2-D and 3-D Visualization Methods of Endoscopic Panoramic Bladder Images

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2–D and 3–D Visualization Methods of Endoscopic Panoramic Bladder Images

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ABSTRACT

While several mosaicking algorithms have been developed to compose endoscopic images of the internal urinary bladder wall into panoramic images, the quantitative evaluation of these output images in terms of geometrical distortions have often not been discussed. However, the visualization of the distortion level is highly desired for an objective image-based medical diagnosis. Thus, we present in this paper a method to create quality maps from the characteristics of transformation parameters, which were applied to the endoscopic images during the registration process of the mosaicking algorithm. For a global first view impression, the quality maps are laid over the panoramic image and highlight image regions in pseudo–colors according to their local distortions. This illustration supports then surgeons to identify geometrically distorted structures easily in the panoramic image, which allow more objective medical interpretations of tumor tissue in shape and size. Aside from introducing quality maps in 2–D, we also discuss a visualization method to map panoramic images onto a 3–D spherical bladder model. Reference points are manually selected by the surgeon in the panoramic image and the 3–D model. Then the panoramic image is mapped by the Hammer–Aitoff equal–area projection onto the 3–D surface using texture mapping. Finally the textured bladder model can be freely moved in a virtual environment for inspection. Using a two–hemisphere bladder representation, references between panoramic image regions and their corresponding space coordinates within the bladder model are reconstructed. This additional spatial 3–D information thus assists the surgeon in navigation, documentation, as well as surgical planning.

Keywords: Bladder, Fluorescence Endoscopy, Image Mosaicking, Quality Maps, Visualization

1. INTRODUCTION

Urinary bladder cancer is a common cancer disease with over 70,000 new cases estimated in 2010 in the US.1 Its medical diagnosis and therapy are usually carried out by a cystoscopy. During this intervention an endoscope is guided through the urethra to examine the internal bladder wall. To provide a high contrast between malignant and benign tissue photodynamic diagnostics (PDD) based on fluorescence excitation is used. Using a 5–ALA marker and a bluish narrow band illumination, malignant tissue fluoresces reddish. This results then in a visually improved tissue contrast compared to white light endoscopy.2 Cystoscopic images usually cover only a very limited field of view (FOV), which impede navigation in the bladder. Thus, different mosaicking algorithms for white light3 as well as PDD modality2 have been developed to compose overview images. These panoramic views provide a larger FOV, assist in navigation and surgical planning, and complement medical evidence protocols.

Although the performance, registration errors, and blending strategies of many mosaicking algorithms have been evaluated,2,4–6 the spatial distributions of geometric image distortions have not been discussed. In general any map projection introduces distortions into the panoramic view,7,8 if images taken from a 3–D sphere–shaped bladder surface are mapped onto a 2–D reference plane. Although distortions of standard map

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projections are analytically computable, the free-hand movement of the endoscope leads to superpositions and local variations of angle, distance, and area distortions across the whole panorama image. Thus, explicit and quantitative measurements of the level of the local distortions in the panoramic images are required for an objective medical diagnosis.

Another difficulty to interpret image mosaics is the loss of absolute positions. Using standard monocular video endoscope systems without any camera calibration and tracking devices, the relation between the positions of the endoscope in 3-D and the image coordinates is unknown. Although the mosaicking algorithm transforms all single images into one global reference frame, only relative motion information is provided. Thus, a visualization method which puts panoramic views at their corresponding positions on the surface of an anatomical 3-D bladder model is desired. Using an adequate map projection, the relation between diagnostically relevant image regions, like tumors and their 3-D positions at the bladder wall can be reconstructed by texture mapping. Furthermore it provides surgeons with an interactive 3-D overview of all medical findings in the panoramic view, and assists in the analysis of tumor distributions, which indicate the likelihood of carcinoma in situ.

Addressing both problems, we present in this paper a visualization method, which analyzes and highlights local image distortions in panoramic images using pseudo-colored quality maps. Furthermore we discuss a semi-automatic 3-D visualization algorithm to reconstruct the correlation between image and space coordinates based on an equal-area map projection and a spherical 3-D bladder model.

2. IMAGE MOSAICKING

The composition of single endoscopic images into one panoramic image is performed by our real-time capable PDD mosaicking algorithm. During this process the endoscopic images are first rectified from lens distortions based on camera calibration data. Then, distinctive SURF (Speeded Up Robust Features) feature points are extracted on multi-scales. Minimizing the Euclidean distance between feature descriptors, point correspondences between subsequent images are determined. Based on these matches a homography matrix $H$ with six degrees of freedom is estimated iteratively. It describes the affine transformation $\vec{p}_i = H \cdot \vec{p}_j$ of all matching points $\vec{p}_i \leftrightarrow \vec{p}_j$ with the lowest overall reprojection error. Written in homogeneous coordinates $H$ becomes

$$H = \begin{bmatrix} c & d & e \\ f & g & h \\ 0 & 0 & 1 \end{bmatrix} = \begin{bmatrix} A & \vec{t} \\ 0^{T} & 1 \end{bmatrix},$$

with a $2 \times 2$ non-singular matrix $A$ and a translation vector $\vec{t} = (t_x, t_y)^T$. Although strong perspective distortions can only be compensated by a projective transformation, eq. 1 sufficiently describes the global image transformation between two subsequent images, since the endoscope is guided closely ($\leq 1$cm) along the bladder wall under PDD illumination. To compose one single mosaic, all input images are transformed into the same reference frame, determined e.g. by the first image $I_1$ of the sequence. The global transformation $H^2_{n,1}$ of image $I_n$ is then given by

$$H^2_{n,1} = H_{2,1} \cdot \ldots \cdot H_{n-1,n-2} \cdot H_{n,n-1} = \prod_{i=1}^{n-1} H_{i+1,i}.$$

After registration, overlap regions are linear interpolated to reduce seam, blur and ghosting artifacts. Finally, the successively growing panoramic views are displayed on a clinical monitor.

3. 2-D QUALITY MAPS

Since no additional information about the camera position and the bladder surface is available, the images are mapped by the mosaicking algorithm onto a planar projection plane. Any map projection of a 3-D surface onto a 2-D reference plane introduces distinct distortions. Approximating the bladder by a sphere,
folding of the surface into single stripes results in a "peeling" problem, as illustrated in Fig. 1. Since the area of each stripe is preserved, adjacent latitudes are progressively bended since the upper and lower edges differ in length. Although misalignment of single mosaics caused by drift and registration errors can be compensated in the panoramic image by drift compensation and bundle adjustment, these distortions are unavoidable and remain persistently. To overcome this problem, the bladder surface can be modeled by single hemicube planes, leading to separate local panoramas with lower but still some geometric distortions. This approach provides panoramic views of the surgical field, which can already assist the surgeon in orientation and reidentifying tumors. However, the medical diagnosis based on the visual inspection of the panoramic views is limited, since the image compositions do not provide unbiased data in all image regions. Thus, quality maps which identify, quantify, and assess local image distortions are highly desired for an objective visual interpretation.

To analyze and track the characteristics of single transformation parameters across the whole panoramic image, we decompose each matrix $H^q_{n,1}$ of eq. 1 and eq. 2 by the QR decomposition

$$H^q_{n,1} = \begin{pmatrix} \cos(\alpha) & -\sin(\alpha) & 0 \\ \sin(\alpha) & \cos(\alpha) & 0 \\ 0 & 0 & 1 \end{pmatrix} \ \begin{pmatrix} 1 & t_x \\ 0 & 1 & t_y \\ 0 & 0 & 1 \end{pmatrix} \ \begin{pmatrix} s_x & a & 0 \\ 0 & s_y & 0 \\ 0 & 0 & 1 \end{pmatrix}$$

$H^q_{n,1}$ becomes the product of a rotation, shift, and scale & shear matrix with six degrees of freedom in total. Since physicians are primarily interested in a correct and consistent appearance of structures like tumors and lesions, in size and shape over the whole panoramic image, the area magnification $m = |H^q_{n,1}| = s_x s_y$ and the shear distortion parameter $a$ are selected for evaluation. Traditionally, the characteristics of these parameters can be easily plotted over the image numbers, as shown in Fig. 2.

Figure 2. Characteristics of the area magnification $m = s_x s_y$ (left) and the shear parameter $a$ (center) over the image numbers of the panorama of the genuine bladder sequence Cystoscopy 1. PDD panorama with its trajectory (right).
Figure 3. Panoramic images with their trajectories (top row) composed from a bladder phantom (left column), and two genuine cystoscopy sequences (center and right column). Pseudo-color quality maps highlight the area magnification $m = s_x s_y$ (center row) and the values of shear parameter $a$ (bottom row) according to their colorbars.
Although this representation illustrates the whole parameter characteristics of the image composition, spatial correlations are missing. Thus, the identification of abnormal image regions, which e.g. show high area magnifications or strong distortions, becomes very difficult. To overcome this problem, we developed color–coded 2–D quality maps to visualize the parameter characteristics locally across the panoramic image. Using a pseudo–color illustration, the values of the transformation parameters \( m \) and \( a \) of each image are displayed over the intensity channel of the panoramic image. Examples composed from a phantom and genuine cystoscopy sequences are given in Fig. 3.

A variation to the reference values of \( m = s_x s_y = 1 \) and \( a = 0 \) (colored in green) is highlighted by a bluish color for lower and a reddish color for higher values, respectively (cf. colorbars in Fig. 3). In contrast to the plots in Fig. 2 the characteristics of the parameters can now be easily assigned to local regions of the panoramic image. In the example, the area magnification about a value of \( m = 1.6 \) at the image numbers 12–17 (c.f. Fig. 2) is located in the upper right corner of the panorama image of Cystoscopy 1, as shown in Fig. 3. Also the increase of the shear parameter \( a \) at the end of the video sequence becomes iteratively apparent due to the pseudo–colors. Besides the here chosen parameter \( m \) and \( a \), other values like e.g. the scale ratio \( r = \frac{s_x}{s_y} \) can be easily processed and evaluated in the same way. Since single spatially adjacent images in the panorama are interpolated within the transition zone by a linear cross–blending method,

Thus, the color–coded illustration assists the physician to interpret the geometrical distortions of the panorama image at a single glance. Also the switching between the panoramic image and the semi–transparent quality maps indicates directly the location of distorted image structures, which can now be considered for an objective diagnosis. Furthermore the parameter values at each pixel region can be displayed for a quantitative analysis using a data cursor.

4. 3–D VISUALIZATION

Addressing the issue of unknown space coordinates, we describe in the following a method to map image mosaics directly onto the surface of a 3–D bladder model. Motivated by common medical urinary depictions as shown in Fig. 4, two aligned hemispheres are chosen for visualization. To determine the relation between

\[
\begin{align*}
x &= \frac{2\sqrt{2}\cos\theta\sin\frac{\phi}{2}}{z}, \\
y &= \frac{\sqrt{2}\sin\theta}{z}, \\
z &= \sqrt{1 + \cos\theta\cos\frac{\phi}{2}},
\end{align*}
\]

for mesh parameterization. The 3–D surface coordinates \((\theta, \phi)\) of the sphere are directly mapped onto the plane \((x, y)\). The sphere is transformed in an ellipse with axes in a 2 : 1 ratio. The distribution of the

![Figure 4. Urological depiction with labeled bladder regions.](image)
triangles is more homogeneous and the areas are consistent, compared to the parameterization generated by the spring-mass model with cotangent weights, as shown in Fig. 6. Although the Hammer–Aitoff projection is only limited to a spherical object, it provides a fast and computationally inexpensive transformation, which does not introduce additional area distortions.

The initial alignment between the panoramic image and the mesh parameterization is done by manually selecting three points in the image and their correspondences in the 3-D model using ray tracing. Based on these matches, an affine transformation matrix $A_T$ (cf. eq. 3) is calculated, and the panorama image is transformed into the parameterization framework, as illustrated in Fig. 7. Using barycentric coordinates and the inverse projection of eq. 4, each face of the 3-D surface is then textured. Thus, the whole panoramic image of the bladder scan is displayed onto the hemispheres. At this point, the location of tumors and lesions
displayed in the panoramic image are correlated to 3-D positions of the bladder model. An example is given in Fig. 8.

Although the applied transformations do not introduce further area distortions in the panoramic image, geometrical distortions due to perspective views will still remain in the texture information. Furthermore error caused by the "peeling" effect (cf. Fig. 1) can only be fully compensated by exactly known and reconstructed camera positions. Thus, the quality maps of the panoramic images (cf. Fig. 3) could also be transformed into the mesh parameterization framework and displayed onto the 3-D model.

Consequently, this visualization method provides a more intuitive illustration of the internal bladder surface than 2-D planar images alone. The physician can now virtually inspect the whole bladder by controlling the camera view interactively and analyze the spatial distribution of findings across the bladder in 3-D for documentation and surgery planning.

5. CONCLUSIONS

Based on the evaluation of single transformation parameters, which were determined by image mosaicking algorithms during the stitching process, we designed color-coded quality maps to describe local variations of area magnifications and shear values. After the maps are overlaid with the original panoramic images, geometrical distorted image regions can be easily identified, supporting a more objective medical interpretation of the provided image data. For an adequate visualization method of bladder images in 3-D we propose a basic spherical 3-D bladder model and an equal-area mapping using Hammer–Aitoff projection. Setting reference points by the surgeon, the positions of image structures in 3-D are reconstructed. This additional information supports then the analysis of tumor distributions across the bladder wall, as well as it assists in navigation and surgical planning. Although preliminary feedback of surgeons showed already high potential of these visualization methods, further objective clinical evaluations will be carried out in future work.

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