Efficiently Computing and Registering Digitally Reconstructed Radiographs

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ABSTRACT

Generation of digitally reconstructed radiographs (DRRs) is computationally expensive and has been cited as the rate-limiting step in the execution time of intensity-based two-dimensional/three-dimensional registration algorithms. This paper considers the problem of generating DRRs by conventional ray tracing. Experiments confirm that good quality reconstructions can be obtained using this approach in a few seconds. We evaluate the approach for automatic patient setup prior to radiotherapy treatment by performing intensity based 2D-3D registration using normalized cross correlation. Preliminary results using a pelvic CT data set show the method is accurate to about ± 2 pixels (i.e. ± 0.3 mm).

1   Introduction

Advances in diagnostic imaging technology [5] and the increased availability of computer tomographic (CT) and magnetic resonance imaging (MRI) enables many invasive and non-invasive surgical procedures to be pre-planned. Computer assisted planning is routine in radiotherapy treatment and is increasingly being used to plan complex surgery. Indeed, virtual surgery may now be practiced in a number of teaching hospitals and surgical simulation is an essential stage if robotic surgery is being considered. If a volumetric data set has been used for planning radiotherapy treatment or robotic surgery the patient must be accurately positioned when the procedure is carried out to ensure that the prescribed treatment is delivered as planned. Where the anatomy is rigid (e.g. head) this may be achieved by using mechanical fixations (e.g. stereotactic surgery), with radiographs used for quality assurance. More generally, images acquired prior to surgery may be used directly for patient positioning. For example, in radiotherapy treatment, radiographs captured immediately before treatment delivery (so called portal images), perform an important role in patient setup protocols.

Although recent advances in technology have enabled acquisition of 3-D volumetric data immediately before surgery (e.g. using cone beam [1], portable or 'open' CT scanners) patient setup for radiotherapy treatment is usually performed by registering 2-D images. In this case a 2-D radiograph, reconstructed from the volumetric data set is matched with a conventional radiograph (or portal image) acquired immediately before the patient undergoes treatment (see Figure 1). To estimate the pose of a patient, radiographs must be quickly reconstructed from 3-D CT data and automatically compared with those captured conventionally. Figure 2 shows a schematic representation of this approach; the blocks representing transformation, reconstruction and similarity measure forming the focus for this investigation. The paper proceeds as follows, section 2 considers the problem of image reconstruction and section 3 reviews some of the more popular approaches to image registration. Preliminary results of an experimental evaluation of one approach is presented in section 4 and finally some conclusions are drawn and further work is identified.

2   Reconstruction of Digital Radiographs

Two main processes are responsible for the absorption of X-ray photons by matter: the photoelectric effect and Compton scattering [4, 6]. The mathematical expression describing this process is

$$I = I_0 \exp^{-\mu x}$$

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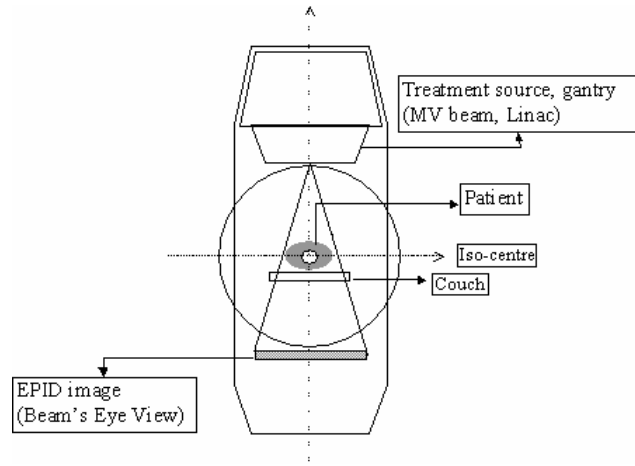
Figure 1: Image capture for patient setup before radiotherapy treatment
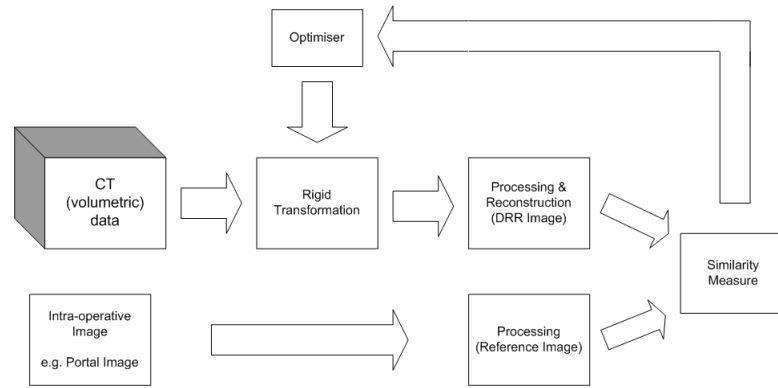

Figure 2: Schematic representation of image based pose recovery

where I_0 represents the intensity of the incident ray, μ the linear attenuation coefficient, and x the thickness of the material. The linear attenuation coefficient relates the attenuation of the beam due to all attenuation processes, i.e.

$$\mu = \mu_{en} + \mu_{scat}$$

where μ_{en} represents the part of the total linear absorption coefficient attributable to the photoelectric effect and u_{scat} that due to Compton scattering. μ is influenced by the atomic number of the material (Z), the material density (ρ), and the photon energy (E). The mass attenuation coefficient (μ/ρ) represents the radiation removed from a beam of unit cross-sectional area by a medium of unit mass; typical values are tabulated [10].

CT data is typically represented as an 3-D array of values, each representing a CT number

$$\text{CT number} = 1000 \left[\frac{\mu_i - \mu_w}{\mu_w} \right]$$

where μ_i is the attenuation value of a particular volume element of tissue (voxel) and μ_w is the linear attenuation coefficient of water for the average energy in the CT beam. The reconstruction of radiographic projections from CT data has been addressed by many authors [7, 8] and is usually accomplished by simulating the attenuation of a ray passing through the CT volume by incrementally summing the attenuation due to each voxel as shown in Figure 3. i.e.

$$I = I_0 \exp^{\Sigma - \mu_i x_i}$$

where μ_i represents the linear absorption coefficient and x_i the path length of a ray intersecting voxel i. The intersection of a ray with a (generic) cube is a fundamental operation that is efficiently solved in many elementary texts on computer graphics (e.g. [11]). A key step is to establish a parametric representation for the ray such that any point along it can be calculated by varying t. i.e.
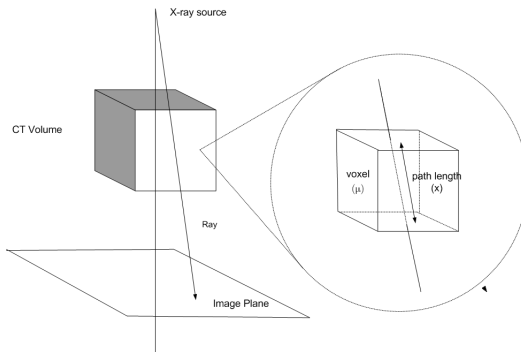
$$r(t) = source + \overline{D}_{rc} t$$

Figure 3: X-ray projection formed by raytracing

where *source* represents the position of the x-ray source and $\overline{D}_{rc}$ represents the direction of a ray striking the image plane at row $= r$, column $= c$ (the rc^{th} ray). The intersection of the ray with any cube arbitrarily positioned in the scene can then be calculated by computing an affine that will transform the cube (and the ray) to the generic geometry (i.e. cube at the origin). Similarly, accessing the CT number for a particular voxel can be accomplished by computing an affine that transforms the CT volume (cube) to a geometry that directly yields the 3-D array index. Unfortunately this task is computationally complex. Two forward and inverse affine matrices are needed for the CT volume (generic + index) and one generic forward and inverse affine matrix is needed for each intersected voxel. For a CT volume comprising 256 x 256 x 256 voxels, approximately 772 (256 x 2 + 256 + 4) affine transformations are needed (each requiring 16 floating point multiplications/additions) per ray. Accurately forming an x-ray image comprising (say) 256 x 256 rays takes several minutes and represents a significant processing bottleneck. This has motivated researchers to seek fast approaches to generating DRRs ([9] represents a recently published example).

Leaving aside [9], there are two main techniques that can be used for estimating the DRR. We will refer to these as *shear warp* and *ray sampling* approaches. Shear warp [14] applies a transformation to the volume such that all the rays intersect orthogonally. Thus, voxels can be summed along rows or columns (after interpolation). This method is attractive as it does not require any ray tracing (as such) and easily be implemented in MATLAB. However, although the implementation is compact, we found it to perform only slightly quicker than the most accurate approach. Ray sampling applies the index transform to the parameterized ray so that points on the ray are converted directly into array indices for the volume. Using this technique, provided the number of samples is similar to the number of intersected voxels, the summation can be performed with reasonable accuracy. As only one affine transformation is needed per ray the reconstruction takes only a few seconds (volume = 256x256x256, radiograph=256x256). A ray tracing demonstrator has been implemented [13] using the CImg image processing library [12].

3   Image Registration Techniques

We have developed an interactive GUI using MATLAB which uses a normalised cross-correlation index or manually placed point correspondences to match two radiographs. Images can initially be assessed interactively using a *split and merge* display mode (Figure 4). The normalised cross correlation index

$$\gamma(u,v) = \frac{\sum_{x,y}[f(x,y) - \overline{f}_{u,v}][t(x-u,y-v) - \overline{t}]}{\{\sum_{x,y}[f(x,y) - \overline{f}_{u,v}]^2 \sum_{x,y}[t(x-u,y-v) - \overline{t}]^2\}^{0.5}}$$

where $\overline{t}$ is the mean of the feature and $\overline{f}_{x,y}$ is the mean of $f(x,y)$ in the region under the feature, is used for automatic image registration. This similarity measure was chosen as it is a well established technique that has been implemented in MATLAB's image processing toolbox. The algorithm searches the similarity surface (Figure 5 for the best match, in this case quite obvious. This approach registers the images in a few seconds (MATLAB implementation); these times seem to confirm the view that DRR is the major computational expense in 2D/3D image based registration.

4   Results

Figure 6(a) shows an example DRR produced by our implementation. The imaging geometry can be changed by moving the image plane, rotating the gantry, or moving the couch (Figure 7). The image plane simulates
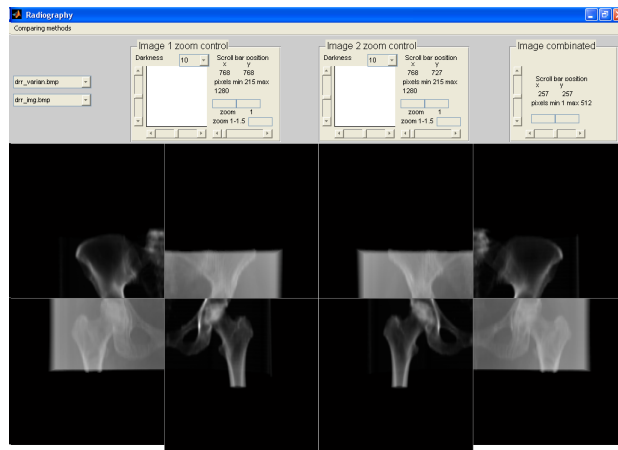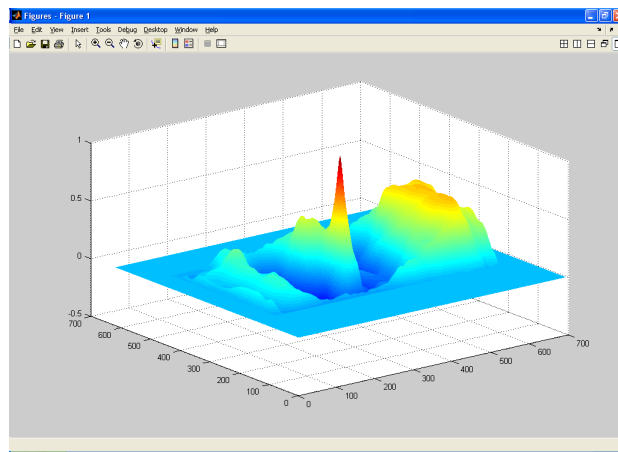
Figure 4: Interactive Split and Merge GUI

Figure 5: Correlation surface

a Varian A500 amorphous silicon detector (ASD); 40x30 cm. comprising 512x384 detectors. Feedback is provided in an associated window (Figure 6(b)). To test the approach we generated a series of DRR images corresponding to a couch translations of ± 5 mm. Figure 8 shows this translation was recovered by the normalized cross correlation image registration. This is represented graphically at Figure 9

5   Conclusions

We presented some preliminary results using an image processing technique to estimate patient setup error prior to radiotherapy treatment. The method uses a ray tracing approach to digitally reconstruct radiographs from a CT data set. Image registration was performed using normalised cross correlation. Preliminary results seem to be consistent with those reported in earlier studies carried out by others [15–17]. Further work is needed to assess the effect of varying some of the parameters; for example t_{step} greatly affects the performance of the ray tracing algorithm and may not be too critical to the performance of the registration algorithm. A more rigorous evaluation is required, ideally using a calibrated phantom, and other approaches to image registration (e.g. using Mutual Information) could be explored.

Acknowledgements

Figure 6: (a) Example DRR, (b) 'Tellback' display
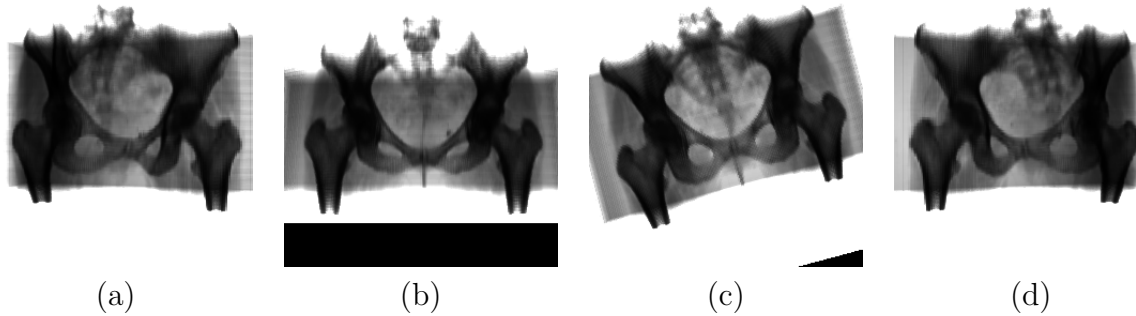
Figure 7: Pose examples: (a) Gantry +15° (b)-(d) Couch +15°, +15°, −15°, (pitch, yaw, roll)

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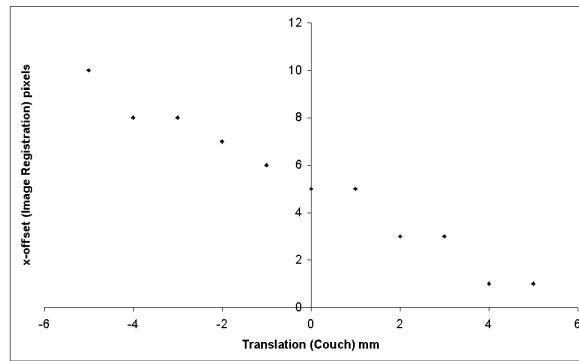| Couch Position (mm)	+5	+4	+3	+2	+1	0	-1	-2	-3	-4	-5
Corr. Coef.	0.759	0.758	0.756	0.750	0.752	0.749	0.743	0.749	0.743	0.751	0.752
x-offset (pixels)	1	1	3	3	5	5	6	7	8	8	10
y-offset (pixels)	-41	-40	-41	-41	-41	-41	-41	-41	-41	-42	-41

Figure 8: Result of image based registration

Figure 9: Graph representing recovered translation

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