A NEW SYSTEM FOR FREEHAND ULTRASOUND TRACKING APPLICATION IN MULTIMODALITY IMAGE MATCHING

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Abstract

This paper describes a localization system composed of a stereovision system with two cameras and an ultrasound system. After calibration of 2 cameras, a mask of infra-red LEDs is mounted on the probe. A method for spatial and temporal calibration is presented and used to calibrate the ultrasound system. Position and orientation of each ultrasonic cross-section are precisely measured and a 3D localization is available from the US images with an accuracy of 0.4 mm. We present two clinical applications of the system which gave good results in each case.

Key Words

Ultrasound imaging, stereovision, calibration, Measurement and instrumentation.

1 Introduction

Ultrasonic imaging has become an important modality in the field of medical imaging systems, due to its flexibility and its non-invasive principle. It is used for the diagnosis and treatment of cancer of many organs. Ultrasound is also the method of reference for intra-operative visualization [1], [2].

Currently, most of the ultrasound systems provide a 2D image, the orientation of which depending on the orientation of the probe. But no spatial information are available. To establish his diagnosis, the expert must then mentally rebuild the characteristics of volume by coordinating the images with the position of the probe. The interest of 3D ultrasound is to carry out this integration of the images and the positions in an automatic way. Performing this kind of examinations requires an acquisition followed by the reconstruction and finally visualization. Three types of solutions exist: 3D probes, mechanical systems and freehand systems. The phase of acquisition requires an accurate image plan localization generated each time in a 3D reference frame. Systems of localization can be mechanical, acoustic, optical or magnetic [3].

However, a freehand system of 3D localization of US slices can be exploited in other cases. Indeed, such a system is able to help the expert for the control and checking of patient position during conformal radiotherapy sessions, following tools or radioactive implants in brachytherapy and finally for the matching with other imaging modalities as CT or MRI.

For this kind of applications the systems described in literature show their limits due to their complexity or by the non compatibility with the material used in treatment room.

In this paper, we present an accurate localisation system, easy to setup and without interference with other installations. A method for spatial and temporal calibration of the ultrasound system is presented.

2 Materials and Methods

The aim is to locate each pixel on the US slice in a 3D space $R_{com}$ (figure 1):

Fig. 1 Tracking transformation

$T_{track}$ which gives the 3D ($x$, $y$, $z$) position of pixel ($u$, $v$) on the US image is composed by :

$T_{Track}=T_{Probe} \ast T_{US}$

where :

$T_{US}$ is the rigid transformation from $R_{us}$ to $R_{probe}$ and

$T_{Probe}$ rigid transformation from $R_{probe}$ to $R_{com}$. 
2-1 Camera calibration

We have installed a stereovision system including 2 monochromic CCD video cameras (512 X 512 pixels), sensitive to infra-red radiation. The cameras are calibrated [4], [5] using a calibration frame with a plan composed of 56 infra-red diodes (LEDs). The plan can be installed in 4 positions what gives 224 reference points. We used these points to solve, by the singular value decomposition method (SVD), the system:

\[ \begin{bmatrix} u \\ v \\ 1 \end{bmatrix} = M \begin{bmatrix} x \\ y \\ z \end{bmatrix} \]  

(1)

where \( M \) is the homogeneous matrix of each camera. 

\((x, y, z)\) coordinates of a point in \( R \) frame and \((u,v)\) the coordinates of its projection on the camera image.

We chose \( R_{frame} \) as \( R_{com} \).

2-2 Tracking the probe : \( T_{probe} \)

In order to get position and orientation of the probe we have installed 3 LEDs and we have measured their coordinates in the \( R_{probe} \) frame.

From the images obtained by the cameras, the LEDs are paired using the epipolar constraint and their 3D coordinates are calculated in \( R_{com} \).

Then, for each position of the probe, \( T_{probe} \) is evaluated using the quaternion-based algorithm, which is more efficient in the three dimensions case than the SVD method [6].

2-3 Ultrasound Calibration

Calibration involves finding the displacement and rotation of the US image reference axis from the probe reference, in order to form the matrix of \( T_{US} \) which gives the coordinates of points of the US image into the probe reference.

A three-wire phantom [7], [8] (with metallic wires of 1mm diameter), made from PVC was used.

We have put 3 LEDs on the phantom and we measured their 3D coordinates in the phantom frame (figure 2).

The LEDs are used to calculate \( M_{\text{phantom}}^{\text{com}} \) the matrix of \( T_{\text{phantom}} \) the rigid transformation from \( R_{\text{phantom}} \) : phantom frame To \( R_{\text{com}} \).

A point in the US image reference \( R_{US} \) is transformed in a point in phantom reference \( R_{\text{phantom}} \) by:

\[ \begin{bmatrix} x \\ y \\ 1 \end{bmatrix} = M_{\text{com}}^{\text{phantom}} M_{\text{com}}^{\text{Probe}} M_{\text{Probe}}^{\text{US}} \begin{bmatrix} u \\ v \\ 1 \end{bmatrix} \]  

(II)

\( M_{\text{com}}^{\text{phantom}} \) is given by the stereovision system (2-2)

Each wire of the phantom is scanned along its length from a variety of directions. The wire appears as a detectable spot in the US image.

As each wire is aligned on an axis, two zeros components appear on the left hand side of equation (II), they give two equations with the six unknowns 3 rotations \( R(\alpha, \beta, \gamma) \) and 3 translations \( T(x, y, z)^t \) of the matrix.

Extraction of the 2D coordinates of intersection point is done automatically by thresholding the image, to extract the region of interest, and then computing its centre of gravity.

For \( N \) images, we obtain \( 2N \) equations stacked in a nonlinear system and solved by the Levenberg-Marquardt algorithm [9] by minimization of RMS between positions extracted from images and positions on axes.

3 Results

In order to evaluate our system, we have first evaluated the precision at each step.

3-1 Camera calibration

We have used 56X2 points of the calibration frame to calculate the homogeneous matrix of each camera. The coordinates of the 112 remaining points are retro-projected on the camera’s image.

We have compared the real positions and the retro-projected we have obtained:

<table>
<thead>
<tr>
<th>( \Delta u ) (pixel)</th>
<th>( \Delta v ) (pixel)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.145</td>
<td>0.119</td>
</tr>
<tr>
<td>0.272</td>
<td>0.298</td>
</tr>
<tr>
<td>0.073</td>
<td>0.053</td>
</tr>
<tr>
<td>0.052</td>
<td>0.071</td>
</tr>
</tbody>
</table>

Table 1 Errors of camera calibration

Combining the 2D information resulting from each camera, the stereovision system is able to measure the 3D positions of LEDs with an accuracy of about 0.2 mm at a rate of 24.192 MHz.
3-2 Localization of the probe

The matrix $M_{\text{Probe}}^{\text{com}}$ calculated for each position of the probe is evaluated using a micrometric bench. For 250 displacements in translation, the absolute error of position is independent of its position and its orientation (0.15 mm on average) and for 60 rotations measures, the absolute angular error remained lower than 0.55°.

3-3 Ultrasound System calibration

We have used a Sonosite hand-carried ultrasound system with a 5 Mhz convexe probe on which we have installed 3 LEDs (figure 3).

![Fig. 3 Ultrasound System with LEDs](image)

We have calculated $M_{\text{US}}^{\text{Probe}}$ as described in 2-3. To evaluate reproductability of the results, we have done 10 different computations and we compared parameters of the matrix, we obtained:

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Mean</th>
<th>Standard deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\alpha$ (degree)</td>
<td>-0.10</td>
<td>0.05</td>
</tr>
<tr>
<td>$\beta$ (degree)</td>
<td>-1.25</td>
<td>0.012</td>
</tr>
<tr>
<td>$\gamma$ (degree)</td>
<td>-29.62</td>
<td>0.001</td>
</tr>
<tr>
<td>x (mm)</td>
<td>-85.50</td>
<td>0.53</td>
</tr>
<tr>
<td>y (mm)</td>
<td>34.54</td>
<td>0.44</td>
</tr>
<tr>
<td>z (mm)</td>
<td>-165.20</td>
<td>0.13</td>
</tr>
</tbody>
</table>

Table 2 parameters of ultrasound calibration

3-4 Localization system

To measure the accuracy of the 3D position estimated from ultrasonic images we have performed two tests:

Test 1: A cylindrical phantom of 100 mm length and 12 mm diameter has been put in water path and the 3D position of its tips measured from ultrasonic image (figure 4).

![Fig. 4 :Ultrasound images of test object](image)

The tips were manually specified from images and their positions were estimated from 10 images taken with different probe positions and orientations. Calculated dimensions have given average length of 100.40 mm and average diameter of 12.13 mm.

Test 2: In this test, 2 intersecting wires have been used. The 3D position of intersection point was measured both from ultrasonic images and by the stereovision system with a pen equipped with a LED. Then using the micrometric bench, we have found displacements between 5 mm to 30 mm and we have measured the coordinates of the new position at every time from the US image. Errors in the new coordinates calculated remained lower than 0.4 mm (figure 5).

![Fig. 5 Variations in positions with different displacements](image)

4 Experiments

4-1 Patient setup in radiation therapy

During conformal radiation therapy sessions, the patient must be positioned 20 to 40 times. The most common method of reproducing his position uses skin marks and isocentric lasers in treatment room.

However, for the prostate this repositioning can prove to be insufficient, indeed the prostate can move up to 30 mm from the marks.

For this case, our system can improve the setup. To achieve this aim, it is enough to locate the isocenter of the treatment machine in the camera calibration frame using the lasers.

Once beam coordinate calibrated, localization can be done in the isocenter frame and the patient position can be checked according to the position of the prostate extracted from US images (figure 6).

![Fig. 6 US image with prostate outlined](image)
Matching with pre-operative data

Intra-operative visualization of pre-operatively defined data have become a field of considerable interest. Usually, preprocessed diagnosis image data obtained from CT or MR serve as basis for planning and reconstruction of 3D models. These models are centered in the MR/CT coordinate system. Intra-operative registration is necessary to relate these coordinates to those from operating room.

We have used this method of localization to register US images of prostate with MR/CT images. We used the spatial information given by the localization system to reslice the MR/CT reconstructed volume in the same orientation than the US image using a multi-planar reconstruction algorithm (MPR) (figure 7). Then prostate boundaries are segmented from the two images modalities and aligned by a rigid transformation obtained after optimization, guided by the Iterative Closest Point (ICP) algorithm [6].

The advantage of this method is its velocity because it avoids doing a 3D segmentation from the volume since we obtain a novel pre-operative in the same plan than the intra-operative one.

![Fig 7. Pelvis reconstructed and MPR trace](image)

5 Conclusion

A robust localization system based on stereovision and freehand ultrasound images has been proposed. Two cases of applications, where the system had given good results, are presented. Calibration of both camera and ultrasound are based on phantoms easy to use and simple to build. An other attractive feature of the system is its compatibility with other instruments being usually in the treatment room.

6 Acknowledgement

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References


