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Abstract. Intraoperative soft tissue deformation, referred to as brain shift, compromises the application of current image-guided surgery navigation systems in neurosurgery. A computational model driven by sparse data has been proposed as a cost-effective method to compensate for cortical surface and volumetric displacements. We present a mock environment developed to acquire stereoisimages from a tracked operating microscope and to reconstruct three-dimensional point clouds from these images. A reconstruction error of 1 mm is estimated by using a phantom with a known geometry and independently measured deformation extent. The microscope is tracked via an attached tracking rigid body that facilitates the recording of the position of the microscope via a commercial optical tracking system as it moves during the procedure. Point clouds, reconstructed under different microscope positions, are registered into the same space to compute the feature displacements. Using our mock craniotomy device, realistic cortical deformations are generated. When comparing our tracked microscope stereo-pair measure of mock vessel displacements to that of the measurement determined by the independent optically tracked stylus marking, the displacement error was ~ 2 mm on average. These results demonstrate the practicality of using tracked stereoscopic microscope as an alternative to laser range scanners to collect sufficient intraoperative information for brain shift correction. © 2017 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: 10.1117/1.JMI.4.3.035002]

Keywords: brain shift; stereoscopic microscope; intraoperative imaging; stereopsis; reconstruction; tracking; accuracy.

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1 Introduction

Image-guided surgery (IGS)¹ provides a standard of care platform for guiding surgeons during brain tumor resection. Unfortunately, commercial IGS navigation systems do not have mechanisms to account for nonrigid tissue deformations, which commonly arise from cerebrospinal fluid drainage, tissue swelling due to edema, tissue contraction due to hyperosmotic drugs, or tissue retraction/resection.² Solutions like intraoperative magnetic resonance (iMR) imaging,³ intraoperative computed tomography (iCT),⁴ and intraoperative ultrasound (iUS)⁵ have been proposed to compensate for soft-tissue changes. However, deficiencies like ionizing radiation of iCT, high cost of iMR, and poor image contrast/quality of iUS have compelled researchers to look for alternatives. Another cost effective method is to use sparse data acquired intraoperatively to drive a biomechanical model to update and register preoperative images to the intraoperative field.⁶ Laser range scanners (LRS) and stereoscopic microscopes are two widely used surface data acquisition techniques.^{7,8} Both devices can be used in an operating room (OR) and generate three-dimensional (3-D) cortical surface cloud data as well as a texture map. Typically, the microscope is continuously used during the whole surgery. Thus, it can provide high-resolution, consistent intraoperative information in near real time with very limited interruption to the

surgical workflow. Performing a single LRS acquisition is more disruptive because it takes about 15 to 30 s and requires the surgical microscope to be moved away from the surgical field of view (FOV).

Standard stereomicroscopes allow surgeons to appreciate the 3-D nature of their domain when performing surgery. This facilitates the physical execution of surgery. With respect to tracked microscopes, once calibrated, these will typically track the focal point of the scope in 3-D space. This allows one to probe the tissue one point at a time to determine the associated imaging data. If one wanted to measure the optical field quickly, such as in measuring the distribution of deformations on the brain surface, the process of refocusing the focal point for each measurement would be highly impractical. The proposed work herein is focused at a tracked stereo-pair reconstruction; put more simply, the system herein is focused at measuring the whole visible brain surface dynamically throughout surgery capturing multidimensional displacements. These measurements may subsequently be used within nonrigid image-to-physical registration methodologies to account for soft-tissue brain shift, which can compromise neuronavigation systems. The proposed tracked stereo-scope solution allows for those measurements potentially to be continually captured throughout surgery without disrupting workflow and with optimal visibility of the tissue of interest. In the recent work, we demonstrated using position-fixed stereo-pair cameras for surface measurements. This work used two

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identical Grasshopper digital cameras produced by Point Grey Research, Inc. (Richmond, British Columbia, Canada) and served as an initial testing prototype for our mock cortical surface environment.⁹

Here, we have extended those approaches to a surgical microscope that is used clinically at our institution. We have developed an interactive environment that permits acquisition of stereo-image capture from the microscope, calibration of the cameras, and adjustment of parameters used for 3-D point cloud reconstruction. We have also equipped the microscope with a rigid body tracking star (Microscope Tracking Array, Brainlab Inc., Westchester, Illinois) to permit tracking of the microscope position within an IGS system. We show that with this device we can register 3-D point clouds that are acquired with the microscope in different positions thus permitting an intraoperative calculation of cortical surface displacement. While the availability of microscope focal point tracking is possible with commercial IGS systems (although accuracy on these commercial systems is not widely reported), the implementation of a fully 3-D tracked approach necessary for measuring full cortical surface field displacements is lacking. The methods we are utilizing for stereovision reconstruction are well-referenced and our goal is to explore this within the scope environment and within the context of brain shift measurement.

2 Methods

Section 2.1 describes the equipment used for data acquisition. Sections 2.2 and 2.3 explain the detail of stereovision techniques and integrated system, respectively. Section 2.4 shows the method of microscope tracking, and Sec. 2.5 discusses the validation experiments.

2.1 Data Acquisition

At the Vanderbilt University Medical Center (VUMC), OPMI Pentero (Carl Zeiss, Inc., Oberkochen, Germany) surgical microscopes are routinely used during neurosurgery procedures [see Fig. 1(a)]. The scope used for this study is equipped with two charged-coupled device cameras, Zeiss MediLive Trio, with a video frame rate of ~30 frames per second. The images or

video stream in the FOV can be displayed on a touch screen monitor, controlled using a joystick, recorded, and exported by video output interface. Through an IEEE 1394 interface, captured data can be saved on a desktop or laptop. In our previous work, a user-friendly graphical user interface (GUI) was proposed to facilitate the acquisition and was designed to be compatible with all USB, Point Grey Research, or other IEEE-1394b (FireWire) digital cameras.^{9,10}

Our craniotomy simulation device is designed to permit in-plane stretching as well as out-of-plane deformation of a membrane to simulate realistic displacements occurring during the surgery [see Fig. 1(c)]. Two circular ring inserts can be placed in the device aperture to simulate different craniotomy sizes, which can vary substantially in different cases. Outside the main circle lie five hemispherical divots that are used for registration with an optically tracked stylus, which permits tracking the deformations in our phantom setup without using the stereo-pair localization. Clamped by the four screws on each side of the device, the membrane inside the circle is made of Dragon skin high-performance silicone rubber and marked with a realistic mock cortical vessel pattern. Lateral shift can be simulated by extending the right side of the device and stretching the silicon membrane through a lead-screw mechanism. In addition to lateral motion, the membrane can be pushed down more than 1 cm by four additional medium-size screws placed around the main circle, which indicates that brain sag often manifests as a withdrawal of the brain surface into the cranium on average of 1 cm although it depends on patient head orientation. Figure 1(d) shows a stair-block phantom that is used for calibration and verification experiment. At the center of each block, there is one red disc with a dent in the circle center that can be digitized by a tracked stylus. Other parts of the phantom are painted nonreflectively in order to improve the performance of disparity computing.

2.2 Stereovision

The first step in stereocalibration is capturing images of calibration objects. Based on our experimental experience, taking 12 to 18 pairs of images usually produce good results. Using less pairs to calibrate does degrade robustness. We have several chessboards with different sized patterns. For this specific case, we use a planar chessboard pattern that has 8×6 internal corners with a block size 13.5 mm as shown in Fig. 2. The chessboard images need to be captured in different positions and orientations that cover the whole FOV. In order to determine both the intrinsic and extrinsic parameters of the stereocameras, a process of calibration is needed to find the relationship between camera and world coordinate system. In our work, we rely on a classical and widely used method¹¹ for camera calibration, which builds the epipolar geometry based on the pinhole model to search for corresponding points in stereo-matching. The fundamental matrix F , which encodes the epipolar geometry of two views, is defined by mapping a point in the left image x_1 to its corresponding epipolar line l_2 in the right image. For the corresponding point x_2 in the right image lying on line l_2 , the following equation is satisfied:

$$x_2^T F x_1 = 0. \quad (1)$$

The image pairs of the checkerboard pattern provide a number of correspondences that are used to solve F using a least squares method.¹¹ Then, the camera matrix of each camera can be

Fig. 1 Surgical operating microscope with (a) optical marker attached, (b) optical tracking device, (c) simulated craniotomy device, and (d) calibration phantom.

Fig. 2 Three pairs of stereomages placed at different positions with corner detected.

retrieved using the fundamental matrix F . The accuracy of calibration is evaluated by computing the distance between the corresponding corners and the estimated epipolar lines, since in theory the corresponding corner lies on the epipolar line.

The ultimate aim of the reconstruction process is to reproject two-dimensional (2-D) image points to physical world positions. The perspective reprojective matrix Q is defined

$$Q = \begin{bmatrix} 1 & 0 & 0 & -c_{x1} \\ 0 & 1 & 0 & -c_{y1} \\ 0 & 0 & 0 & f \\ 0 & 0 & -\frac{1}{T} & \frac{c_{x1}-c_{x2}}{T} \end{bmatrix}, \quad (2)$$

where point (c_{x1}, c_{y1}) is the principal point in left image, f is the focal length (when we reproject the left image, f is the focal length of the left camera, and vice versa, for the other), and T is the horizontal distance between the two cameras centers C_1 and C_2 . c_{x2} is the x coordinate of the principal point in the right image. The bottom right corner $(c_{x1} - c_{x2})/T$ is equal to zero if two principal rays intersect at infinity in a frontal parallel configuration. A 2-D homogeneous point (associated with disparity d) can be transformed into a 3-D point $(X/W, Y/W, Z/W)$ in world space by the following mapping:

$$Q \begin{bmatrix} x \\ y \\ d \\ 1 \end{bmatrix} = \begin{bmatrix} X \\ Y \\ Z \\ W \end{bmatrix}, \quad (3)$$

where W is the extra dimension in projective geometry. The image acquisition interface guarantees simultaneous capture

of stereopair, however, the left and right images are not perfectly parallel-aligned. Image pairs thus need to be rectified before computing disparity. Hartley¹² has proposed an algorithm that finds homographies that map the epipoles to infinity using just the fundamental matrix F . The main drawback of this method is the ambiguity in image scale, which can lead to inaccuracy in the size of the reconstructed 3-D objects. Another algorithm has been proposed by Bouquet,¹³ which takes advantage of two calibrated cameras' rotation and translation parameters to minimize reprojection distortions as well as maximizing the view overlap area. In this work, we have used Bouquet's approach. After the rectification process is completed, the disparity map can be computed by finding the difference in x coordinates of the same features in the rectified left and right images. A number of algorithms have been proposed to calculate the disparity map, e.g., block matching (BM),¹⁴ semi-global block matching (SGBM),¹⁵ and a variety of other algorithms that have been evaluated and compared on an online platform.^{16,17} Considering simplicity, potential for near real-time performance, and availability of open-source libraries such as Open Computer Vision Library (OpenCV),¹⁸ we choose the BM approach for our stereo-reconstruction algorithm. Most stereo-matching algorithms are challenged by uniform textureless areas and regions with depth discontinuities. One possible solution is to use postfiltering¹⁹ to align the disparity map edges with the source image. This has been adopted here within OpenCV. Figure 3 compares the disparity map result; the upper row shows the results with stereoBM alone, whereas the lower row shows the results with stereoBM and postfiltering.

2.3 Integrated System

Point cloud reconstruction from microscopic stereo-pair images requires several steps: camera calibration, image rectification, disparity computation, and 3-D point cloud reconstruction. Each of these steps may require parameter adjustments to produce acceptable results. To facilitate the process and make it achievable by users that are not experts in computer vision, we have developed an interactive environment. This environment includes a GUI through which parameters can be adjusted using check-boxes, spin-boxes, and line-edit widgets. Intermediate and final results are also shown. This GUI is written in C++ using Qt²⁰ and can be run on Windows and Linux platforms. We used the OpenCV library¹⁸ for the computer vision algorithms, and the Point Cloud Library²¹ was used to display and process point cloud data.

Figure 4 shows the main window of the software. On the left side of the interface is a tree-view structure used to select the input images, the output point clouds as well as some intermediate results. Moving to the right, a display area facilitates visualization of the stereo-pair images (left and right), and the panel below contains the series of actions and options necessary to perform the point cloud reconstruction. The first step involves stereocapture of a calibration checkerboard pattern, and the user is guided through the process. The next step is the localization of the corners in the checkerboard images, and the calibration of the stereocameras using the method described in Ref. 11.

Once the calibration is complete, the output camera matrices are used to rectify left and right images. The disparity map can then be computed using either a BM or an SGBM algorithm.¹⁶ Both BM and SGBM algorithms use nine parameters that can be adjusted. These can be adjusted using sliders while showing the disparity image produced by the current parameter values

Fig. 3 Compare the disparity map with and without postfiltering. (a) StereoBM and (b) StereoBM with postfiltering.

Fig. 4 Integrated reconstruction software: main window.

[see Fig. 5(a)]. Note that all parameters for computing disparity maps begin with default values, which in many instances work quite well. When adjustments are needed, the user interface we have developed to achieve rapid tuning, Fig. 5(a), has been enormously helpful. Once the disparity image computation is deemed acceptable by the user, the point cloud can be computed for each disparity image [see Fig. 5(b)]. After computing the reconstruction,²² the display panel is invoked to immediately provide a visualization of the point clouds.

2.4 Microscope Tracking

Microscope tracking can be performed via the use of a rigidly attached optically tracked reference body, a calibration procedure to compute the transformation between the coordinate system specified by the rigid tracking body, and the coordinate system of the reconstructed stereo-pair point cloud.²³ We use a Polaris Spectra optical tracking system [see Fig. 1(b)] developed by Northern Digital, Inc. (Waterloo, Ontario, Canada), which has a reported tracking accuracy of 0.25- to 0.3-mm root

Fig. 5 Integrated reconstruction software: (a) disparity tuner panel and (b) point cloud display window.

mean square.²⁴ A commercially available rigid body reference with microscope mounting attachment (Microscope Tracking Array, Brainlab Inc., Westchester, Illinois) was employed to facilitate tracking of the surgical microscope.

The setup for microscope tracking calibration is sketched in Fig. 6(a). To summarize, the goal of the microscope tracking calibration is to determine the rigid body transform that provides a mapping between the coordinate system of the reconstructed stereo-pair point cloud (X_{cam}) and the coordinate system of the rigid body attached to the surgical microscope (X_{star}). This calibration transform ($T_{cam-star}$) is computed using a calibration phantom that is comprised of a series of fiducial disks that can

Fig. 6 The setup of (a) calibration procedure and (b) tracking experiment.

be localized in both the reconstructed stereo-pair point cloud space and the coordinate system space of the microscope rigid body. The fiducial points are located in the reconstructed stereo-pair space via computation of disk centroids from the reconstructed point cloud. The fiducial point locations in the microscope rigid body space are determined via an optically tracked probe that can digitize the individual fiducial points relative to the microscope rigid body within the optical tracking system. Once these individual fiducial points have been localized in each space, a point-based registration²⁵ is computed to determine the camera calibration transform ($T_{cam-star}$).

2.5 Validation Experiments

The validation experiment includes two parts: microscope tracking validation and vessel displacement validation. The aim of the microscope tracking validation is to ensure that the calibration transformation ($T_{cam-star}$) is accurate via the transformation of a series of point cloud reconstructions, acquired at different microscope locations, to a reference coordinate space [see Fig. 6(b)]. At an initial position of the surgical microscope ($X_{cam,1}$), the tracked location of the scanner was recorded ($T_{star-opt,1}$), and a stereo-pair reconstruction was performed of the calibration phantom. With the phantom and optical tracking system in a fixed position, the surgical microscope was then moved to a second position ($X_{cam,2}$). Again, the tracked location

of the scanner was recorded ($T_{\text{star-opt},2}$), and a stereo-pair reconstruction was performed. Given the recorded data from the two locations, the fixed microscope calibration transformation ($T_{\text{cam-star}}$) and the fixed optical tracking coordinate system (X_{opt}), the stereo-pair reconstructed point clouds acquired at the two microscope locations can be transformed into the same space using the following equations:

$$X_{\text{opt}} = [T_{\text{star} \rightarrow \text{opt},1}][T_{\text{cam} \rightarrow \text{star}}]X_{\text{cam},1}, \quad (4)$$

$$X_{\text{opt}} = [T_{\text{star} \rightarrow \text{opt},2}][T_{\text{cam} \rightarrow \text{star}}]X_{\text{cam},2}. \quad (5)$$

The steps are as follows:

1. Place the scope at base position, then place the calibration phantom under the scope. Reconstruct the point cloud of the phantom.
2. Digitize the centroids of the nine disks and record the position of the scope using the optical tracking system. The transformation matrix can be computed.
3. Move the scope to another position while keeping the phantom static. Reconstruct the phantom and record the scope position.
4. Repeat step 3.
5. Use the transformation matrix from step 2 to transform the point cloud in step 3 and 4 back to base position and then evaluate the error.

Fig. 7 Three states of generating simulated brain shift: (a) baseline, (b) stretch (lateral shift), and (c) stretch as well as sag.

For the vessel displacement computation, we use the craniotomy device (described in Sec. 2.1) to simulate brain shift via three states (baseline, stretch, and sag as shown in Fig. 7). The experimental steps are as follows:

1. Set the craniotomy device in the baseline state and place it under the scope.
2. Reconstruct 3-D point cloud of the device and record the ground truth positions of vessel features marked on the membrane using optical tracking system.
3. Move the scope to a new position and apply a 2-cm horizontal stretch (lateral shift) using the screw mechanism. Then, repeat step 2.
4. Move the scope to another new position, and use the four screws around the craniotomy to displace the membrane downward by 1.6 cm (this simulates sag). Then, repeat step 2.
5. Use the previous transformation matrix to transform point cloud in step 3 and step 4 to base space. The movement of features marked on the membrane can be computed after the application of the tracking transformations.

To validate our system, the deformations are recorded using the optically tracked stylus. This serves as the ground truth that

Fig. 8 (a)–(c) Stereo-pair derived point clouds from three different microscope positions, respectively, (d) the point cloud from (b) aligned to the coordinate reference of (a), (e) the point cloud from (c) aligned to the coordinate reference of (a), and (f) the overlay of all clouds (a)–(c) to the reference of (a).

can be subsequently compared to the tracked microscope stereo-pair measurements.

3 Results

Result for the stereo-pair camera calibration experiment is shown in the GUI (see Fig. 4). The average epiline error is 0.2 pixels, and the stereo-projection error is reported as 0.41 pixels. Knowing the geometry of the calibration phantom, a reconstruction error of ~ 1 mm is estimated by systematically comparing the distance between nine divots. Since the microscope is tracked, the transformation between point clouds generated via scope acquisitions at different positions can be computed. We reconstructed the point cloud of the calibration phantom by moving the scope to three positions [see Figs. 8(a)–8(c)].

By applying the transformation matrix to the second and third point clouds (highlighted in yellow and blue, respectively, in Fig. 8), these two point clouds can be transformed into the same space as the initial point cloud [see Figs. 8(d) and 8(e)]. Figure 8(f) shows the results of registering all three

point clouds together. The mean distance error is 0.64 mm in the x -direction, 0.89 mm in the y -direction, and 2.92 mm in the z -direction. These errors are caused by reconstruction error and tracking error. The displacement is computed by registering the simulated vessel features on the simulated craniotomy device to the same space (baseline state) shown in Fig. 9. Note that feature No. 1 is not included in the results because it is no longer visible following the stretch (lateral shift) operation of the craniotomy device, which can also happen in actual surgeries when part of the brain slides past the craniotomy. As each measurement is independently taken and as our goal is to measure the widest distribution of shift on the surface, it will not impact the results much besides one less measurement. The difference between displacement in stereopsis (measured in 3-D point cloud) and tracking system (ground truth) is calculated as displacement error and is found to be ~ 2 mm on average (see Fig. 10).

4 Discussions

The phantom experiments performed for the purpose of validating the tracking calibration for the surgical microscope use three different microscope positions. In fact, we can continuously acquire stereo-pair data and render 3-D point clouds in a

Fig. 9 Bar of (a) displacement and (b) displacement error.

Fig. 10 Box plot of (a) displacement and (b) displacement error.

consistent space. In our experience within the OR, the scopes are positioned quite similarly in space with only small changes from a practical use standpoint. However, for the work herein, while we chose only three microscope positions, we did purposefully position broadly across the mock-OR space for testing (as compared to its *in vivo* counterpart use, which is typically more limited). The experiments yield a number of error metrics that provide some insight into the range of possible error sources that are contributing factors. These contributing factors include the error due to the stereo-pair reconstruction process, manual error associated with the fiducial digitization and centroid extraction, suboptimal calibration phantom design, and the tracking error associated with the rigid body attachment to the microscope and the tracked stylus.

The stereo-pair reconstruction error highly depends on the quality of the disparity computation. Camera calibration affects the rectification step, which is critical for computing disparity. The scope tracking error is computed by comparing the point clouds that are transformed to the same scope coordinates. The X-, Y-, and Z-direction mean distance errors are computed, and the results show that the error in the Z-direction is higher than in the other two directions. This may be due to the error in estimating depth from disparity. The point cloud reconstruction is more accurate in the horizontal direction than in the vertical direction. Considering that the maximum height difference of the steps on the phantom is nearly 50 mm (larger than the normally reported brain deformation), the mean distance error is, to some degree, acceptable. We also note that stereo-reconstruction techniques are challenged by sharp edges such as the steps on the phantom. We expect that our reconstruction error would be smaller with a phantom that would have smoother edges as would the cortical surface.

Moreover, the error of the tracking system itself and the localization error associated with the use of a tracking stylus also contribute to the calibration error of the surgical microscope. Error associated with the tracking system used in these experiments is reported by the manufacturer and is within an acceptable range. However, any error associated with the calibration files generated for the tracking of the rigid body attached to the microscope as well as the tracked probe can be potential sources of error. Finally, any manual localization inaccuracy in the digitization of the points on the calibration phantom with the optically tracked probe can introduce errors. With respect to conventional image guided neurosurgery, accuracy of 1 to 2 mm is certainly a target goal to strive for (note that this is the typical range of practically tracked tools). While 2 to 3 mm is not ideal, we would suggest that the accuracy is clinically relevant as brain shift that routinely occurs in tumor resection surgery is on the order of 1 to 2 cm. Therefore, the accuracy of this system is sufficient to provide some benefit to soft-tissue deformation correction IGS systems. We do, however, acknowledge that more work is needed to improve the fidelity, but the work here is highly encouraging.

The main goal of this work is to demonstrate the ability to take advantage of stereovision-based techniques to track tissue deformations with arbitrary movement of the surgical microscope. While the results presented indicate that the presented method of microscope tracking calibration is promising, there are a number of avenues for improving the calibration results. Primarily, refinement in reconstruction techniques should facilitate more accurate calibrations. Additionally, the use of an optimally designed calibration phantom would facilitate more

accurate results as well. Finally, our calibration method is only useful for fixed microscope setting where the focal length and zoom factor cannot change during the whole procedure. Typically, when magnifications are changed, surgeons are navigating deeper into the brain tissue. In our experience, tissue features for stereo-pair reconstruction are not readily present as the tissue is invaded. In fact, when navigating deep into the tissue, typically the common focal point localization associated with commercial tracked scopes is all that is needed, i.e., make the microscope essentially similar to a tracked stylus. With that said, we note clearly here that this routine function does not diminish the work presented herein. More specifically, with a single tracked focal point, it would be impractical to refocus the scope on each tissue feature to continually track deformations. With that understanding and in considering measuring deformations of the feature-rich cortical surface, in our experience, this can be achieved by coming back to the original FOV with the near-original magnification to make measurements of the visible cortical surface at the start and at different times during surgery. However, the ability to recalibrate at any time, including when the OR field is sterile, is an important feature to a generalizable framework. We are working toward a solution that combines tracked stylus localization and collocated stereo-pair point reconstruction object that essentially allows for very fast calibration. Future methods will entail the ability to generate calibration data that allows the use of multiple microscope settings throughout the surgical procedure.

5 Conclusions

In this paper, we have proposed a GUI-based system that integrates all necessary functionality for reconstructing FOV of stereopair or microscope cameras including (1) capturing stereo-pair images or video streams, (2) extracting checkerboard corners, (3) calibrating stereo-cameras, (4) computing disparity, and (5) displaying point clouds. Moreover, the parameters associated with the disparity computation can be modified in an interactive GUI to improve the results. This reconstruction system is functional, user-friendly, and requires only minimal prior knowledge. By applying image-to-physical space registration, the stereoscopic microscope can be tracked and freely moved without disrupting the surgical procedure. The reconstruction accuracy and displacement comparison results suggest that this system could be used to gather cortical data to measure and compensate for brain shift during IGS. This would extend the capability of conventional navigation system.

Disclosures

No conflicts of interest including financial or others are declared by the authors.

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References

1. R. L. Galloway, Jr., "The process and development of image-guided procedures," *Annu. Rev. Biomed. Eng.* **3**(1), 83–108 (2001).
 2. A. Nabavi et al., "Serial intraoperative magnetic resonance imaging of brain shift," *Neurosurgery* **48**(4), 787–798 (2001).
 3. C. Nimsy et al., "Quantification of, visualization of, and compensation for brain shift using intraoperative magnetic resonance imaging," *Neurosurgery* **47**(5), 1070–1080 (2000).
 4. T. Gunnarsson et al., "Mobile computerized tomography scanning in the neurosurgery intensive care unit: increase in patient safety and reduction of staff workload," *J. Neurosurg.* **93**(3), 432–436 (2000).
 5. G. Unsgaard et al., "Intra-operative 3D ultrasound in neurosurgery," *Acta Neurochir.* **148**(3), 235–253 (2006).
 6. M. I. Miga et al., "Modeling of retraction and resection for intraoperative updating of images," *Neurosurgery* **49**(1), 75–85 (2001).
 7. M. I. Miga et al., "Cortical surface registration for image-guided neurosurgery using laser-range scanning," *IEEE Trans. Med. Imaging* **22**(8), 973–985 (2003).
 8. A. N. Kumar et al., "Persistent and automatic intraoperative 3D digitization of surfaces under dynamic magnifications of an operating microscope," *Med. Image Anal.* **19**(1), 30–45 (2015).
 9. X. Yang et al., "A novel craniotomy simulation system for evaluation of stereo-pair reconstruction fidelity and tracking," *Proc. SPIE* **9786**, 978612 (2016).
 10. X. Yang, "Stereo-pair capture software," Vanderbilt.edu, <https://my.vanderbilt.edu/xiaochen/archives/75> (August 2015).
 11. Z. Zhang, "A flexible new technique for camera calibration," *IEEE Trans. Pattern Anal. Mach. Intell.* **22**(11), 1330–1334 (2000).
 12. R. I. Hartley, "Theory and practice of projective rectification," *Int. J. Comput. Vision* **35**(2), 115–127 (1999).
 13. J.-Y. Bouguet, "Visual methods for three-dimensional modeling," PhD thesis, Citeseer (1999).
 14. K. Konolige, "Small vision systems: hardware and implementation," in *Robotics Research*, pp. 203–212, Springer, New York (1998).
 15. H. Hirschmuller, "Stereo processing by semiglobal matching and mutual information," *IEEE Trans. Pattern Anal. Mach. Intell.* **30**(2), 328–341 (2008).
 16. D. Scharstein and R. Szeliski, "A taxonomy and evaluation of dense two-frame stereo correspondence algorithms," *Int. J. Comput. Vision* **47**(1–3), 7–42 (2002).
 17. D. Scharstein and R. Szeliski, "Middlebury stereo evaluation—version 3," <http://vision.middlebury.edu/stereo/eval3/> (April 2017).
 18. G. Bradski and A. Kaehler, *Learning OpenCV: Computer vision with the OpenCV library*, O'Reilly Media, Inc., Sebastopol, California (2008).
 19. OpenCV, "Disparity map post-filtering," http://docs.opencv.org/trunk/d3/d14/tutorial_ximgproc_disparity_filtering.html (April 2017).
 20. "Qt powerful, interactive and cross-platform applications," <https://www.qt.io/> (July 2016).
 21. "Point cloud library (PCL)," <http://www.pointclouds.org/> (July 2016).
 22. R. Hartley and A. Zisserman, *Multiple View Geometry in Computer Vision*, Cambridge University Press, Cambridge, United Kingdom (2003).
 23. D. M. Cash et al., "Incorporation of a laser range scanner into image-guided liver surgery: Surface acquisition, registration, and tracking," *Med. Phys.* **30**(7), 1671–1682 (2003).
 24. P. Spectra, "Polaris optical tracking system," <http://www.ndigital.com/medical/products/polaris-family/> (July 2016).
 25. M. Fitzpatrick, "Algorithms for registration, error prediction, and rotation parameter conversion in Matlab," http://eecs.vanderbilt.edu/people/mikefitzpatrick/computer%20programs/computer_algorithms_in_Matlab.htm (July 2016).
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