be reconstructed or not. Recent three-dimensional (3-D) software can calculate not only each segmental volume of liver but also the territorial volume of hepatic vein. The purpose of this study was to evaluate the accuracy and usefulness of preoperative 3-D simulation in living donor liver transplantation.

Methods

From February 2004 to December 2011, 3-D simulation software (Organs Volume Analysis, Hitachi Medico, Tokyo, Japan) was applied to the preoperative volumetric estimation of 189 living donors. 3-D image of the liver were constructed using DICOM data of computed tomography, and the volume of each segment was measured based on auto-segmentation algorithm, which delineates the borderline of segments according to vasculature of portal vein or hepatic vein.[1] Graft type was determined on the basis of the volumetry to ensure sufficient volume of the graft and the liver remnant.[2] When right liver graft was harvested, tributaries of the middle hepatic vein were reconstructed based on the volumetry of hepatic venous territories.[3] Actual graft weight measured after harvesting was compared with estimated graft volume by preoperative 3-D simulation.

Results

Based on the 3-D volumetry, left liver graft (n = 62), right liver graft (n = 100), extended right liver graft (n = 27) were harvested. The shape of the demarcation line between the graft and the liver remnant was properly demonstrated by preoperative 3-D simulation (Fig. 1). The territory of middle hepatic vein which was delineated intraoperatively was also closely similar to the 3-D image (Fig. 2). The estimated graft volume significantly correlated with the actual graft weight ($R^2 = 0.783$, $p < 0.0001$). Neither surgery-related death nor severe postoperative liver dysfunction of donor was observed.

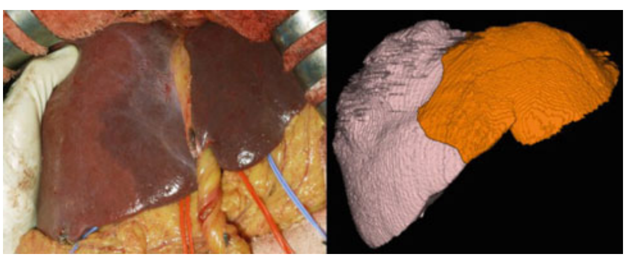

Fig. 1 Intraoperative finding and 3-D image of donor liver

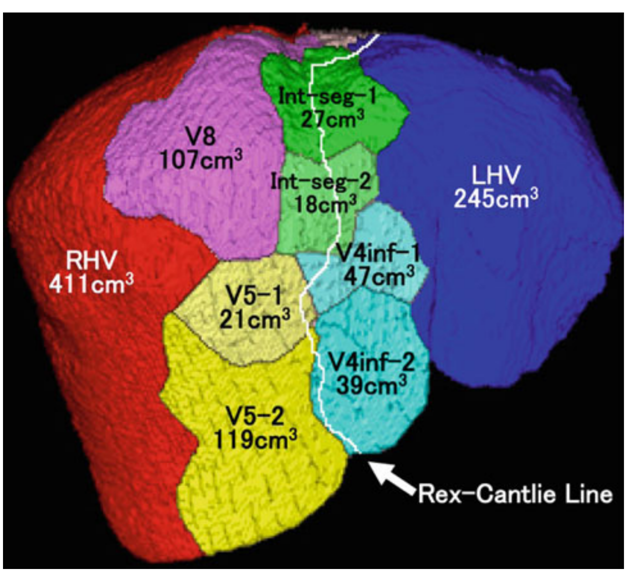

Fig. 2 Hepatic venous volumetry of donor liver

Conclusion

Preoperative 3-D simulation provided accurate volumetric information of living donor liver. This method is very useful for surgical planning of donor hepatectomy and venous reconstruction.

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Accuracy evaluation of an image overlay in an instrument guidance system for laparoscopic liver surgery

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Keywords Laparoscopic surgery · Augmented reality · Liver surgery · Instrument guidance system

Purpose

Benefits of laparoscopic liver surgery with respect to open surgery are well known and include reduced patient trauma and perioperative blood loss. However, video images acquired through an optical device (i.e. endoscope), are the only source of visual guidance into the body cavities. Thus, drawbacks such as the keyhole view of the operating field and 2-dimensional video-optic representation of the operative situs limit the diffusion of this technique [3, 5].

To overcome these limitations, instrument guidance systems dedicated for laparoscopic surgery have been proposed [3]. Through tracking of the laparoscopic camera and instruments, as well as registering the liver to an available image data set (CT, MRI, MeVis), we developed an augmented reality (AR) framework in which the endoscope's video stream is augmented with relevant medical information (i.e. positions of tumors).

In order to provide a clinically applicable instrument guidance system (IGS) for laparoscopic liver surgery, the accuracy of the AR framework plays an important role. This work aims at evaluating the accuracy of the image overlay of the endoscopic image and the preoperative data set.

Methods

An IGS for open liver surgery (CAScination, CH) is extended to integrate a calibrated view of a laparoscopic camera [4]. A standard laparoscopic optics with 30° inclination is used and connected to video camera module (Karl Storz Endoskope, GER). The video signal is integrated into the instrument guidance system. Instrument tracking is provided by an optical tracking system (Polaris Vicra, Northern Digital Inc., Canada), tracking the distal ends of the laparoscopic instruments. MeVis planning data is used as 3D medical image data input, and is registered to the patient through a locally-rigid landmarked-based registration [4]. Then, in order to achieve an accurate image overlay of the planning data and the endoscope video stream, the endoscope camera is calibrated using an optical-tracked, Zhang based calibration. Virtual images are finally rendered using a virtual camera defined by the endoscope's intrinsic and extrinsic parameters.

The calibration process uses a checkerboard composed of a rectangular grid (9 × 7 black-white pattern) attached on a metal plate

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together with four passive markers. While the endoscope remains in the same position, different images of the checkerboard at different angles are acquired together with the checkerboard position with respect to the tracking device coordinate system [6]. The calibration is performed by using Open CV library [7], yielding the intrinsic parameters of the camera and the extrinsic parameters relating the checkerboard coordinate system to the coordinate system of the endoscope camera for each checkerboard position.

In order to track the endoscope, the rigid body transformation from the coordinate system of the tracker attached to the endoscope to the coordinate system of the endoscope camera is required. For each extrinsic parameter, this transformation is obtained by solving the following:

$$\left({}^{\mathrm{endoscope}}\mathrm{T}_{\mathrm{camera}}\right) = \left({}^{\mathrm{tracker}}\mathrm{T}_{\mathrm{endoscope}}\right)^{-1}\cdot\left({}^{\mathrm{tracker}}\mathrm{T}_{\mathrm{checkerboard}}\right)\cdot\left({}^{\mathrm{camera}}\mathrm{T}_{\mathrm{checkerboard}}\right)^{-1}$$

where ${}^{\mathrm{tracker}}\mathrm{T}_{\mathrm{endoscope}}$ is the transform relating the tracking device coordinate system to the endoscope's coordinate system and ${}^{\mathrm{tracker}}\mathrm{T}_{\mathrm{checkerboard}}$ is the transform relating the tracking device coordinate system to the checkerboard's coordinate system (see Fig. 1). These two transforms are given by the optical tracking system. The last transform, ${}^{\mathrm{camera}}\mathrm{T}_{\mathrm{checkerboard}}$ relates the coordinate system of the endoscope camera to the checkerboard's coordinate system and is given by the extrinsic parameters. Since the transformation ${}^{\mathrm{endoscope}}\mathrm{T}_{\mathrm{camera}}$ is a static parameter, the error introduced during the calibration phase yields in variance across the transformations. We selected the transformation minimizing the reprojection error, as the distance between the corners of the grid in the 2D image of the checkerboard and in the 3D checkerboard model, reprojected on the 2D image using the extrinsic parameters [2].

Finally, the virtual camera's point of view is defined by setting its position and intrinsic parameters to those of the endoscope camera. The endoscope image is undistorted using the distortion maps calculated during the calibration procedure and superimposed with the virtual camera view of the 3D image of the planning data [2].

The accuracy of the system was evaluated through two measures. First, in order to evaluate the accuracy of the calibration, the reprojection error was computed for each checkerboard orientation. Since the Camera Calibration Toolbox for MATLAB provides useful and intuitive errors views, and computes the calibration in the same way as the Open CV library, the reprojection error was computed with the former. The uncertainty corresponding to the calibrated extrinsic parameters, computed as three times the standard deviations of the reprojection errors, was also calculated [1, 2].

Then, the overall accuracy of the augmented reality system was evaluated by using a rapid prototyped model of a human liver with a superimposed 1-cm surface grid (Fig. 2a). After calibration of the endoscope camera, augmented reality images were created by superimposing the endoscope image with a 3D image of the model, using 3 different orientations of the endoscope with respect to the model, and 2 distances between the endoscope camera and the model (approximately 2 and 35 cm) [2]. For each AR image, the discrepancies between the grids in the endoscope image and in the 3D image were then measured in 8 different nodes: 4 in the center of the image, and 4 in the borders.

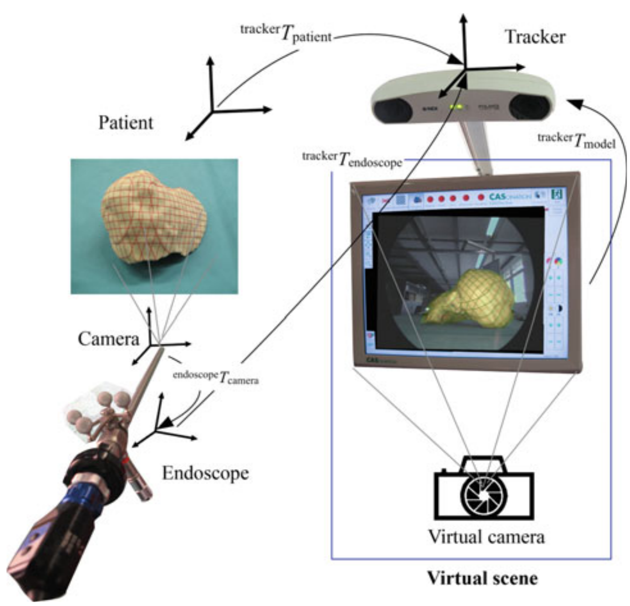

Fig. 1 IGS functional model

Results

Table 1 presents the uncertainty corresponding to the calibrated extrinsic parameters, computed as three times the standard deviations of the reprojection errors.

Figure 3a depicts the error in pixels of each reprojected 2D point for the calibration yielding the lowest reprojection error. The different colors represent the different images used to perform the calibration [1].

Figure 3b shows the box plots related to the misalignment of the superimposition between the endoscope image and the 3D image depicted in Fig. 2b. The minimal and maximal values, as well as the lowerquartile, median, and upperquartile, were computed for the central and margin nodes.

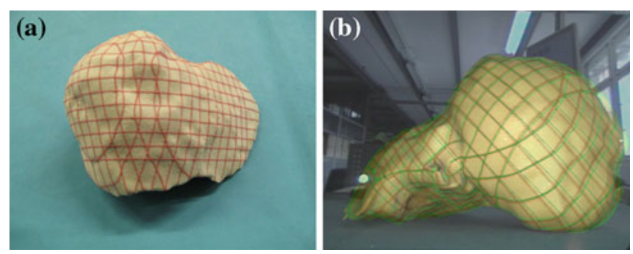

Fig. 2 **a** Rapid prototyped model of a human liver. **b** Image overlay of the AR framework

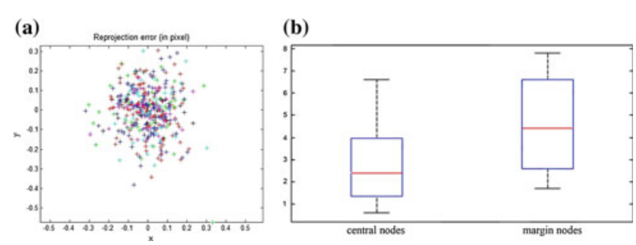

Fig. 3 **a** Reprojection error of the extrinsic parameters related to the best calibration. **b** *Box plot* of the misalignment (in mm) between 3D and endoscopic image

Table 1 Calibration uncertainty

	R(wx, wy,wz) [radians]	T(X,Y,Z) [mm]
EmaX	(0.003,0.003,0.005)	(1.1,1.2,1.1)
Emin	(0.002,0.002,0.004)	(0.7,0.6,0.7)
Emedian	(0.002,0.003,0.004)	(0.9,0.8,0.8)

Conclusion

The accuracy evaluation and the results of the AR framework of our system were presented. We showed that the endoscope image could be overlaid with a 3D image with a mean error of 3.5 ± 1.9 mm. Successful application of image overlay in laparoscopic IGS can potentially lead to better orientation for the surgeon, better identification of structures at risk and to better outcomes. In the future we aim to increase the accuracy of the image overlay and to provide a wide range of AR methodologies and techniques.

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Usage of 3 dimensional preoperative planning and 3 dimensional individualized cutting device for maxillofacial osteotomy

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Keywords Maxillofacial osteotomy · Indivisualized cutting device · Preoperative planning · CAD

Purpose

It is very difficult to plan maxillofacial osteotomy because it is complicated to determine cutting planes which are optimal for superficial part and profound part at the same time. They should also avoid nerves, vessels and so forth. Even 3D CAT scans and MPR viewer were used we still have difficulties to make proper plans. Our purpose is to establish a safer, easier, more accurate, and less time consuming method for preoperative planning and surgical procedures, using 3 dimensional CAOS technology.

Method

For the solution we have been using 3 dimensional preoperative planning and 3 dimensional individualized cutting devices. Three facial hemidisplasia patients were treated by this method. First, CAT scan DICOM data are transferred to STL format CAD data by Mimics® (Materialize, Belgium). Then preoperative planning was done by Magics® (Materialize, Belgium). Mirrored geometries of opposite side were used for the reference. In 1 case, Lefort 1 shortening osteotomy was planned. With this case, the 3 dimensional relationships among the planes were very important. Deciding shortening distance and angle, cutting planes were carefully determined not to destroy palate nor jaw joints, avoiding nerves and vessels. Then the reduction was also simulated and the details such as small overlaps were adjusted. Three dimensional observations were quite useful to check those things. A particular osteotomy with combination of the part of wedges was finally determined. Then expanding the cutting planes, individualized cutting device to be attached on the surface of the bone were machined by a desktop computer numeric machine (Modella MDX-40A, Roland DG, Japan). In another case, mandibular distraction osteogenesis was planned. In this case, osteotomy of outer table of mandible was also required. To avoid mental foramen, the room for the device was too small, dividable device was made and assembled inside the site.

Results

In all cases accurate osteotomies were performed without damaging nerves and vessels with shorter surgical time. The bleeding was small, no blood transfusion was done.

Preoperative planning for hemidisplastic face is very complicated. As the palate has arcade shape, the osteotomy should be curved or bent. It is not parallel to the teeth, so the osteotomy should have slope. If the slope is too much it destroys posterior part of the mandible which leads to huge hemorrhage. Three dimensional CAD technologies could contribute to safer, more accurate and more delicate planning of the osteotomy. Even optimal planning has been prepared, when a surgeon guess the inside mandible from the surface, a small error of the angle can result bigger error inside. Manual readjustment would often be required which would make osteotomy edge rough and less stable. Our individualized cutting device is to be attached on the surface of mandible without any additional measurement. It could provide much more accurate osteotomy with smooth cut surface which we can expect more stability in easier and safer procedures.

Making a plastic model of the whole cranium is also one of the useful way for the surgeon to imagine the osteotomy pre- and intra-operatively. But once the osteotomy was tried on the model, trial of another osteotomy would be impossible. Using 3 dimensional CAD, we can try as many times as we need until we are happy with it. At present the cost of the model is about 2000 Euros. Our device has been made on our desktop and each device costs only about 10 Euros.

Conclusion

The usage of three dimensional preoperative planning and making individualized cutting device could provide safer, more accurate, and more delicate procedures for maxillofacial osteotomy.

Forming the interface between doctor and designing engineer: an efficient software tool to define auxiliary geometries for the design of individualised lower jaw implants

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Keywords Customised software solution · Individualised medical product · Image processing · Auxiliary geometries

Purpose

Manufacturing individualised medical products supported by customised software solutions is becoming more and more important [1, 2]. However, these software tools must be designed so that they are easy to handle and understand. It is not feasible to assume that the surgeon will devote much time to becoming familiar with the