Introduction

Registration of intraoperative ultrasound to preoperative CT data can be used to non-invasively locate bones of interest and allow for computer assisted surgical navigation of the operative field. Ultrasound (US) images of the pelvic surface, along with the respective position and orientation of each image are processed to provide a set of 3D points which define the location of the bone surface. The registration algorithm begins with an initial estimate from three anatomic landmarks and uses an adaptation of the iterative closest points (ICP) algorithm to refine the registration result [1]. The accuracy of the registration method was tested on an anatomic pelvis model using a fiducial based registration as the ground truth registration result. The development of minimally-invasive surgical techniques would benefit from the ability to non-invasively register the intraoperative anatomy to a preoperative surgical plan.

Methods

Ultrasound images were collected along with their 3D position and orientation from the optical tracking system (Optotrak, Northern Digital, Toronto, Canada)[2,3]. The ultrasound images are acquired via video capture of the output of the ultrasound machine (Siemens SONOLINE Elegra, Issaquah, WA). An optical tracker was rigidly attached to a molded shell that fits precisely over the ultrasound probe. This allowed measurement of the position and orientation of each of the ultrasound images with respect to the operating room coordinate system. The CT images were reconstructed into a 3D surface model. To register the pelvis, the algorithm begins with an initial alignment using three anatomic landmark points (pubic tubercle, and the anterior and posterior superior iliac spines) to provide a starting point for the registration process. These three points are collected on a physical model with a graphical user interface during the development of the surgical plan. The best fit between the three paired anatomic landmark points is computed with a closed form least squares solution [1], producing the initial registration estimate. The ultrasound images are processed to recover the location of the bone surface reflection. The bone surface reflects nearly all of the ultrasound energy, thus typically the bone surface produces a bright contour followed by a dark shadow deep to the bone surface in the ultrasound image. The image gradient between the bone shadow and the bone contour is located with a directional edge detector (Figure 1). The bone contours in each US image are then converted into a 3D point set suitable for registration with the iterative closest points algorithm. The ultrasound images contain artifacts inherent to the imaging process which result in error points that do not represent the bone surface. These error points are eliminated based on the residual errors (distance form the bone surface) during the registration process.

To simulate the effect of the overlying muscle and fat in real ultrasound images, random noise was added to the region between the bone contour and the skin surface. From these simulated ultrasound images the bone contours were reliably detected with the method described above.

![Figure 1. Ultrasound image of patient anatomy and two sequential steps of the edge detection method used to recover the bone contour.](image)

![Figure 2. Strips of points from ultrasound images following registration to pelvic surface model.](image)

Table 1. Maximum Error in the Registration Result

<table>
<thead>
<tr>
<th>Axis</th>
<th>Initial Offset</th>
<th>RMS Error</th>
<th>Error in Each Axis</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rotation (degrees)</td>
<td>10 10 10</td>
<td>0.32 0.98 -0.67</td>
<td></td>
</tr>
<tr>
<td>Translation (mm)</td>
<td>20 20 20</td>
<td>0.96 1.98 -2.37</td>
<td></td>
</tr>
</tbody>
</table>

Discussion

Using the above method to simulate muscle and fat layers in ultrasound images of a phantom pelvis, as few as 20 images over the iliac wing were sufficient to obtain clinically acceptable accuracy. Currently, the registration accuracy is limited by inaccuracies of the calibration protocol used to define the spatial relationship of the tracked US probe and the US image plane. Refinements of the bone contour detection method may also further improve the overall accuracy.

References


Acknowledgements: This work was supported by the Medical Robotics and Computer Assisted Surgery Fellowship of the Shadyside Hospital Foundation.