Improving image quality of synchrotron CT by scattered X-ray correction

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Abstract. We developed a high-resolution CT system using synchrotron radiation. The system acquires high-quality CT images because the synchrotron CT has no beam-hardening effect that results from using monochromatic X-ray. It is said that X-ray scattering has no effects on the synchrotron CT images because a detector is set apart from a subject. However, the images have a disadvantage in non-uniformity caused by scattered X-rays owing to an area X-ray beam and a two-dimensional detector. The X-ray scattering decreased the accuracy of the CT number and increased the number of artifacts in CT images. We developed an accurate process for correcting scattered X-rays for the synchrotron CT. The correction process was applied for the synchrotron CT at 35-keV radiation. When a 10-mm-diameter 20%-hydroxyapatite homogeneous column phantom was used, the scattered X-rays led to a reduction of about 45% in the pixel values of CT images. With the correction, the pixel values were almost completely recovered.

1. Introduction

Dental cone-beam CT (Computed Tomography) can obtain two-dimensional and three-dimensional images quickly because it uses conical X-rays and an area detector (Fig. 1). The images are useful for planning dental implant and orthodontic treatments (Fig. 2). However, acquiring a high-precision relation between CT number (pixel values of CT images) and bone density is needed to improve the accuracy of planning.
We acquire the relation with a high-resolution CT system using synchrotron radiation at the High Energy Accelerator Research Organization, KEK. The system acquires high-quality CT images because the synchrotron CT has no beam-hardening effect that results from using monochromatic X-rays. It is said that X-ray scattering has no effects on the synchrotron CT images because a detector is set apart from a subject. However, the images have a disadvantage in non-uniformity caused by scattered X-rays owing to an area X-ray beam and a two-dimensional detector. The X-ray scattering decreased the accuracy of the CT number and increased the number of artifacts in CT images. The purpose of this study was to develop an accurate process for correcting scattered X-rays for the synchrotron CT.

2. Methods

2.1 Formulation of scatter-correcting function

First, we formulated an approximate scatter component, S, in a measured transmission image, as a function of a mean value, T, of the measured image, and a scatter parameter, k, which is defined as the ratio of scatter generation and attenuation coefficient.\(^1\) The k value is peculiar to the imaging apparatus and geometry. It depends on the irradiation area and the characteristics of the X-ray slit for scatter reduction and should be determined experimentally.

Equations 1, 2, and 3 were formulated.

True transmittance: \[ P = \exp(-\mu \cdot L) \] \hspace{1cm} ... Eq. 1

Scatter component: \[ S = \frac{\alpha}{\mu} \cdot P \cdot \{\exp[k \cdot L] - 1\} \] \hspace{1cm} ... Eq. 2

Measured transmittance: \[ T = P + S \] \hspace{1cm} ... Eq. 3

Here, \( L \) is a length through a subject, \( \mu \) is a scatter attenuation coefficient, and \( \alpha \) is a scatter generation coefficient.

Supposing that a scatter parameter \( k \) is \[ k = \frac{\alpha}{\mu}, \]
then scatter component \( S \) becomes \[ S = T - \frac{1}{1 + \frac{T}{\alpha}} \] \hspace{1cm} ... Eq. 4

True transmittance \( P \) is taken as \[ P = T - S \] \hspace{1cm} ... Eq. 5

2.2 Correction process

The correction process was developed using scatter correction functions (Fig. 3). In the correction process, all of the transmittance images were corrected by subtracting the scatter component calculated using the formulated function and the scatter parameter. Scatter correction of the CT images was done with a CT reconstruction process using the corrected transmittance images.

![Correction process diagram](image-url)
3. Results

3.1 Synchrotron CT Apparatus

The correction process was applied for the synchrotron CT at 35-keV radiation (Fig. 4). We measured 10-mm-diameter homogeneous column phantoms containing hydroxyapatite to examine the relation between CT number and bone density. The distance between a subject and an area detector was about 200 mm. The pixel size of the detector was about 0.015 mm. A scatter component was measured by using a very narrow X-ray slit for collimating the scattered X-rays. The height of the slit was 3 mm and the width was 15 mm. Without the slit, the contrast between air region and column region was decreased. When a 20%-hydroxyapatite column was used, scattered X-rays led to a shift of about 9% in the profile of the transmittance image (Fig. 5).

3.2 Scatter component

The scatter component was determined as the difference between a measured transmittance for a usual wide collimator gap (about 30 mm) and that for a very narrow gap (about 3 mm) with the slit (Fig. 5). An approximate scatter component in a measured transmittance image was formulated as a function of the pixel value of the measured image and a scatter parameter, as Eq. 4. The scatter parameter was acquired as the best fitting parameter for a plot of the pixel value versus the scatter component. As a result of Fig. 5, scatter parameter \( k = 0.35 \) was applied for a 20%-hydroxyapatite column (Fig. 6).
3.3 CT number of 20%-hydroxyapatite column

Average pixel values of the CT images were calculated at the center of the hydroxyapatite column (Fig. 7). With the X-ray slit, the relation between the rate of hydroxyapatite and the values was linear, as yellow dots ◇. The value of the 20%- hydroxyapatite column was 1076 HU with the slit, and decreased to 597 HU without the slit, as a red dot □. With the correction, the value without the slit was 1091 HU and was almost recovered, and it appeared as a pink dot △. The error from the CT number with the slit was sharply decreased to 1.4 % with the correction. High-contrast CT images were acquired with no artifact due to the correction. High-contrast CT images were acquired with no artifact due to the correction. (Fig. 8). If the size of the slit becomes small, the accuracy of the correction will be improved.

4. Conclusion

The proposed correction method is very effective for correcting non-uniformity due to the scattered X-ray effect. It enables us to acquire high-quality synchrotron CT images with an accurate CT number, high contrast and few artifacts. These results demonstrate that the dental cone-beam CT images can be used for accurate surgical planning for dental implant and orthodontic treatments.

References

[1] Baba R, Ueda K, Takahashi M, Nakano H and Maki K 2009 Medical Imaging Technology 27(3), 177-184