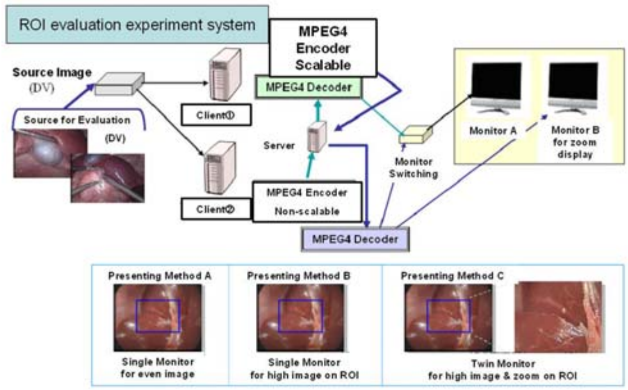

Fig. 1 ROI evaluation experiment system

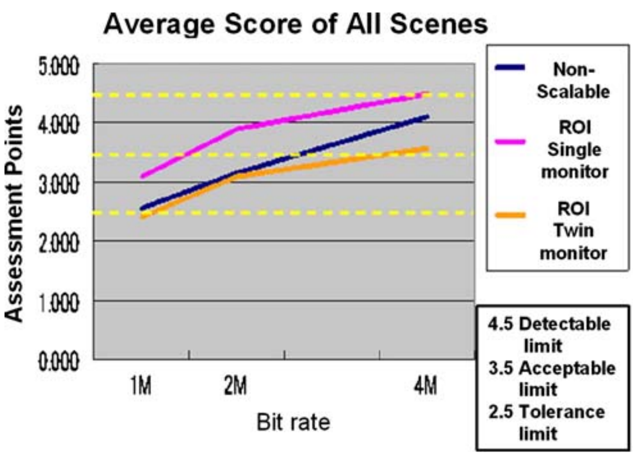

Fig. 2 Assessment points

training images. Next, 20 scenes were selected from the 80 scenes by the chief trainer of the minimally invasive training center at Kyushu University Hospital. Approximately 10 scenes were cut from the 20 nominations. Finally, 4 scenes were selected by the chief trainer. Seven medical doctors took part in evaluating the images in these experiments. They performed a five-grade evaluation using each method. One Mbps, 2 Mbps, and 4 Mbps were used as parameters. The ROI evaluation experiment system is shown in Fig. 1. Nonscalable (A), ROI single monitor (B), and ROI twin monitors (C) were used as the presenting methods.

Results

The relationships between the bit rate and assessment points for each presentation method are shown in Fig. 2. Assessment points represent all average scores in all scenes for all examinees. The ROI single monitor presenting method scored 0.5 points higher than the non-scalable presenting method. The ROI twin monitor presenting method was given the lowest evaluation.

Conclusion

The ROI single monitor presenting method was the most suitable for tele-mentoring, including clinical surgical instructions.

Temporal calibration of tracking and image acquisition

Güler Ö.[1], Yaniv Z.[2], Freysinger W.[1], Cleary K.[2]
[1]Medical University Innsbruck, ENT Clinic, Innsbruck, Austria
[2]Imaging Science and Information Systems (ISIS) Center, Dept. of Radiology, Georgetown University Medical Center, Washington, DC, USA

Keywords CAS · Navigation · Temporal calibration

Purpose

Two common intra-operative imaging modalities utilized in image-guided procedures are ultrasound (US) and endoscopic video. These modalities provide realtime information that can be augmented with data derived from pre-operative images such as CT and MR. To fuse the pre-operative and intra-operative data we need to establish the pose of the intra-operative image relative to the pre-operative data. To enable this, the intra-operative imaging apparatus is calibrated so that the image's spatial location relative to a tracked reference frame mounted on the device is estimated. We can then position the realtime image in its correct location relative to the pre-operative data.

This approach is valid if the latency between the acquisition of the tracking information and intra-operative imaging is negligible, or if the imaging device is stationary. In many procedures this is not the case, images are acquired with a moving device and latency between the image and transformation streams is not negligible. As a result, a temporal calibration step is required. Once the latency between the two data streams is known, the realtime images can be positioned using the correct spatial transformation.

In this work we describe a temporal calibration approach that does not require prior spatial calibration. It is thus applicable both for perspective images such as endoscopic video, and tomographic images such as US.

Methods

To perform temporal calibration a tracked object is moved such that its motion in the acquired images and in three dimensional space describes a back and forth motion along a linear trajectory. We concurrently acquire transformation and image streams from the tracker and imaging apparatus while the motion is performed. For the US case, we move a tracked needle in a water bath with the needle shaft intersecting the image plane, as shown in Fig. 1. For the video case, we place a planar circular fiducial marker in front of the camera so that its motion is parallel to the image plane. Each of the transformations and images is time stamped as it is acquired.

The result of our data acquisition is a set of two dimensional points from the image data and a set of three dimensional points from the tracking data. As most of the motion is along the linear trajectory we extract a one dimensional (1D) signal from each of the data sets via Principle Component Analysis. Our 1D signal is the projection of each of the point sets onto their principle axis. The maximal correlation between these two signals yields the latency.

In this work we implemented this approach only in the context of US imaging and electromagnetic tracking (Fig. 2). Images were

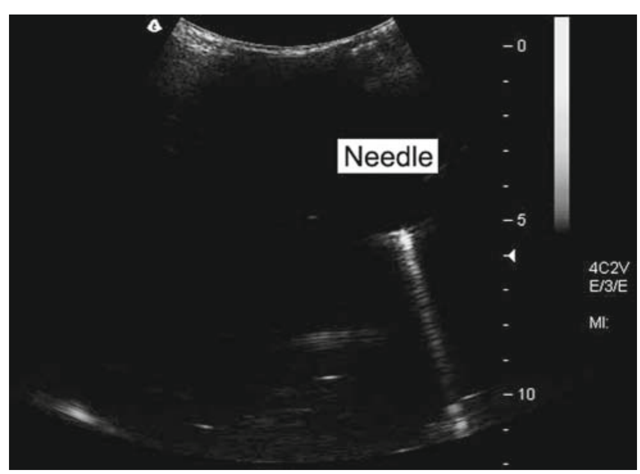

Fig. 1 Ultrasound image showing tracked

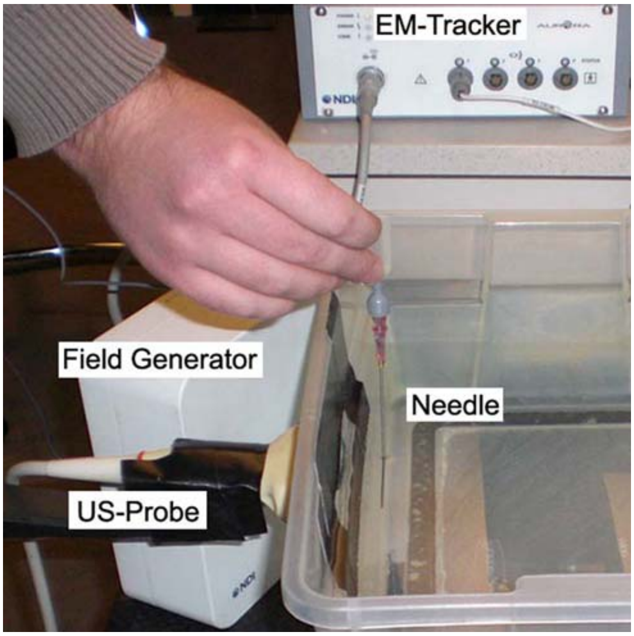

Fig. 2 Experimental setup

acquired with the Terason T2000 portable US system (Teratech Corp., Burlington MA, USA), and tracking with the Aurora tracking system (Northern Digital Inc., Waterloo Ontario, Canada). The ultrasound images were acquired at a rate of 15 Hz with tracking data acquired at a rate of 40 Hz for a period of 20 sec.

Results

The experiments show a remarkable delay between the two signals. Therefore it is necessary to perform a temporal calibration before intervention. The period of 20 seconds is enough and can be decreased to achieve faster calibration results. Ongoing experiments will show the influence of device overheating on the latency between the signals.

Conclusion

We have presented initial work on temporal calibration between image and tracking data. The proposed approach is applicable both for US and perspective images such as video endoscopy. The method is currently being incorporated into the open source Image-Guided Surgery Toolkit (IGSTK).

Acknowledgement

This research was funded by FWF Austrian Science Fund: Project-Nr.: 20604-B13.

3-D measurement for minimally invasive surgery by structured light system using kaleidoscope

Namae T.[1], Koishi T.[2], Nakaguchi T.[1], Tsumura N.[1], Miyake Y.[3]
[1]Chiba University, Graduate School of Advanced Integration Science, Chiba, Japan
[2]Chiba University, Graduate School of Science and Technology, Chiba, Japan
[3]Chiba University, Research Center for Frontier Medical Engineering, Chiba, Japan

Keywords Minimally invasive surgery · 3D measurement · Kaleidoscope · Phase shift method

Purpose

Non-contact 3-D measurement plays an important role in various fields. In recent years there have been applications for minimally invasive surgery such as the reconstruction of organ surfaces in the body using structured light system. This system uses an image fiber to guide structured light to the inside of the body, whereas the image fiber needs a large number of cores to obtain dense reconstruction. Consequently, a hole from which the image fiber is inserted becomes large and patient's burdens increase. In this study, we propose a method that projects high-resolution pattern using kaleidoscope.

Methods

We developed a new optical system for measuring 3D shape, which consists of a kaleidoscope, LCD, a convex lens and a camera. The camera captures structured light projected on objects. The kaleidoscope is rectangular solid which diameter is 20 $\times$ 10 mm rectangular bottom and 120 mm height. Figure 1 shows the geometry of our optical system. The convex lens is placed at outgoing side of the kaleidoscope. The display is placed on the incident side of the kaleidoscope to project structured light. Regarding structured light systems, many techniques have been proposed. Phase-shift method is one of the efficient methods because of its dense reconstruction despite requiring only a few patterns. In general, this method project three kinds of sinusoidal patterns which phase shifts $2/3\pi$. Then the phase value can be calculated by capturing these three images. Although typical methods generally project several periods of sinusoidal pattern, our method projects only single period generated by LCD. Second, Third and later periods are formed by the rays reflected inside the kaleidoscope. The phase value obtained in this way has some difference compared with typical phase-shift methods, because the projected pattern that reflects even number of times are symmetric to the projected pattern that reflects odd number times. Accordingly, the phase value is also symmetric. In order to compensate this inversion, we assume that the wave direction of the sinusoidal patterns and the x-axis of the image coordinate system are parallel. Based on this assumption, inverted regions can be specified by the gradient of the phase value.

Results

For the purpose of evaluating the measurement accuracy, we measured a white plane. Then we fit the measured coordinate to an ideal flat plane and calculate RMSE of each period of the projection pattern, which is shown in Fig. 2. Here, *First-cycle* means a region where direct ray is projected. *Near-second-cycle* means the closer region to the camera, where the projected rays reflect only once inside

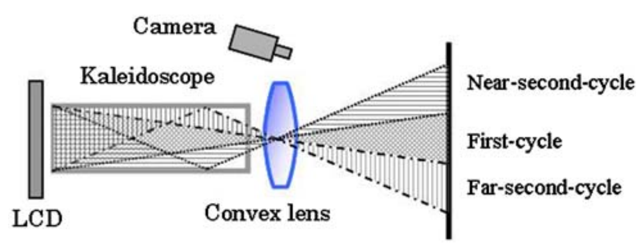

Fig. 1 Geometry of our optical system consisting of kaleidoscope, convex lens, LCD and camera

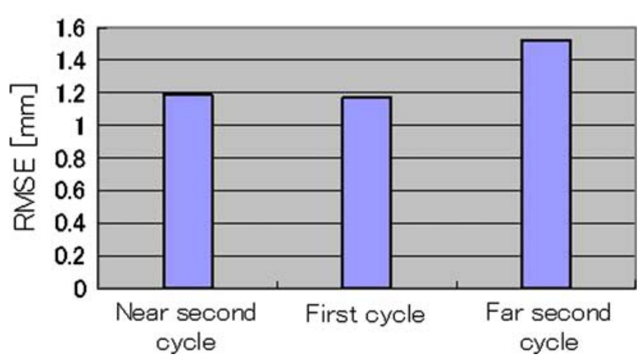

Fig. 2 Measurement error of the proposed method. Error values are calculated by measuring a white plane and fitting the measured coordinates