Accuracy of co-registration of planning images with Cone Beam CT images

WHITE PAPER

Introduction

Leksell Gamma Knife® Icon™ with integrated Cone Beam CT (CBCT) system provides increased workflow flexibility with e.g. planning on frameless images and subsequent frameless dose delivery. The stereotactic reference is given by the CBCT images taken prior to treatment. Co-registration of the planning image volume with the CBCT volume gives the transformation mapping of the planned isocenter positions to stereotactic coordinates.

The accuracy of co-registration depends on mutual information, implying that deteriorating effects i.e. scatter and noise in the CBCT images may affect the accuracy of the co-registration. Also, co-registration accuracy will depend on parameters in the reconstruction algorithm, as well as specific co-registration parameters. Thus it is important to quantify the uncertainty and to optimize relevant parameters in the algorithms to ensure that high positional accuracy is maintained in frameless workflows in radiosurgery.

Materials and methods

To estimate the co-registration accuracy, existing MR and CT images of a large set of patients have been used. Based on the CT images, CBCT images have been synthetically generated. In order to do so, the CBCT system and the physical processes, relevant to creating the projections, must be accurately modeled. Only CTs covering the entire head and with as little artifacts as possible (streaks, beam hardening) are used to avoid introducing non physical errors. The workflow of generating synthetic images is shown in Figure 1.

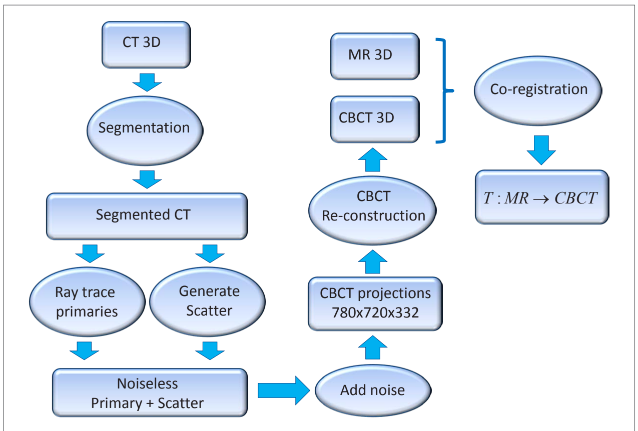

Figure 1. Workflow for generating synthetic images.

The first step is to apply a simple threshold algorithm to segment the CT volume into air, soft tissue and bone based on the Hounsfield values.

For a large set of energies the total attenuation is calculated for every voxel in the segmented volume. This information is needed in the ray-tracing step.

In the second step the photons from the X-ray source are ray-traced through the segmented volume to the pixels of the detector for every projection. The spectral shaping effects of the bowtie filter are included, as well as the detailed attenuation information of the segmented volume. The effect from scattering in the patient is Monte Carlo simulated in much fewer projection angles and tallied in larger pixels than in the actual detector. Using a method developed by G. Bootsma at Princess Margaret Hospital [1] the scatter is subsequently filtered and up-sampled to the resolution of the detector and to all projections.

The third step is to add Poisson noise to every projection. Noise is proportional to one over the square root of mAs and depends on the intensity value at each pixel of the detector. It is assumed that the intensity is proportional to the energy fluence.

After being generated, the projection images are reconstructed and the reconstructed volume is co-registered with MR images of the same patient using the Leksell GammaPlan® rigid co-registration implementation.

The registration uses Normalized Mutual Information (NMI) [2] to estimate the level of correspondence between two images given a registration transformation, and a numerical optimizer based upon Simulated Annealing to find the best such transformation[3]. The NMI is computed using histogramming and a stochastic sampling of the image intensities using linear Partial Volume Interpolation [4]. The most crucial parameter for the registration is the number of sample points used to compute the normalized mutual information. MRI images seem to require a large number of samples to reduce the variability in the Mutual Information measure. This is likely because of the geometric distortions typically present in MRI images. To reduce the uncertainties, a refinement-step has been added which fine-tunes the result using a much larger number of samples.

To numerically assess the uncertainties in the registration without knowing the correct transformation with sufficient accuracy, the estimated mean Target Registration Error (TRE) has been used. The target points are randomly selected within a sphere approximating the head. The mean TRE is the mean distance between the registered position of each target point and the average registration position of that point, see Figure 2.

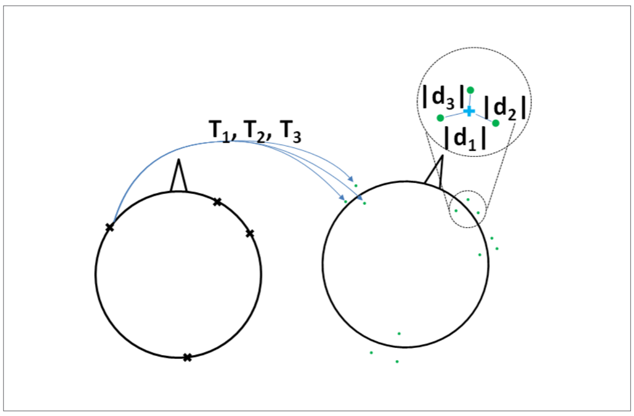

Figure 2. The estimated mean Target Registration Error (TRE) is the mean distance from all transformed points to the mean positions over all transformations and all target points.

Experiments on other images with known transformation have confirmed systematic errors to be negligible.

Results

The mean TRE appears fairly constant for CBCT-MRI registration of different patients, around 0.3mm for Patient 1-4 which all have an MRI resolution of 1mm^3 (see Table 1 which also contains results for corresponding MR-CT registrations). Patient 5 has a higher TRE which seems to correspond with the 50% larger slice distance for this patient. In Figure 3, histogram of the 3D positional errors for a large number of target points and registrations is shown for patient 1. As is shown there will be positions that have a significant larger positional error than the mean error. However, the large errors generally occur far from the center of the head and a substantial part of the head will have much lower positional error as shown in Figure 4. Here, the errors are transformed to the original MR image.

Another general finding of CBCT-MR co-registration is that the accuracy is in general sub-mm but the accuracy is less than for CT-MR registration. This is likely due to the effect of scatter that deteriorates CBCT images but are not present in CT images. However, adding the refinement step makes the CBCT-MR registration as accurate as the CT-MR registration without this step. CBCT-CBCT registration which has relevance for fractionated treatment is very accurate.

As for physical parameters the analysis shows that the accuracy is only slightly improved for mAs/projection exceeding 0.4. Moreover, tests performed at different kVp's show that the accuracy is approximately the same in the 80-100 kVp range assuming the equivalent dose being deposited in the patient. Finally, applying different filter types and crop frequencies have little effect on the accuracy.

Patient	Estimated TRE (std dev) [mm]	
	CBCT-MR	CT-MR
Patient 1 (1 mm resolution)	0.35 (0.01)	0.16 (0.006)
Patient 2	0.33 (0.01)	0.19 (0.007)
Patient 3	0.31 (0.01)	0.17 (0.006)
Patient 4	0.34 (0.01)	0.16 (0.006)
Patient 5 (1.5 mm slice dist.)	0.54 (0.02)	0.26 (0.01)
Typical CBCT-CBCT/CT < 0.1 mm		

Table 1. Co-registration results for 5 different patients.

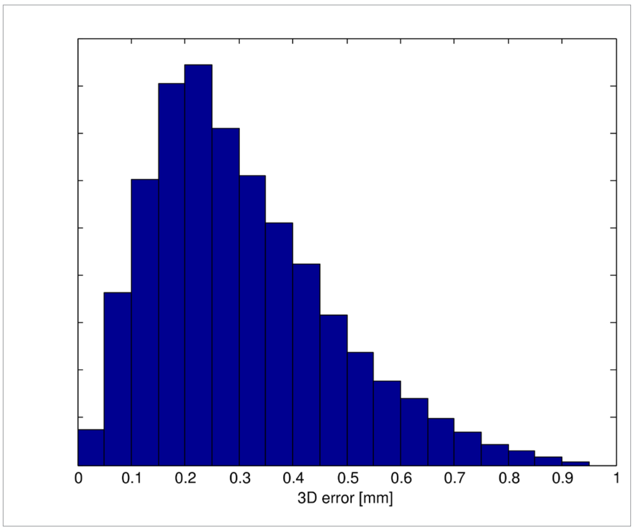

Figure 3. Histogram over 3D errors at different positions in the head for patient 1.

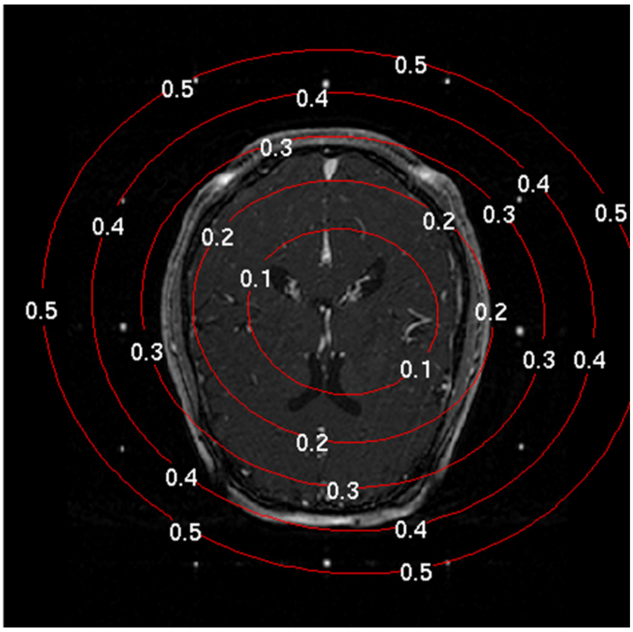

Figure 4. Euclidean norm of standard deviation (mm) in transformed target coordinates CBCT-MRI registration of patient 1.

Conclusions

Co-registration and reconstruction parameters have been optimized to minimize positional uncertainty. It is thus feasible to co-register MR and CBCT images with small uncertainties (in general sub-mm) in isocenter positions.

Future work

Future work aims at improving the physical modeling of the CBCT projections by incorporating detailed models of the detector to include effects from lag and ghosting. In addition to that, more sophisticated segmentation models of the CT images will be developed to more accurately simulate the photon transport through the head. Systematic studies of the effect of scatter and beam hardening on the co-registration accuracy will be performed and quantified.

ABOUT ELEKTA

A human care company, Elekta pioneers significant innovations and clinical solutions harnessing both external and internal radiation therapy for treating cancer and brain disorders. Elekta provides intelligent and resource-efficient technologies that improve, prolong and save patient lives. We go beyond collaboration, seeking long-term relationships built on trust with a shared vision, and inspiring confidence among healthcare providers and their patients.

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