition is often performed as well, it not only disturbs smooth operation but also may affect patient's condition if performed many times. Olbrich et al. propose a gating method for visualizing superimposition only at expiration phase, which is regarded as the most stable phase of a respiratory cycle (Olbrich et al., 2005). However, its visualization is not continuous but intermittent. Thus, repeatedly, superimposed images and raw laparoscopic images

^{*} Corresponding author. Tel.: +81 6 6879 3564; fax: +81 6 6879 3569. *E-mail address*: nakamoto@image.med.osaka-u.ac.jp (M. Nakamoto).

^{1361-8415/\$ -} see front matter @ 2007 Elsevier B.V. All rights reserved. doi:10.1016/j.media.2007.07.009

(non-superimposed) are displayed alternately. Such a discontinuous and blink visualization may affect safe surgery, although every effort is exercised by a surgeon to perform a careful and precise operation (i.e. resection of the liver avoiding vessels). On the other hand, recovery of the liver motion from preoperative MR images has been reported (Rohlfing et al., 2004; Blackall et al., 2005; Tokuda et al., 2006; von Siebenthal et al., 2007), albeit with the following disadvantages in intraoperative use: (1) preoperative images may not capture the actual intraoperative dynamic motion of the liver in laparoscopic setup in which the abdominal cavity is filled with gas; and (2) acquisition of time-varying 3D MR images in acceptable spatiotemporal resolution is time-consuming.

In this paper, we describe a method for intraoperative recovery of respiratory motion and deformation of the liver using a freehand 3D laparoscopic US (LUS) system to generate augmented reality visualization of 4D US. The advantages of our method are as follows:

- Data acquisition protocol is simple and rapid so that it can be performed intraoperatively. Acquisition of several cross-sectional time-varying 2D US images is only required in addition to conventional 3D US image acquisition performed by freehand scanning with an LUS probe.
- Intraoperative dynamic motion and deformation of the liver in laparoscopic setup are recovered.

The proposed method utilizes the assumptions on characteristics of the liver motion indicated previously (Rohlfing et al., 2004). Its main component exists in the cranio-caudal (5–25 mm) and anterior-posterior (1–12 mm) directions, while that in the left-right direction is sufficiently small (1-3 mm) (Rohlfing et al., 2004). That is, we assume that non-rigid deformation mainly exists in the sagittal planes and is negligible in the left-right direction (although rigid motion component across the sagittal planes is incorporated in our method). In addition, we assume that the liver motion is periodical because the patient is under general anesthesia with breathing controlled by a respirator. The proposed method combines rigid 3D motions of the image planes corresponding to LUS probe motion and time-varying in-plane 2D deformation fields estimated from time-varying 2D US images acquired at several (quasi-)sagittal planes to recover 4D deformation field of the liver. Since a LUS probe is in contact with hepatic surface during US image acquisition, we assume that the motion of corresponds to that of hepatic surface with moderate contact pressure. Animal experiments are carried out to evaluate and demonstrate the feasibility of the proposed method.

2. Methods

2.1. System overview and data acquisition

We employ a freehand 3D laparoscopic ultrasound (LUS) system, in which the position and orientation of the LUS

probe tip are measured using an attached six degree-of-freedom miniature magnetic tracker (Fig. 1). Spatial relationship between the tracker and the LUS image is determined by preoperative calibration (Sato et al., 1998; Nakamoto et al., 2002). In order to calibrate delays among optical tracker, magnetic tracker and ultrasound machine, temporal calibration is performed by using the method described in Nakada et al. (2003). Magnetic field distortion caused by metallic objects is intraoperatively corrected in a rapid manner as described in Nakada et al. (2003).

Time-varying 2D US image sequences of several (quasi-)sagittal planes as well as their positions and orientations are acquired during two or more respiratory cycles (Fig. 2). We assume that the LUS probe tip is put directly on hepatic surface with moderate contact pressure so that they move concurrently. The time-varying dataset $D_i(t)$ of *i*th sagittal plane is defined as

$$D_i(t) = \{ (I_i(\mathbf{p}, t), T_i(t), R_i(t)) | t = 1, \dots, N_i \},$$
(1)

where $I_i(\mathbf{p}, t)$ is time-varying 2D US images, and $T_i(t)$ and $R_i(t)$ are positions and orientations of the US images, respectively. **p** denotes 2D coordinates of the US images, *t* denotes a frame number, and N_i is the total number of acquired frames for *i*th sagittal plane.

2.2. Respiratory phase analysis and temporal registration

2.2.1. Definition of respiratory phases

Reference time points in the respiratory cycle for $D_i(t)$ are defined for the necessary temporal registration among acquired time-varying datasets, $D_i(t)$, of different sagittal planes. We define the start and end points of deformation (SPD and EPD) as well as those of inspiration (SPI and EPI) for each time-varying dataset by analyzing the position of US image planes and time-varying similarity between temporally adjacent US images. Normalized cross correlation (NCC) is used as a similarity measure, which is considered to be in inverse proportion to the amount of deformation. US image position is defined as the distance from the origin of magnetic tracker frame to that of US image frame, i.e. $|T_i(t)|$.

Fig. 3 shows an example of time-varying similarity and position of the image plane. Respiratory cycle is divided into static and dynamic phases based on the magnitude of similarity. SPD is defined as the point where the similarity begins to decrease (point D in Fig. 3). EPD is defined as the point where the similarity returns to a large value (point B in Fig. 3). Using SPD and EPD, dynamic and static phases are determined. Since the similarity has a nearly constant and large value in static phase (from point B to D), we regard that in-plane deformation is negligible in this phase, and thus estimation of in-plane deformation in static phase is not performed for computational efficiency (although rigid motion is incorporated).

EPI is defined as the peak point (point A) during dynamic phase (from point D to B), and used as a reference point for temporal registration because it can be localized



Fig. 1. Intraoperative position measurement and laparoscopic ultrasound image acquisition of the liver. Upper: spacial relationships from sagittal and coronal view. Lower: representative probe placement for ultrasound image acquisition.



Fig. 2. Time series of 2D sagittal ultrasound images.

stably and accurately. The point where the change in both US images and position is the smallest is defined as SPI (point C), and the US image at SPI is used as a reference image for in-plane non-rigid registration described in Section 2.3 because it is regarded as the most stable state. The remainder of this section describes the methods for

localizing points SPI, EPI, SPD, and EPD and temporal registration in details.

2.2.2. Methods for respiratory phase segmentation

Respiratory phase analysis is based on 1D signal analysis of time-varying similarity and position shown in Fig. 3.



Fig. 3. Analysis of respiratory phase: (A) End point of inspiration (EPI); (B) Start point of deformation (SPD); (C) Start point of inspiration (SPI); (D) End point of deformation (EPD). (Position is appropriately scaled and shifted so that comparison with the similarity is facilitated).

Because all reference points except SPI are located at the peak or shoulder positions of time-varying similarity, where the similarity begins decreasing and/or ends increasing, we use local minima of the Gaussian-smoothed second derivative ($\sigma = 5.0$ frames) of time-varying similarity (Fig. 3) to localize these reference points. Local maxima of the second derivative with large magnitude values are also detected in order to identify each reference point. The prominent local maxima of the second derivative are easily identifiable as shown in Fig. 3. EPD (point B) is identified as the local minimum of the second derivative which is firstly encountered on the right side of the local maximum. EPI (point A) is identified as the maximum of the similarity located between the local minimum of the second derivative encountered firstly on the left side and the prominent local maximum. Further, SPD (point D) is identified as the local minimum of the second derivative with the minimum value in the interval from EPD (point B) to EPI (point A). Respiratory cycle length is determined as the average interval length between adjacent EPIs.

It is not appropriate to identify SPIs (point C) based on the second derivative because SPIs are located in the plateau of the similarity curve. By analyzing time-varying position, SPI is determined as the center point of the period minimizing the position variation $P_i(t)$ in the interval from EPD to SPD (point B–D), which provides an approximate center of the plateau. The position variation $P_i(t)$ is defined as

$$P_i(t) = \sum_{s=-w}^{w} ||T_i(t)| - |T_i(t+s)||,$$
(2)

where *w* is a range of summation and is set to a quarter of the cycle length.

2.2.3. Temporal registration of US image sequences

Temporal registration among time-varying datasets is performed by aligning EPIs (point A) of the datasets. One reference EPI located in the middle of the sequence is selected from each dataset, and the datasets are shifted so that the reference EPIs are aligned. Since estimated respiratory cycle length differs slightly among datasets, the average \overline{L} of all estimated lengths is used. We assume that a respiration cycle starts from SPI and ends at next SPI (point C). The positions of SPIs, that is, both ends of a respiration cycle, are redefined in each dataset after temporal registration by obtaining the average of interval length from SPI to EPI, γ , and then redefined SPI positions before and after the reference EPI are determined as SPI' = EPI - γ and SPI'' = SPI' + \overline{L} , respectively. Using redefined SPIs, a normalized frame number t'is obtained as t' = t - SPI'. In normalized respiration cycle, SPIs are located at t' = 0 and \overline{L} , and EPIs at $t' = \gamma$ in all the datasets. It should be noted that SPDs and EPDs are not aligned among the datasets even after the temporal registration although their positions are shifted as SPD' = SPD - SPI' and EPD' = EPD - SPI', respectively.

2.3. Estimation of in-plane deformation

To estimate time-varying in-plane deformation fields from US image sequences, we perform intensity-based non-rigid registration (Rueckert et al., 1999) during dynamic phase (from point D to B). Normalized cross correlation (NCC) is used as a similarity measure. We assume that in-plane deformation is negligible during static phase (from point B to D in Fig. 3), and thus the estimation of deformation field is not performed (i.e. the deformation field of static phase is filled by zero vectors). Given timevarying US images of *i*th sagittal plane indexed by the normalized frame number t' = t - SPI', 2D deformation field at *t*'th frame is defined as $\mathbf{m}_i(\mathbf{p}, t')$ maximizing

$$NCC(F(I_i(\mathbf{p}, 0), \mathbf{m}_i(\mathbf{p}, t')), I_i(\mathbf{p}, t')),$$
(3)

where NCC(*I*, *J*) denotes NCC of images *I* and *J*, *F*(*I*, **m**) denotes the deformed image of *I* by deformation field **m**. To accurately estimate the deformation of the liver, hepatic region is manually segmented beforehand in the first frame (at t' = 0), which is not time-consuming because only one short curve is specified at the bottom of the liver in one image in the sequence. To improve the robustness of registration, we employ the coarse-fine approach described in Rohlfing et al. (2004). First, deformation is estimated by using coarse control point grid. Next, grid size is refined to half, and thus more detailed deformation is estimated by using refined control point grid. How the initial deformation for coarse estimation is provided is as follows.

Dynamic phase can be divided into a former and a latter phases, which are intervals from SPD to EPI (point D–A) and from EPI to EPD (point A–B), respectively. In the latter phase, the similarity abruptly decreases and increases (Fig. 3) while the similarity variation in the former phase is not so large as compared with the latter one. It is considered that large changes in US images are caused by rapid movement of the liver due to inspiration in the latter phase. Based on the above observation, different methods are used to provide the initial deformation field in the former and latter phases. In the former phase where the changes are not large, the difference between temporally adjacent t'th and (t' + 1)th deformations is regarded as small. Therefore, estimated t'th coarse deformation is used as the initial coarse deformation for the estimation of (t' + 1)th deformation. In the latter phase, because our preliminary experiments indicate that the result of the previous frame is not sufficient for the initial deformation due to rapid movement of the liver, we utilize the estimation of the deformation fields in the former phase. The US image with the highest similarity to the current image to be processed is selected in the former phase, and then coarse deformation field of the selected image is given as the initial deformation field (Fig. 4).

2.4. Recovery of 4D motion and deformation

The recovery processes of 4D liver motion and deformation consist of the following three stages (Fig. 5): (1) Recovery of 3D positions and displacement vectors corresponding to pixel positions of 2D US images; (2) Recovery of continuous 3D deformation field from scattered dataset of 3D positions and displacement vectors; and (3) Recovery of a 4D US model by combining a static 3D US model and the recovered time-varying 3D deformation field. The details of each stage are described below.

2.4.1. Recovery of 3D positions and displacement vectors

Using the method described in Section 2.3, 2D displacement vector $\mathbf{m}_i(\mathbf{p}, t')$ is obtained at each pixel position \mathbf{p} and frame t' of 2D US image sequences. By combining the position and orientation of US image $(T_i(t'), R_i(t'))$, measured using a magnetic tracker, 2D position **p** of US images and the corresponding time-varying 2D displacement vector $\mathbf{m}_i(\mathbf{p}, t')$ are converted to time-varying 3D position $\mathbf{X}_i(\mathbf{p}, t')$ and displacement vector $\mathbf{U}_i(\mathbf{p}, t')$, respectively. Time-varying 3D position $\mathbf{X}_i(\mathbf{p}, t')$ is given by

$$\mathbf{X}_{i}(\mathbf{p}, t') = R_{i}(t')(R_{\rm us}(p_{\rm x}, p_{\rm y}, 0)^{\rm T} + T_{\rm us}) + T_{i}(t'), \qquad (4)$$

where $\mathbf{p} = (p_x, p_y)$, and R_{us} and T_{us} are a 3 × 3 rotation matrix and a 3D vector to transform coordinates from the US plane frame to the magnetic tracker frame, respectively, obtained by US calibration performed beforehand. Time-varying 3D displacement vector $\mathbf{U}_i(\mathbf{p}, t')$ is given by

$$\mathbf{U}_i(\mathbf{p},t') = (\mathbf{X}_i(\mathbf{p},t') - \mathbf{X}_i(\mathbf{p},0)) + R_i(t')R_{us}(m_x,m_y,0)^{\mathrm{T}}, \quad (5)$$

where $\mathbf{m}_i(\mathbf{p}, t') = (m_x, m_y)$. Note that 3D vector $\mathbf{U}_i(\mathbf{p}, t')$ represents displacement from 3D position $\mathbf{X}_i(\mathbf{p}, 0)$ at t' = 0.

2.4.2. Recovery of continuous 3D deformation field

We describe the deformation field at each frame t' as displacements from 3D positions in the first frame at t' = 0. Given datasets of $\mathbf{X}_i(\mathbf{p}, 0)$ and $\mathbf{U}_i(\mathbf{p}, t')$ on each sagittal plane *i*, continuous 3D deformation field $\mathbf{u}(\mathbf{x})$ at each frame t' is recovered, where \mathbf{x} is a continuous variable of 3D position. Here, it should be noted that \mathbf{p} and t' are discrete variables and the sagittal planes are placed manually at intervals. Thus, we have discrete and sparse datasets of $\mathbf{X}_i(\mathbf{p}, 0)$ and $\mathbf{U}_i(\mathbf{p}, t')$. To obtain continuous 3D functions described as

$$u = f_u(x, y, z), \tag{6}$$

$$v = f_v(x, y, z),\tag{7}$$

$$w = f_w(x, y, z), \tag{8}$$



Fig. 4. Estimation processes of in-plane deformation. How initial coarse deformation fields are provided is illustrated.



4D US vessel model

Fig. 5. Reconstruction of 4D US model. See Movies 1, 2, and 3 in Appendix.

where $\mathbf{u} = (u, v, w)$ and $\mathbf{x} = (x, y, z)$, $f_u(x, y, z)$, $f_v(x, y, z)$, and $f_w(x, y, z)$ are described using multilevel B-splines (Lee et al., 1997). The problem is regarded as finding multilevel B-splines coefficients at each frame t' minimizing

$$\sum_{\mathbf{p},i} (U_i(\mathbf{p},t') - f_u(X_i(\mathbf{p},0), Y_i(\mathbf{p},0), Z_i(\mathbf{p},0)))^2,$$
(9)

$$\sum_{\mathbf{p},i} (V_i(\mathbf{p},t') - f_v(X_i(\mathbf{p},0), Y_i(\mathbf{p},0), Z_i(\mathbf{p},0)))^2,$$
(10)

$$\sum_{\mathbf{p},i} (W_i(\mathbf{p},t') - f_w(X_i(\mathbf{p},0), Y_i(\mathbf{p},0), Z_i(\mathbf{p},0)))^2,$$
(11)

where $\mathbf{X}_{i}(\mathbf{p}, 0) = (X_{i}(\mathbf{p}, 0), Y_{i}(\mathbf{p}, 0), Z_{i}(\mathbf{p}, 0)), \quad \mathbf{U}_{i}(\mathbf{p}, t') = (U_{i}(\mathbf{p}, t'), V_{i}(\mathbf{p}, t'), W_{i}(\mathbf{p}, t')), \text{ and } \mathbf{p} = \{(0, 0), (0, 1), (0, 2), \ldots, \}$

 $(N_x - 1, N_y - 1)$, in which we assume that the matrix size of US images is $N_x \times N_y$.

2.4.3. Recovery of 4D ultrasound model

3D deformation field $\mathbf{u}(\mathbf{x})$ is recovered at each frame t'using the method described in Section 2.4.2. Let $\mathbf{u}(\mathbf{x})$ at t'th frame be $\mathbf{u}(\mathbf{x}, t')$ ($t' = 0, ..., \overline{L} - 1$), which represents 4D deformation field. Let \mathbf{V}_k ($k = 1, ..., \overline{N}_v$) be the vertices of a 3D polygon model reconstructed from static 3D US images acquired at t' = 0, where N_v is the number of vertices of the model. The time-varying 3D vertex positions of the 4D US model are given by $\mathbf{V}_k + \mathbf{u}(\mathbf{V}_k, t')$.

3. Experimental results

3.1. Experimental conditions

We performed *in vivo* experiments using a pig under laparoscopic control to evaluate the proposed method. The animal was under general anesthesia with breathing controlled by a respirator whose ventilation volume was 400 cm^3 . SSD-5500 (ALOKA Co., Japan) ultrasound system and UST-5536-7.5 (ALOKA Co., Japan) LUS probe were employed. The field of view of the US images was $27 \times 56 \text{ mm}^2$, and the frame rate was 30 frames per second (fps). microBIRD (Ascension Technology Co., Burlington, VT) was used as a six degree-of-freedom magnetic tracker to track the LUS probe tip. Polaris (Northern Digital Inc., Waterloo, Ontario, Canada) was employed for correcting magnetic field distortion of microBIRD by using a magneto-optic hybrid tracker (Nakada et al., 2003). Fig. 6 shows the setup for *in vivo* experiments.

We acquired two datasets around the porta hepatic of the pig, which covered the volumes of $27 \times 56 \times 60 \text{ mm}^3$ and $27 \times 56 \times 90 \text{ mm}^3$, respectively. These datasets consisted of six and seven 2D-US sagittal planes of average intervals of 12.6 mm and 15.7 mm, respectively. Parameters used for the estimation of in-plane deformation and recov-



Fig. 6. Experimental setup.

ery of continuous 3D deformation field are summarized in Tables 1 and 2, respectively. The frame rate of the reconstructed 4D model was reduced from 30 fps to 15 fps due to the limitation of memory space. Computation time for estimation of in-plane deformation and reconstruction of the 4D model were around 20 min and 10 min (Xeon $3.0 \text{ GHz} \times 2$, 2 GB memory), respectively.

3.2. Evaluation of temporal registration

Fig. 7 shows time-varying similarity in each plane before and after temporal registration. The origin of time in Fig. 7a represents the start of data acquisition. Because data acquisition started from different time points in inspiration phase, EPIs were not aligned before temporal registration. After the registration, EPIs were aligned (point A in Fig. 7b). It was observed that time-varying similarities of different planes had similar characteristics, that is, peaks located at the end of inspiration and plateaus around the start of inspiration. We confirmed that the reference points could be localized in sufficient accuracy and stability.

To evaluate the reproducibility of respiratory phase analysis, average and standard deviation of cycle length and interval from EPI to SPI in different planes were calculated, as summarized in Table 3. The standard deviations of them were 0.12 s or less, 2% of cycle length. The difference in respiratory cycle length was quite small between Datasets 1 and 2.

We compared the respiratory cycle estimated using US image sequences with that obtained by optical marker tracking. Optical markers were attached to abdominal skin of the pig, and tracked by Polaris (Fig. 6). US image acquisition and optical marker tracking were performed simultaneously. Because the primary component of marker motion was in the anterior–posterior direction, the time point where time-varying position along the anterior–posterior

Table 1

Experimental conditions of non-rigid registration for estimating in-plane deformation

| Image matrix size | 93×190 |
|---|----------------------|
| Pixel size (mm ²) | 0.298×0.298 |
| λ | 0.05 |
| Initial control point grid spacing (mm) | 5.7 |
| Refined control point grid spacing (mm) | 2.8 |

 λ is the weighting parameter which defines the tradeoff between the alignment of the two images and the smoothness of the transformation

Table 2

Experimental conditions of multilevel B-spline interpolation for recovery of continuous 3D deformation field

| | Dataset 1 | Dataset 2 |
|---|--------------------------------|-----------------|
| Range of interpolation (mm ³) | $78.8 \times 72.2 \times 64.3$ | 73.8×158.1×92.2 |
| Number of grid refinement step | 6 | 6 |
| Initial grid spacing (mm) | 84.0 | 168.7 |
| Final grid spacing (mm) | 1.3 | 2.6 |



Fig. 7. Temporal registration (Dataset 1).

Table 3 Respiration cycle analysis

| 1 | <i>.</i> | | |
|-----------|------------------|------------------------------|-----------------------------|
| | Number of planes | Respiratory cycle length (s) | Interval from A to C (s) |
| Dataset 1 | 6 | 6.0 ± 0.01 | 2.6 ± 0.05 |
| Dataset 2 | 7 | 6.0 ± 0.02 | 3.0 ± 0.12 |

direction reached the local maximum was defined as EPI by optical marker tracking. We regarded the time point obtained by the above method as a bronze standard for comparison. Fig. 8 shows the difference between estimated EPI and bronze standard at each plane of Dataset 1. Average and standard deviation of the differences were 0.11 ± 0.06 s. The cycles estimated using US images were in good agreement with these obtained by optical marker tracking.

3.3. Evaluation of in-plane deformation estimation

Fig. 9 shows representative original images, deformed images by estimated deformation field, and their compari-



Fig. 8. Comparison between the proposed method and marker tracking method to determine EPI.

sons. Hepatic region in the reference images at frame t' = 0was segmented and non-hepatic area was filled by zero intensity. Also in the deformed image, zero intensity was filled outside the deformed version of hepatic region. In column (c), subtraction images between original (in column (a)) and deformed (in column (b)) images are shown. In the subtraction images, intensities inside deformed hepatic region were nearly zero. In column (d), checkerboard display of the original and the deformed images are shown. Discontinuity was not observed even around characteristic areas such as vessels depicted as low intensity spots. Columns (c) and (d) demonstrate that the deformed images were well aligned to the original ones. In column (e), estimated deformation vectors are shown as white lines. The directions of deformation vectors changed smoothly, and turbulent flow was not observed.

We quantitatively evaluated the estimation of in-plane deformation. Vessels in the liver depicted as black spots in 2D US images were employed to measure estimation error. The error at frame t' was defined as the average distance between the points, which were the centroids of black spots, in the original image $I_i(\mathbf{p}, t')$ and their corresponding points in the deformed image $F(I_i(\mathbf{p}, 0), \mathbf{m}_i(\mathbf{p}, t'))$. Nine and six points, which were found in all the images in dynamic phase, were manually selected from Datasets 1 and 2, respectively. Fig. 10 shows the time series of the error. EPI was located at 1.9 s along the time axis in Fig. 10. Average errors over time of Datasets 1 and 2 were 0.3 and 0.5 mm, respectively. The error tended to increase during inspiration phase and decrease during expiration phase (Fig. 10). The maximum errors of Datasets 1 and 2, which were observed around EPI, were 0.8 and 1.1 mm, respectively.

3.4. Recovery of motion and deformation

Fig. 11a shows the magnitude of motion and deformation of the pig liver recovered from Dataset 1. The estimated respiratory cycle duration was 6.0 s. EPI was



Fig. 9. Visual assessment of estimation of in-plane deformation (Dataset 1): (a) original image; (b) deformed image; (c) subtraction image between original and deformed images; (d) checkerboard display to facilitate intuitive understanding of registration accuracy of original and deformed images; and (e) estimated deformation vectors on deformed images. See Movie 4 in Appendix.

located at 2.6 s along the time axis. The magnitude of recovered motion was the largest in the cranio-caudal, second largest in the anterior–posterior, and the smallest in

the left-right directions, with maximum at EPI in all three directions. Table 4 shows the maximum magnitude of motion in each direction.



Fig. 10. Accuracy of estimated in-plane deformation. EPI was located at 1.9 s along the time axis.



(a) Magnitude of recovered motion of the liver along the body axes. C-C: cranio-caudal direction. A-P: anterior-posterior direction. L-R: left-right direction.



(b) Evaluation of interpolation accuracy in recovered continuous deformation field.

Fig. 11. Recovered motion of the liver and evaluation of interpolation accuracy (Dataset 1). EPI was located at 2.6 s along the time axis.

The recovered 4D deformation fields were combined with static 3D US data of the same regions as those covered by the datasets acquired for recovery of motion and

| Table 4 | |
|--|---|
| Maximum magnitude of liver motion (unit: mm) | |
| | 1 |

| | C-C | A–P | L–R |
|-----------|-----|------|-----|
| Dataset 1 | 7.7 | 10.7 | 1.4 |
| Dataset 2 | 7.7 | 7.4 | 1.6 |
| | | | |

deformation in order to reconstruct the 4D US data. Fig. 12a shows the static 3D liver vessel model reconstructed from 3D US images and US image frames of sagittal planes used for recovery of motion and deformation. 3D US images were acquired at expiration phase during breath holding. Fig. 12b shows the 4D liver vessel model reconstructed by combining the 3D liver vessel model and the recovered 4D deformation field. Distinctive motion was observed from the lateral view, where vessels moved up and down along different paths during expiration and inspiration phases, respectively.

3.5. Accuracy evaluation of recovered 4D deformation field

To evaluate the effects of intervals of sagittal planes on interpolation accuracy, we performed leave-*N*-out cross validation. Recovery of continuous 4D deformation field was performed by removing selected *N* planes and interpolating 3D displacement vectors on remaining planes. The 3D displacement vectors at frame t' on the selected *N* planes were used for validation by comparing them with corresponding vectors in recovered continuous 4D deformation field of the whole volume.

Validation planes, that is, removed planes, were selected based on three conditions; (I) a single plane, (II) every one plane, and (III) two consecutive planes (upper row in Fig. 13). The average interval was the smallest in condition (I) and the largest in condition (III) (middle row in Fig. 13). The error $E_{i,t'}$ was defined as

$$E_{i,t'} = \sqrt{\frac{\sum_{x,y,z \in \mathcal{R}_i} |\mathbf{U}_{val}(x,y,z,t') - \mathbf{U}_{est}(x,y,z,t')|^2}{n}},$$
(12)

where $n = \sum_{x,y,z \in R_i} 1$, and R_i was the point set on the selected planes. $\mathbf{U}_{val}(x, y, z, t)$ and $\mathbf{U}_{est}(x, y, z, t)$ were original and recovered deformation vectors at position (x, y, z) and frame t', respectively. Here, the original deformation vectors represent the 3D displacement vectors obtained in Section 2.4.1 while the recovered ones were obtained by multilevel B-spline interpolation described in Section 2.4.2.

Table 5 summarizes the results. The maximum error under condition (II) during one cycle, which was around 2.0 mm, was comparable to that under condition (I). It was larger under condition (III) while there was no significant difference in the average error among the three conditions. Fig. 11b shows the time series of the error of each condition. The maximum error was observed around EPI. The magnitude of error along time was correlated to that of respiratory motion, and the error remained at



(a) 3D US liver vessel model and US image frames of sagittal planes.



(b) Reconstructed 4D liver vessel model. Upper row: Frontal views. Lower row: Lateral views. From left to right, 3D models at different time frames are shown.

Fig. 12. Reconstructed 3D and 4D liver vessel models (Dataset 1). See Movies 5 and 6 in Appendix.



Fig. 13. Effect of estimation error of 3D deformation field on recovered 4D model (Dataset 1). Upper row: remained and removed sagittal planes are depicted by blue and red, respectively. Middle row: intervals caused by removed planes. Lower row: differences in recovered motion.

Table 5 Evaluation on interpolation accuracy in recovery of continuous deformation field of whole volume (unit: mm)

| | Condition | Ι | II | III |
|-----------|------------------|------|------|------|
| Dataset 1 | Average interval | 15.8 | 25.2 | 31.3 |
| | Maximum error | 1.9 | 1.8 | 2.2 |
| | Average error | 1.3 | 1.2 | 1.2 |
| Dataset 2 | Average interval | 15.7 | 18.8 | 42.2 |
| | Maximum error | 2.4 | 2.3 | 2.7 |
| | Average error | 1.0 | 1.0 | 1.1 |

1.0–1.5 mm at the end of cycle. There was no significant difference between under condition (I) and under condition (II) in the range from the beginning to 4.0 s. The maximum error was observed around EPI under condition (III). The lower row of Fig. 13 shows the difference (light colored) in the recovered 4D vessel models at EPI between deformation fields interpolated using all the planes and those under the three conditions described above. It was observed around the position of removed planes and it was relatively small along the left–right direction. We confirmed that similar results were observed in Dataset 2.

4. Discussion

4.1. Main features of the proposed method

The proposed method extends the laparoscopic 3D US system that we have developed (Nakamoto et al., 2002) to 4D without any additional instrument and equipment. An additional requirement of data acquisition for recovery of a 4D US model is to acquire 2D US images at several sagittal planes during two or three respiratory cycles, which typically takes a couple of minutes. Accuracy evaluation of recovered 4D deformation field shows that the error caused by interpolation is around 2 mm even at EPI only using 2D US images on three additional planes (Table 5). Although computation time, which is around 30 min in the current implementation, is not practical for clinical application, it may be significantly reduced by introducing parallel processing and/or GPU-based technologies (Ino et al., 2005; Ino et al., 2006; Levin et al., 2005; Vetter et al., 2007).

The volume of interest covered by the proposed method is around $27 \times 56 \times 60-90 \text{ mm}^3$ in our experiments. The reconstructed volume is much smaller as compared with those using MRI-based methods (Rohlfing et al., 2004; Blackall et al., 2005; Tokuda et al., 2006; von Siebenthal et al., 2007) that are able to reconstruct 4D model of the entire liver. In actual clinical situation, however, intraoperative ultrasound imaging is employed to confirm internal critical structures such as tumors and blood vessels within relatively narrow areas directly related to resection of the liver. Therefore, the volume of interest covered by the proposed method should be acceptable.

During acquisition of time-varying 2D US images of the sagittal planes, the tip of the LUS probe should be moved

together with the liver surface by keeping moderate contact pressure. Because the LUS probe is inserted into abdominal cavity through a trocar, a question is raised as to whether the trocar disturbs the above described movements. Around twenty LUS datasets acquired in our in vivo experiments under laparoscopic control show that the trocar does not disturb the probe movement. In order to acquire time-varying 2D US images successfully as described above, a surgeon needs to take account of the following points: (1) The LUS probe tip should be kept in contact with hepatic surface by moderate pressure so as not to disturb the motion of the organ; (2) The LUS probe should be supported so that it may move concurrently with the liver; and (3) Trocars should be arranged so as not to disturb the LUS probe movement. In our experiments, the participating surgeon adjusts to the above requirements after some trials.

Another issue is that the LUS probe should be aligned on (quasi-)sagittal planes. If the arrangement of trocars is not appropriate, it may be difficult to align the LUS probe tip to the cranio-caudal direction. In laparoscopic surgery of the liver, trocars are located at lower positions from the liver, and thus surgical instruments are inserted from the caudal side toward the cranial side. Therefore, the inserted LUS probe tip is naturally aligned to the craniocaudal direction.

4.2. Validity of recovered motion and deformation

The proposed method assumes that (1) deformation is cyclic and reproducible, and (2) deformation along the left–right direction is negligible.

Regarding the first assumption, we have confirmed that variations of interval lengths among the reference points in respiration phase analysis are less than 5% of cycle length in most cases and intra-plane variation of respiratory cycle length is less than 3% of cycle length in all cases. These results suggest that respiratory cycle length is almost constant and cyclic. Time-varying similarity also shows a cyclic nature (Fig. 7b). This suggests that similar deformation is repeated in each cycle. In order to confirm the reproducibility of respiratory motion, we perform preliminary pig experiments. Several marks are indicated on hepatic surface of the animal and we repeatedly measure their 3D positions using an optical 3D digitizer at expiration point. Respiration is suspended during the measurement and started again after. This procedure is repeated. The repeatability in position is constantly around 3 mm under laparoscopic control while around 6 mm and increasing as time passed in open surgery of the same pig. We consider that 3 mm error is not large if uncertainty of the measurement method itself is taken into account. The abdomen is filled with gas under laparoscopic control, and thus the organs are in a more stable condition due to gas pressure as compared with in open surgery. Therefore, we consider that the first assumption is valid under laparoscopic control as well as in controlled respiration condition.

Regarding the second assumption, the magnitude of motion component along the left-right direction, which can be recovered by magnetic tracking of the LUS probe on hepatic surface, is 1–2 mm (Table 4) and is regarded as small. This result is consistent with the second assumption. Visual assessment of deformed images (Fig. 9) and registration error of landmarks (Fig. 10) suggest that estimation of in-plane deformation works well. These results also imply that deformation along the left-right direction is small because if large deformation along the left-right direction occurs, tracked and unwanted landmarks disappear and appear, respectively, as the frame progresses. These changes in US images cause failure in non-rigid registration.

The motion of the liver has been investigated by several studies (Rohlfing et al., 2004; Davies et al., 1994; Brandner et al., 2006; Beddar et al., 2007). It has been reported that the main components of hepatic motion are in the craniocaudal and anterior-posterior directions and the motion along the left-right direction is small. The recovered 4D model in our experiments shows similar tendency as indicated in Table 4. Our object in the present study is a pig liver while the targets in previous reports (Rohlfing et al., 2004; Davies et al., 1994; Brandner et al., 2006; Beddar et al., 2007) are human ones. However, the physical relationships among the lung, the diaphragm, and the liver in pig, which mainly influence respiratory motion of the liver, is similar to those in human. Therefore, the proposed method is potentially applicable in clinical practice.

4.3. Issues on respiratory phase analysis

The proposed method employs temporal registration algorithm based on analysis of time-varying similarity between adjacent frames to align datasets acquired on different planes. Because EPI is used to determine a reference point for registration and respiratory cycle length, the accuracy of EPI localization is important. We confirm that the peak of time-varying similarity can be accurately located as EPI and datasets can be aligned using EPIs in all cases as shown in Fig. 7b. The standard deviation of obtained cycle lengths is small enough.

Methods for measuring respiratory motion are reported (Vedam et al., 2003; Tsunashima et al., 2004; Xu and Hamilton, 2006). Unlike those studies, the proposed method detects SPD and EPD in order to determine the interval where estimation of in-plane deformation is required. The interval length between SPD and EPD is around a half of respiration cycle, and thus computation time for estimating in-plane deformation increases two times if SPD and EPD are not determined.

When 4D model overlays onto live laparoscopic view, synchronization between the 4D model and patient's respiration is required. In order to detect patient's respiratory phase, tracking fiducial markers attaching to the skin has been reported (Vedam et al., 2003; Beddar et al., 2007). Our experimental results show that EPIs obtained by the proposed method are well correlated to those acquired by marker tracking (Fig. 8). Because it is easy to integrate fiducial marker tracking to surgical navigation systems, synchronized 4D augmented reality visualization can be developed straightforwardly.

4.4. Possible applications

Two possible applications of the 4D liver model are considered. One is stabilization of surgical robotic manipulators. If a respiratory phase can be provided to a surgical robotic manipulator system in realtime, it would be possible to stabilize the manipulator against respiratory motion of the liver by using the 4D model as well as stabilization against heartbeat (Nakamura and Kishi, 2001). The other is to move and deform a virtual liver in a laparoscopic surgical simulator along respiratory cycle. If the 4D liver model is incorporated to the surgical simulator, it is possible to simulate realistic respiratory motion and deformation of the liver in the virtual environment of the simulator. Thus, the 4D liver model acquired in clinical situation would contribute to the improvement of reality of surgical simulators.

5. Conclusion

We have described an intraoperative recovery method of respiratory motion and deformation of the liver using a laparoscopic freehand 3D US system. 4D US images of the liver are recovered using static 3D US image of the organ and time-varying 2D US images acquired on several sagittal planes. The proposed method is evaluated by in vivo experiments using a pig liver. Datasets acquired on different planes are aligned accurately by temporal registration based on respiratory phase analysis. Non-rigid registration for estimation of in-plane deformation is stable and its accuracy is less than 1 mm on average. The characteristics of recovered motion are qualitatively consistent with those reported previously. Future work will include accuracy evaluation of recovered 4D model using markers embedded in the pig liver, augmented reality visualization of 4D US model synchronized with respiration, and reconstruction of a multi-modal 4D model by combining preoperative 3D CT or MR images and intraoperative deformation field.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.media.2007. 07.009.

References

Bao, P., Sinha, T.K., Chen, C.-C.R., Warmath, J.R., Galloway, R.L., Herline, A.J., 2007. A prototype ultrasound-guided laparoscopic radiofrequency ablation system. Surgical Endoscopy 21 (1), 74–79.

- Beddar, A.S., Kainz, K., Briere, T.M., Tsunashima, Y., Pan, T., Prado, K., Mohan, R., Gillin, M., Krishnan, S., 2007. Correlation between internal fiducial tumor motion and external marker motion for liver tumors imaged with 4D-CT. International Journal of Radiation Oncology Biology Physics 67 (2), 630–638.
- Blackall, J.M., Penney, G.P., King, A.P., Hawkes, D.J., 2005. Alignment of sparse freehand 3-D ultrasound with preoperative images of the liver using models of respiratory motion and deformation. IEEE Transactions on Medical Imaging 24 (11), 1405–1416.
- Boctor, E., Fischer, G., Choti, M., Fichtinger, G., Taylor, R., 2004. A dual-armed robotic system for intraoperative ultrasound guided hepatic ablative therapy: a prospective study. In: Proceedings of IEEE International Conference on Robotics and Automation, vol. 3, pp. 2517–2522.
- Boctor, E., deOliveira, M., Choti, M., Ghanem, R., Taylor, R., Hager, G., Fichtinger, G., 2006. Ultrasound monitoring of tissue ablation via deformation model and shape priors. In: Lecture Notes in Computer Science (Proceedings of MICCAI 2006, Part II), vol. 4191, pp. 405– 412.
- Brandner, E.D., Wu, A., Chen, H., Heron, D., Kalnicki, S., Komanduri, K., Gerszten, K., Burton, S., Ahmed, I., Shou, Z., 2006. Abdominal organ motion measured using 4D CT. International Journal of Radiation Oncology Biology Physics 65 (2), 554–560.
- Davies, S., Hill, A., Holmes, R.B., Halliwell, M., Jackson, P., 1994. Ultrasound quantitation of respiratory organ motion in the upper abdomen. British Journal of Radiology 67 (803), 1096–1102.
- Ellsmere, J., Stoll, J., William Wells, I., Kikinis, R., Vosburgh, K., Kane, R., Brooks, D., Rattner, D., 2004. A new visualization technique for laparoscopic ultrasonography. Surgery 136 (1), 84–92.
- Harms, J., Feussner, H., Baumgartner, M., Schneider, A., Donhauser, M., Wessels, G., 2001. Three-dimensional navigated laparoscopic ultrasonography: first experiences with a new minimally invasive diagnostic device. Surgical Endoscopy 15 (12), 1459–1462.
- Hong, J., Dohi, T., Hashizume, M., Konishi, K., Hata, N., 2004. An ultrasound-driven needle-insertion robot for percutaneous cholecystostomy. Physics in Medicine and Biology 49 (3), 441–455.
- Ino, F., Ooyama, K., Hagihara, K., 2005. A data distributed parallel algorithm for nonrigid image registration. Parallel Computing 31 (1), 19–43.
- Ino, F., Kawasaki, Y., Tashiro, T., Nakajima, Y., Sato, Y., Tamura, S., Hagihara, K., 2006. A parallel implementation of 2-D/3-D image registration for computer-assisted surgery. International Journal of Bioinformatics Research and Applications 2 (4), 341–358.
- Kleemann, M., Hildebrand, P., Birth, M., Bruch, H.P., 2006. Laparoscopic ultrasound navigation in liver surgery: technical aspects and accuracy. Surgical Endoscopy 20, 726–729.
- Lee, S., Wolberg, G., Shin, S.Y., 1997. Scattered data interpolation with multilevel B-splines. IEEE Transactions on Visualization and Computer Graphics 3 (3), 228–244.
- Leven, J., Burschka, D., Kumar, R., Zhang, G., Blumenkranz, S., Dai, X.D., Awad, M., Hager, G.D., Marohn, M., Choti, M., Hasser, C.J., Taylor, R.H., 2005. DaVinci Canvas: A telerobotic surgical system with integrated, robot-assisted, laparoscopic ultrasound capability. In: Lecture Notes in Computer Science (Proceedings of MICCAI 2005, Part I), vol. 3749, Palm Springs, CA, pp. 811–818.
- Levin, D., Dey, D., Slomka, P., 2005. Efficient 3D nonlinear warping of computed tomography: Two high performance implementations using OpenGL. In: Progress in Biomedical Optics and Imaging – Proceedings of SPIE, vol. 5744, pp. 34–42.

- Nakada, K., Nakamoto, M., Sato, Y., Konishi, K., Hashizume, M., Tamura, S., 2003. A rapid method for magnetic tracker calibration using a magneto-optic hybrid tracker. In: Lecture Notes in Computer Science (Proceedings of MICCAI 2003, Part II), vol. 2879, Montreal, Canada, pp. 285–293.
- Nakamoto, M., Sato, Y., Miyamoto, M., Nakajima, Y., Konishi, K., Shimada, M., Hashizume, M., Tamura, S., 2002. 3D ultrasound system using a magneto-optic hybrid tracker for augmented reality visualization in laparoscopic liver surgery. In: Lecture Notes in Computer Science (Proceedings of MICCAI 2002, Part II), vol. 2489, Tokyo, Japan, pp. 148–155.
- Nakamura, Y., Kishi, K., 2001. Robotic stabilization that assists cardiac surgery on beating hearts. Studies in Health Technology and Informatics 81, 355–361.
- Olbrich, B., Traub, J., Wiesner, S., Wiechert, A., Feußner, H., Navab, N., 2005. Respiratory motion analysis: towards gated augmentation of the liver. In: Proceedings of Computer Assisted Radiology and Surgery (CARS 2005), Berlin, Germany, pp. 248–253.
- Rohlfing, T., Maurer Jr., C.R., O'Dell, W.G., Zhong, J., 2004. Modeling liver motion and deformation during the respiratory cycle using intensity-based nonrigid registration of gated MR images. Medical Physics 31, 427–432.
- Rueckert, D., Sonoda, L.I., Hayes, C., Hill, D.L.G., Leach, M.O., Hawkes, D.J., 1999. Nonrigid registration using free-form deformations: application to breast MR images. IEEE Transactions on Medical Imaging 18, 712–721.
- Sato, Y., Nakamoto, M., Tamaki, Y., Sasama, T., Sakita, I., Nakajima, Y., Tamura, S., Monden, M., 1998. Image guidance of breast cancer surgery using 3-D ultrasound images and augmented reality visualization. IEEE Transactions on Medical Imaging 17 (5), 681–693.
- Sjølie, E., Langø, T., Ystgaard, B., Tangen, G., Hernes, T., Marvik, R., 2003. 3D ultrasound-based navigation for radiofrequency thermal ablation in the treatment of liver malignancies. Surgical Endoscopy 17 (6), 933–938.
- Tokuda, J., Morikawa, S., Haque, H.A., Tsukamoto, T., Matsumiya, K., Liao, K. Masamune, H., Dohi, T., 2006. New 4-D imaging for realtime intraoperative MRI: Adaptive 4-D scan. In: Lecture Notes in Computer Science (Proceedings of MICCAI 2006, Part I), vol. 4190, Copenhagen, Denmark, pp. 454–461.
- Tsunashima, Y., Sakae, T., Shioyama, Y., Kagei, K., Terunuma, T., Nohtomi, A., Akine, Y., 2004. Correlation between the respiratory waveform measured using a respiratory sensor and 3D tumor motion in gated radiotherapy. International Journal of Radiation Oncology Biology Physics 60 (3), 951–958.
- Vedam, S.S., Kini, V.R., Keall, P.J., Ramakrishnan, V., Mostafavi, H., Mohan, R., 2003. Quantifying the predictability of diaphragm motion during respiration with a noninvasive external marker. Medical Physics 30 (4), 505–513.
- Vetter, C., Guetter, C., Xu, C., 2007. Non-rigid multi-modal registration on the GPU. In: Proceedings of SPIE Medical Imaging 2007: Image Processing, vol. 6512, p. 651228.
- von Siebenthal, M., Szekery, G., Gamper, U., Boesiger, P., Lomax, A., Cattin, P., 2007. 4D MR imaging of respiratory organ motion and its variability. Physics in Medicine and Biology 52, 1547– 1564.
- Xu, Q., Hamilton, R.J., 2006. A novel respiratory detection method based on automated analysis of ultrasound diaphragm video. Medical Physics 33 (4), 916–921.