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Comparison of two detector systems for cone beam CT small animal imaging - a preliminary study

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Purpose

To compare two detector systems - one based on the charge-coupled device (CCD) and image amplifier, the other based on a-Si/CsI flat panel, for cone beam computed-tomography (CT) imaging of small animals.

A high resolution, high framing rate detector system for the cone beam CT imaging of small animals was developed. The system consists of a 2048×3072×12 bit CCD optically coupled to an image amplifier and an x-ray phosphor screen. The CCD has an intrinsic pixel size of 12 μm but the effective pixel size can be adjusted through the magnification adjustment of the optical coupling systems. The system is used in conjunction with an x-ray source and a rotating stage for holding and rotating the scanned object in the cone beam CT imaging experiments. The advantages of the system include but are not limited to the ability to adjust the effective pixel size and to achieve extremely high spatial resolution and temporal resolution. However, the need to use optical coupling compromises the detective quanta efficiency (DQE) of the system. In this paper, the imaging characteristics of the system were presented and compared with those of an a-Si/CsI flat-panel detector system.

Keywords

cone beam CT; CCD; flat panel; small animal imaging

1. INTRODUCTION

Computed-tomography (CT) system has become one of the most important tools in medical diagnosis since it was first introduced in the mid 1970s. The CT image quality has been improved with electronics and computer technology develops. Conventional CT techniques have been based on fan beam geometry and single or multiple array detectors. In the last decade, various cone beam CT (CBCT) techniques have been developed as high performance digital x-ray area detectors and computers became available. These techniques have been applied to obtain 3-dimensional (3D) and even 4-dimensional (4D) images of human patients as well as small animals.

Amorphous silicon CsI (a-Si/CsI) flat-panel (FP) detectors have become commercially available for primary chest and mammographic imaging.¹ It uses a layer of needle structured CsI:Tl scintillator coupled to an array of photodiodes on the a-Si plate. The

resolution and the contrast provided by a-Si/CsI FP detector based CBCT system are much higher than those by conventional CT. However, the pixel size of the detector for large field and high framing rate operation is still limited to about 150 μm .

As one of digital x-ray detectors, charge-coupled device (CCD) based x-ray detector has shown its own special characters and played an important role in the medical imaging.²⁻¹⁷ The current intrinsic pixel size on commercial CCD chip can be as small as 3 μm , and the data readout speed can be high up to 160 byte/s. Now, we are developing a high spatial resolution, high framing rate x-ray detector system for the CBCT imaging of small animals. In this preliminary study, the imaging characteristics of the system were presented and compared with those of an a-Si/CsI FP based CBCT system.

2. METHODS AND MATERIALS

2.1 CCD based cone beam CT system

A newly developed bench top CCD based CBCT system was used to acquire CT projection views and an algorithm based on filtered back-projection was implemented to reconstruct the images. The system consists of an x-ray tube, a rotary table, a phosphor screen, a piece of optical mirror, 2 sets of lenses, an image intensifier, and a CCD camera. All optical components and the image intensifier are enclosed in a lead shield black box. The system setup is illustrated in Figure 1.

In the CT scanning system, the scanned object was placed on a rotary table. The x-ray tube and the x-ray detector were fixed. When the rotary table rotated, x-ray passed through the object and arrived to the phosphor screen. On the screen, most of x-ray were absorbed and converted to visible light photons. The visible image was coupled to the image intensifier by the 45° mirror and the first set of lenses. Then the enhanced image was coupled to the CCD chip by the second set of lenses. Finally the data were acquired and processed in the computer as shown in Figure 1.

In the study, the x-ray was produced by a Varian G1592 x-ray tube (Varian Medical Systems, Salt Lake City, UT) and an Indico 100 x-ray generator (CPI Canada Inc. CMP, Georgetown, Ontario, Canada). The focal spot was 0.6 mm. The kVp was 50 kV. The x-ray tube was operated in continuous mode, and the radiation dose for each projection image was controlled by the integration time of the CCD camera. The phosphor screen was Kodak Lanex Regular intensifying screen (Eastman Kodak Company, Rochester, NY). The source-to-object distance (SOD) was about 600 mm and the source-to-image (screen) distance (SID) was about 800 mm. The image intensifier was ITT FS9911C (ITT Industries, Night Vision, Roanoke, VA) with the resolution of 64 lp/mm. The CCD was a Dalsa 6M18 full frame CCD camera (DALSA, Colorado Springs, CO). The computer had two 3.2 GHz Intel Xeon processors and 4 GB physical memory. The general characters of the CCD based CBCT system are listed in Table 1.

2.2 Flat-panel based cone beam CT system

In order to compare, the same x-ray tube, x-ray generator, rotary table, SOD, and SID were used for both of the two CBCT systems. The components in the dashed block in Figure 1 for

CCD based CBCT were replaced with a flat-panel x-ray detector, Varian PaxScan 4030 CB (Varian Medical Systems, Salt Lake City, UT). Its characters are compared with those of the CCD system in Table 1.

2.3 Cone beam CT images

A mouse sealed in wax was scanned by the two CT systems. Then the projection data were reconstructed with an FDK algorithm. The X-ray energy was 50 kVp. The X-ray exposure levels were 6 mAs/frame $\times$ 450 frames and 2.7mAs/frame $\times$ 300 frames for CCD based CBCT and FP based CBCT, respectively.

3. RESULTS AND DISCUSSION

In Figure 2, the projection images for the same mouse captured by two CBCT systems are shown and compared. The CCD based CBCT has a much higher spatial resolution.

However, the image contrast and the noise level are both worse than those in the FP based CBCT system. The reasons may be as the followings,

a). X-ray focal spot blurring

In the study, the x-ray source was part of the FP based CBCT system. The x-ray focal spot was 0.6 mm. The corresponding size on the detector was about 200 μm . However, the pixel size on the screen in the CCD based CBCT system was as small as 32 μm . As a result, the x-ray focal spot blurring effect was very serious. Obviously, the x-ray focal spot size and the image magnification factor should be carefully selected to optimize spatial resolution and temporal resolution in the future.

b). Exposure level

The pixel size ratio in the two CBCT systems was 1:6. The exposure level used to capture each projection image was 2.2:1. The x-ray photons which arrived to each pixel in the CCD system were far less than those in the FP system. This factor also lowered the signal to noise ratio (SNR) in the CCD system.

c). Inefficient light collection

In our prototype CCD based CBCT system, it was the two sets of optical lenses that were used to optically couple the x-ray image on scintillator to CCD chip instead of the optical fiber taper in prevalent CCD based medical imagers^{2–5}. The optical transmission efficiency of the lens was very poor, so the SNR was further deteriorated. Although the optical signal could be enhanced by using the image intensifier, the SNR was not improved because the noise was also amplified. To make the imaging system better, we need to improve the light transmission efficiency.

In the reconstructed images, Figure 3 (a) and Figure 4 (a) should be able to provide more detail information than Figure 3 (b) and Figure 4 (b). However, the image quality was deeply affected by the artifacts (white dots), which resulted from the noise and the scatter in the projection images. So, more research would be needed to improve the quality of the images.^{18–22}

4. CONCLUSION

We had successfully built a CCD based cone beam CT system and used it for small animal imaging. However, further improvement and optimization are required to achieve better spatial resolution, temporal resolution and contrast sensitivity.

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Figure 1.

Schematic diagram of the CCD based cone beam CT system. The components in the dashed line block were replaced with a flat-panel x-ray detector in the FP based cone beam CT system.

1200 x 1200 pixels, pixel size = 32 μm

(a)

320 x 880 pixels, pixel size = 194 μm

(b)

Figure 2.
Projection images acquired by (a) CCD based cone beam CT and (b) a-Si/CsI flat-panel based cone beam CT

1210 x 1210 pixels, voxel size = 25 μm

(a)

200 x 200 pixels, voxel size = 150 μm

(b)

Figure 3.

Reconstructed images from (a) CCD based cone beam CT system and (b) flat-panel based cone beam CT system.

voxel size: 50 μm

(In order to display the 3D image, the projection images had to be processed by 2x2 binning)

(a)

Voxel size: 150 μm

(b)

Figure 4.
3D maximum intensity projection (MIP) images of mouse head from (a) CCD based cone beam CT system and (b) flat-panel based cone beam CT system.

Table 1

Comparison of CCD based CBCT system and a-Si/CsI FP detector based CBCT system

Parameter	CCD based system	Flat-panel Detector
Maximum pixel number	3072×2048	2048×1536
Pixel depth	12 bit	14 bit
Intrinsic pixel size	12 μm	194 μm
Effective pixel size	Adjustable through lens coupling	388, 776 μm through binning
Detector size	$3.69 \times 2.46 \text{ cm}^2$	$40 \times 30 \text{ cm}^2$
Effective size	Adjustable through lens coupling	Same as above
Frame rate at full resolution	18 fps	7.5 fps
Scintillator	Gadolinium oxysulphide	CsI:Tl