White Paper

Design and performance characteristics of a Cone Beam CT system for Leksell Gamma Knife® Icon™

Introduction

Introducing an image guidance system based on Cone Beam CT (CBCT) and a mask immobilization to the Gamma Knife® system enables the flexibility to perform frameless and fractionated treatments. With the CBCT system mounted on the radiation unit, the CBCT images determine the stereotactic reference and the planning can be performed on non-stereotactic images that are co-registered to the stereotactic CBCT images.

This paper discusses design constraints and considerations to include a CBCT system in Leksell Gamma Knife® Icon™ and the implications this has on image quality. It also discusses CBCT system components.

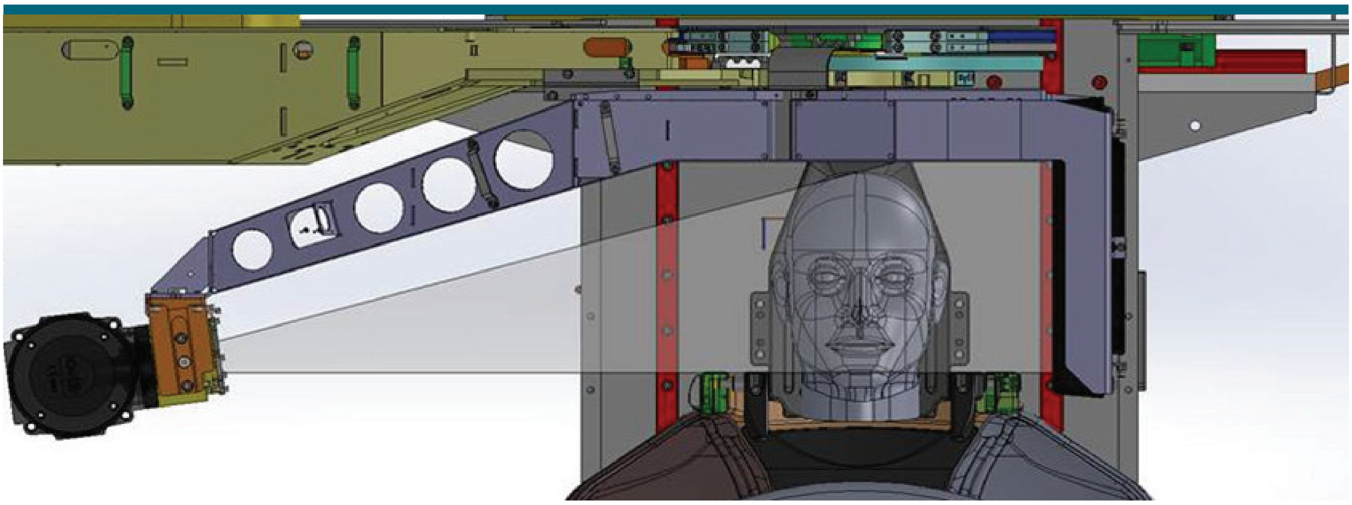


Figure 1.
The x-ray field is aligned with the frame fixation.

Design constraints

Since it is not possible to x-ray through the Patient Positioning System (PPS) or to fit an x-ray tube between the PPS and the patient's head, the limit of the rotational scan range of the C-arm is about 200 degrees. The reconstruction algorithm requires 180 degrees plus a cone angle (=197.4 degrees). Using a Parker weighting function the scan range is sufficient. An implication of the limited space is that the detector needs to pass between the PPS and the patient's head. The detector is close to the object being scanned, so a substantial amount of scatter is absorbed.

Another limitation is that x-ray beams cannot penetrate the frame fixation. By placing the x-ray tube so the field is aligned parallel and close to the frame plane, the field of view in the reconstructed images is as far down as possible (Figure 1).

Together these mechanical limitations affect the image quality and will be discussed later in this paper.

Geometrical characteristics

Table 1 lists the geometrical characteristics of the CBCT system. Most of the parameters are constrained due to the mechanical considerations. The source-to-axis distance is set so the flexing of the C-arm is low but the x-ray tube still is far from the object to give a large reconstructed field of view. The speed of the gantry is the maximum regulations allow.

Characteristics	Value
Source to axis distance (SAD)	790 mm
Source to detector distance (SDD)	1000 mm
Magnification factor	1.27
Reconstructed volume	224 x 224 x 224 mm^3
Cone beam angle	15°
Fan angle	16°
Scan time	30s
Flex	< 0.2 mm

Table 1.
The geometric characteristics of the CBCT system.

Sub components

The main components of the CBCT system are the X-ray tube and the detector. Their data are listed in Table 2.

Components	Properties
Detector	Layers: CsI, TFT (amorphous Si) 780 x 720 pixels (binned mode). Pixel resolution = 0.368 mm
X-ray tube	Energy range: 70-120 kVp. Spot size: 0.6 mm Weight: 17kg

Table 2.
Components.

Reconstruction algorithm

The reconstruction algorithm is an Elekta implementation of the commonly used FDK algorithm. It is implemented using CUDA libraries to make use of the GPU for short reconstruction times. It runs simultaneously with the scanning so the reconstruction is finished at the same time as scanning is completed.

The cone beam geometry together with a circular scanning path does not provide enough data for an exact reconstruction. This limitation in data results in "cone beam artifacts" when reconstructed with the FDK algorithm. The reconstruction is exact at the plane where the x-ray beam intersects the rotational axis orthogonally, i.e., the region closest to the frame fixation. The cone beam artifact increases with larger cone beam angle and makes flat structures in the xy-plane smeared out in the z-direction. Fortunately, there are not many such structures in the head. Combined with a limited scan range, the design—with the orthogonal beam close to the frame fixation giving a large cone beam angle—makes the artifact asymmetric.

Scatter

Scatter is a concern in CBCT imaging. Using fan beam geometry as in conventional CTs, most of the scattered radiation is scattered away from the detector. In a cone beam geometry the scattered radiation has a large possibility of interacting with the detector.

A common method to reduce the scattered radiation is to move the detector further away from the object. However, due to design constraints, this is not possible with this system.

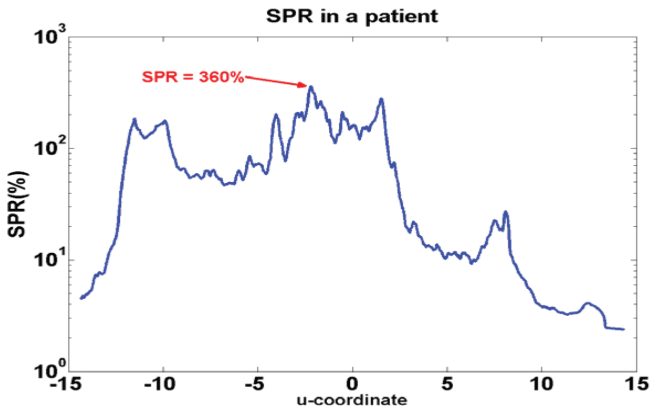
Another method to reduce scatter (and to homogenize the field at the detector) is to use a bowtie filter. To optimize a bowtie filter for this system Monte Carlo simulations were used where the x-ray tube and the radiation transport is modeled in detail. The optimization criteria were to give a flat response in the detector when a 90 mm radius cylindrical phantom (modeling an average sized head), made of water, is imaged in a position that matches the bowtie filter.

The scatter-to-primary radiation ratio along a line in an imaged anatomical phantom can be seen in Figure 2.

Image Quality

The main purpose of the CBCT images is to determine the position of a patient. The image quality is therefore optimized to enable a correct co-registration with planning MR images with the constraint to minimize dose to the patient.

Two scanning presets are given. In Table 3 the two different presets and their image quality implications are described.



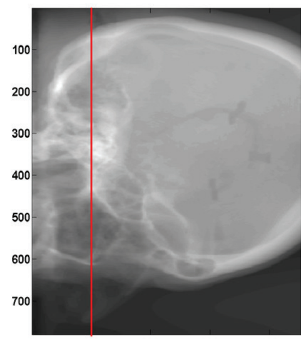


Figure 2.
The scatter to primary ratio along a line in an anatomical phantom.

	Preset 1	Preset 2
mAs/projection	0.4	1.0
kVp	90	90
Number of projections	332	332
Image volume (voxels)	448^3	448^3
Voxel size	0.5 mm	0.5 mm
Resolution	7 lp/cm	8 lp/cm
CTDI	2.5mGy	6.3mGy
CNR	1	1.5

Table 3.
The two presets.

The measures used to define the images include resolution, contrast-to-noise ratio, and linearity.

Resolution

There are several methods to measure resolution. One simple but subjective method is to count the number of line pairs, which gives a good estimation of the resolution. As can be seen in Figure 3, where a reconstructed image of the line pair section in the Catphan phantom is shown, 8 line pairs can be resolved. The main limitation of the resolution in this system is the voxel size. Since the voxel size is 0.5 mm the theoretical limit to what can be resolved is 10 line pairs per cm.

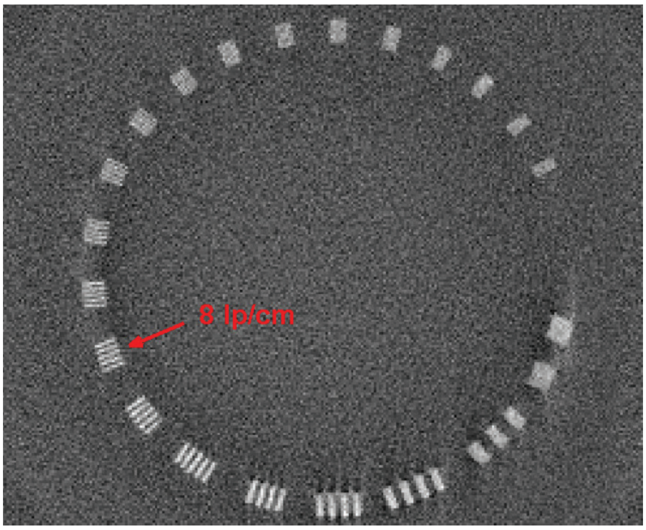


Figure 3.
Reconstructed line pair section of the Catphan phantom.

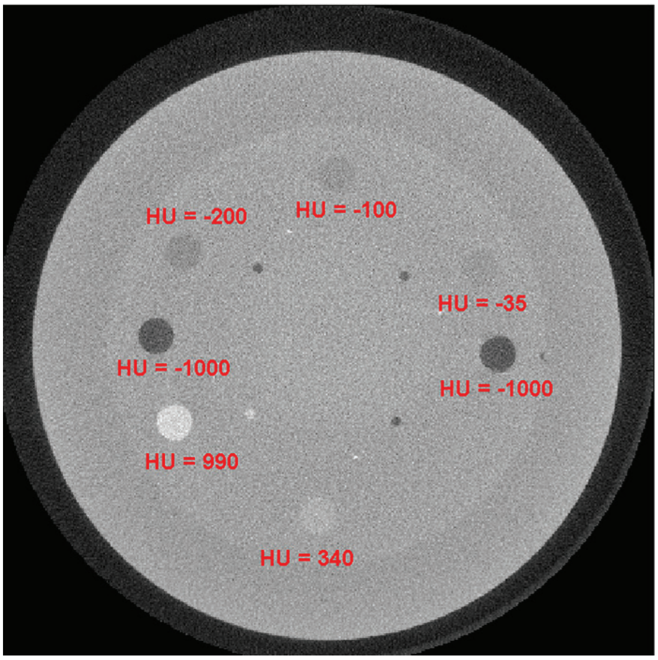


Figure 4.
The insert section of the Catphan phantom.

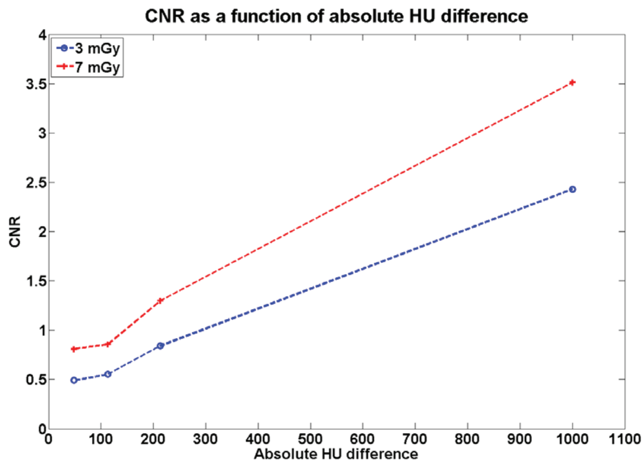


Figure 5.
The result of the CNR measurement for the two different presets.

Contrast

Contrast is a measure that captures how well objects can be distinguished from the background. If the noise is large, the object is hidden in the noise. The contrast-to-noise ratio takes this into account. It is defined as:

$$CNR = \frac{|S_k - S_{uniform}|}{\sqrt{\sigma_k^2 + \sigma_{uniform}^2}}$$

where S_k is the signal for material k and σ_k is the standard deviation of the signal at k. Figure 4 shows a reconstructed image of the insert part of the Catphan phantom and in Figure 5, measurement for the two presets can be seen. If the CNR is 1 for the lower preset, it is 1.5 for the higher preset.

Linearity

Linearity measures how homogeneous the images are. In Figure 6, a profile of a reconstruction of the homogeneous section in the Catphan is plotted. The higher intensities at the sides of the phantom are due to scatter radiation and the top is due to the beam hardening effect in the bowtie filter. This top is not as large as in a smaller phantom or a human head.

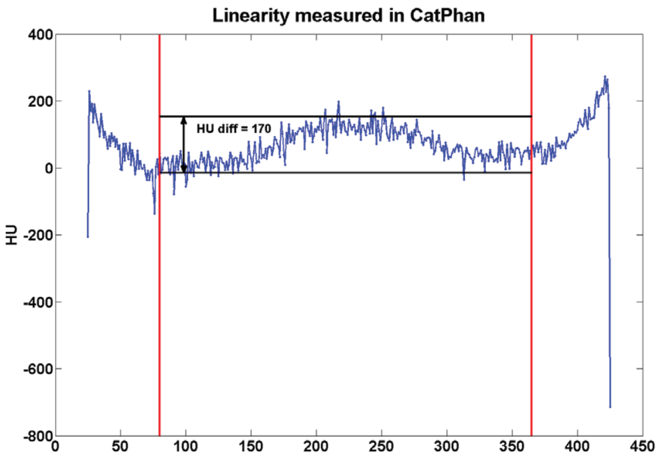


Figure 6.
A profile of the homogeneous section of the Catphan phantom plotted

Dose

CTDI is an integrated dose measure which gives a comparison between CT systems. Dose is measured at 5 positions in a PMMA phantom. One example can be seen in Figure 7 and the weighted CTDI dose is calculated as :

$$CTDI\_w = \frac{1}{3} CTDI_{100(center)} + \frac{2}{3} CTDI_{100\,(peripheral\ mean)}$$

The weighted CTDI for the system is 2.5mGy for the lower preset and 6.3mGy for the higher preset.

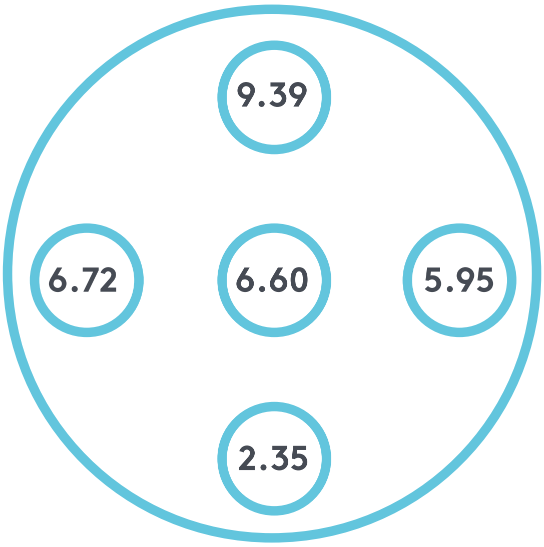


Figure 7.
Example: result of a CTDI measurement.

Summary and conclusion

To characterize the system, projection images of the Catphan quality phantom are reconstructed and evaluated. The image quantities used to evaluate the images are spatial resolution, contrast-to-noise ratio, homogeneity, Hounsfield unit correctness, and scatter to primary ratio. In addition to that CTDI is measured with a CTDI phantom and an ionization chamber.

Given the mechanical limitation imposed by the existing system, the image quality is well within the needs for patient positioning—so, the image quality allows for accurate patient positioning.

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