A bench-top micro-CT capable of simulating head motions

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Received: 7 December 2016 / Revised: 9 February 2017 / Accepted: 21 March 2017 / Published online: 3 April 2017
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Abstract Computational three-dimensional (3D) models of a dental structure generated from 3D dental computed tomography (CT) images are now widely used in digital dentistry. To generate precise 3D models, high-resolution imaging of the dental structure with a dental CT is required. However, a small head motion of the patient during the dental CT scan could degrade the spatial resolution of CT images to the extent that digital dentistry is no longer possible. A bench-top micro-CT has been built to evaluate the head motion effects on the dental CT images. A micro-CT has been built on an optic table with a micro-focus x-ray source and a flat-panel detector. A rotation stage, placed in between the x-ray source and the detector, is mounted on two-directional goniometers that can rotate the rotation stage in two orthogonal directions while the rotation stage is performing the CT scan. The goniometers can make object motions of an arbitrary waveform to simulate head tilting or head nodding. CT images of a phantom have been taken with and without introducing the motions, and the motion effects on the CT images have been evaluated. Object motions parallel to the detector plane have greater effects on the CT images than those against the detector plane. With the bench-top micro-CT, the motion effects have been visually seen at a tiny rotational motion as small as 0.3° . The bench-top micro-CT can be used to evaluate head motion effects on the dental CT images. The projection data, taken with the motion

effects, would be used to develop motion artifact correction methods for a high-resolution dental-CT.

Keywords Bench-top micro-CT · Digital dentistry · Motion artifact · Motion simulation · Dental-CT

1 Introduction

With the aids of three-dimensional (3D) printing technologies, rapid 3D prototyping is now widely spread to most of industrial fields. Computational 3D data for rapid 3D prototyping are made either by mathematical modeling or by 3D scanning of the object of interest. Rapid 3D prototyping is now widely adopted in digital dentistry for which 3D modeling of various dental devices and their computerized machining are needed [1, 2]. Computerized machining or printing of dental prostheses and dental implants has become a routine job in modern dental practice.

In most of digital dentistry, computational 3D dental data are made through optical scanning of real teeth or a teeth model made of plaster [3, 4]. Optical scanning of real teeth can be made by an intra-oral optic scanner, but it is known that an intra-oral optic scanner may introduce significant geometry errors to the 3D model when it comes to an entire teeth scan. When a computational model is made through optical scanning of a physical teeth model made of plaster, the plaster model often has shape errors caused by inaccurate impression of teeth on gums or by contraction of the plaster during the solidification. Any geometry errors in the computational model necessitate additional correcting jobs. Therefore, generating a computational model from high-resolution 3D computed tomography (CT) images is strongly desired. Unlike optic scanners, a 3D dental CT can

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show the periodontal structure as well as the tooth structure, thereby providing much richer information for dental practices [5–7]. If a computational model can be generated directly from the 3D dental CT images, digital dentistry would be more facilitated.

To use dental CT images for generating a computational model, the spatial resolution of dental CT images should be much improved over the current technology. There is a consensus that the spatial resolution of 3D CT images should be at least 100 μm to generate a usable computational model from CT images. Most of dental CTs are based on a cone-beam geometry in which a flat-panel detector (FPD) is used to acquire projection images at given sampling angles [8, 9]. Since the pixel pitch of FPDs, mostly 100–200 μm , is smaller than that of x-ray detectors in typical medical CTs, the detector integration time of dental CTs is rather long to secure reasonable signal-to-noise ratio. Therefore, the scan time of most dental CTs is around 10 s which is so long that patient motions may occur during the scan. Particularly in a dental CT scan, bigger motion artifacts may appear in CT images since the scan is performed with the patient sitting on a chair or standing [10, 11]. Any patient motions, caused by saliva swallowing, coughing, head tilting or nodding, etc., could compromise the spatial resolution of dental CT images. When it comes to high resolution imaging to generate a computational model, appearance of motion artifacts is a great concern since a small patient motion could introduce significant geometry errors to the 3D computational model.

Motion artifact corrections have been extensively studied in the field of image reconstruction [12–16]. In a medical CT scan, the patient's breathing and heart pumping are two major issues for motion artifact correction. In a dental CT scan, however, the patient's head motion, caused by saliva swallowing, coughing, tilting or nodding, is the major issue to be considered. Most of motion artifact correction methods for CT have been validated by visual inspection of the corrected CT images since motionless images are not available. However, visual inspection of the CT images would not be easy if the motion effects are small. Therefore, a physical simulation of head motions during the CT scan is necessary to develop motion artifact correction methods and to validate them. Recently, a physical simulation of head motions using a robotic arm holding a head phantom has been introduced to analyze the motion effects in a C-arm based cone-beam CT scan [17]. Using a robotic arm to make a physical simulation of head motions is a good approach, but it will be difficult to adopt a robotic arm if there is a little space between the human subject and the gantry like a dental CT.

In this paper, we introduce a bench-top micro-CT that can simulate head motions during the dental CT scan. To simulate head motions, a rotation stage, to be used for the

CT scan, is mounted on two-directional goniometers that can simulate head motions in two directions, i.e., head tilting parallel to the detector plane and head nodding against the detector plane. The motion effects are demonstrated in the CT images of a phantom, and the spatial degradations are evaluated from the images taken with introducing motions.

2 Methods

2.1 The bench-top micro-CT

A micro-CT has been built on an optic table as shown in Fig. 1. On the optic table measuring 110 cm $\times$ 80 cm, an x-ray source and an FPD are placed facing to each other. The x-ray source (Jupiter 5000, Oxford Instrument, U.K.) is mounted on a linear motor so that the source-to-detector distance (SDD) can be controlled. The x-ray source has an operating voltage of 10–50 kV with a maximum tube current of 1.0 mA. The focal spot size of the x-ray source is 50 μm and the x-ray exit window is made of 127 μm -thick Beryllium (Be). The FPD (Dexela 1512, Perkin Elmer, UK) has a pixel pitch of 74.8 μm and a matrix size of 1944 $\times$ 1536 making a sensitive area of 145.4 mm² $\times$ 114.0 mm². The FPD has an x-ray energy band of 12–160 keV and its CMOS readout circuit outputs 14-bit digital signal through Giga-Ethernet Vision (GigE) communication channel. The scintillator of the FPD is made of CsI.

A rotation stage is placed on the axis connecting the x-ray source and the FPD to make the CT scan when driven by a stepping motor. The rotation stage has a step angle of 0.10°, and it is mounted on the two-directional goniometers as shown in Fig. 2. The two directional goniometers are driven by two independent stepping motors, one for generating a rotational motion about the z-axis to simulate head tilting and the other for generating a rotational motion about the y-axis to simulate head nodding. The two-directional rotation can be controlled by a step angle of 0.05°. Four stepping motors, one to control SDD (XCVL6200-2-N-F-N, MiSUMi, Japan), another to control CT scan (RMPG60W-N, MiSUMi, Japan), and the others to simulate the head motion (GMPWG60-50-L-N, MiSUMi, Japan), can be driven independently. To drive the stepping motors, two control boards (MSCTL102, Soguru Seiki, Japan) are linked to a personal computer (PC) through a RS232C communication channel. The same PC controls the tube operation, including the tube current and tube voltage, through a serial communication link. The PC receives the x-ray projection data from the FPD through the GigE communication link. The maximum frame rate is 14 Hz when no binning is applied. In 2 $\times$ 2 and 4 $\times$ 4

Fig. 1 A photograph of the bench-top micro-CT consisting of an x-ray tube, a flat panel detector, a rotation stage mounted on an optic table

Fig. 2 The rotation stage mounted on two-directional goniometers. The rotation stage is for the CT scan, and the goniometers are for simulating head motions

binning modes, the maximum frame rate is increased to 56 and 86 Hz, respectively.

Figure 3 shows motion waveforms used in this work. An arbitrary motion waveform can be generated by the system, and the step-like waveforms are chosen to simulate typical head motions, jerky motions. Figure 3 shows the waveforms to simulate a jerky head motion in the middle of the scan. The jerky motions occur linearly over 50 view angles

Fig. 3 The motion waveforms simulating a jerky head motion

in this case. The length of the linear portion should be chosen depending on the motion speed and the frame rate of the CT scan. That is, a longer linear portion should be set for a higher frame rate for a given motion speed. To simulate a tiny head motion, the rotation angle was chosen to range from 0.3° to 1.2° in the figure.

A bench-top CT system and a dental CT system have different scan mechanism. In a bench-top CT system both the detector and the x-ray source are fixed while the object is rotating during the scan. On the other hand, the detector and source of a dental CT system are rotating while the object is fixed during the scan. To mimic the patient motion during a dental-CT scan in our bench-top system, we decompose the motion in the dental CT scan into two components about the z-axis and the y-axis of the bench-top system. For example, if the head motion during a dental CT scan is about the y-axis (head nodding), the head motion in the bench-top CT becomes:

$$\begin{aligned} M_{z\text{-axis}} &= M_{\text{dental}} \cos \theta \\ M_{y\text{-axis}} &= M_{\text{dental}} \sin \theta \end{aligned} \quad (1)$$

where M_{z-axis} and M_{y-axis} are the rotation angles about the z-axis and y-axis, respectively, in the coordinate of the bench-top micro-CT. M_{dental} is the head motion angle during the dental CT scan and θ is the view angle of the dental CT scan (the view number 450 corresponds to the view angle of 360°). Figure 4 shows the motion waveforms in the bench-top micro-CT when the head motion appears in a jerky waveform shown in Fig. 3.

2.2 Evaluation of the motion effects

To observe the head motion effects on CT images, a lab-built phantom has been used in the micro-CT imaging. The phantom has the shape of a half cylinder as shown in Fig. 5a, and it is made of polyethylene. The diameter of the phantom is 30 mm. This phantom is for evaluating the edge spread function when a motion is introduced during the scan. A CT image of the phantom, taken without introducing the motion, is shown in Fig. 5, and it will be used as a reference.

To observe the motion effects, CT scans were performed sequentially along with applying the motion waveforms shown in Fig. 4. Firstly, images were taken without introducing the motion, and they were used as reference images to evaluate the motion effects. After taking the motionless images, images were taken sequentially with increasing the motion amplitude. After taking images with or without introducing the motions, difference images between them were computed to evaluate the motion effects. Mean square errors (MSEs) over the difference images were computed as a metric of the motion effects. The phantom images were taken with a tube current of 0.9 mA and a tube voltage of 50 kV. The number of views was 450 and the frame rate of the detector was 5 Hz. 3D images, with the matrix size of $512 \times 512 \times 512$ and the pixel pitch of 100 μm , were reconstructed from the flat-field corrected

Fig. 4 The goniometer control waveforms in the bench-top micro-CT when a head motion in a dental CT appears in a jerky waveform shown in Fig. 3

projection data with the Feldkamp algorithm [18] and the center slices were chosen to evaluate the motion effects.

2.3 Motion correction

To verify the physical motion simulation is properly working, we correct the motion simulated projection data and reconstruct CT images from the corrected projection data. Since the effects of the motion against the detector plane are very small as compared to the one parallel to the detector plane, we correct the effects of a motion parallel to the detector plane only. At each view angle, we know the exact rotation angle about the z-axis (M_{z-axis}) from the motion control waveform. Therefore, the motion correction is made by rotating the projection data in the reverse direction by the known angle,

$$\begin{bmatrix} x_2 \\ y_2 \end{bmatrix} = \begin{bmatrix} \cos(M_{z-axis}) & -\sin(M_{z-axis}) \\ \sin(M_{z-axis}) & \cos(M_{z-axis}) \end{bmatrix} \begin{bmatrix} x_1 - x_0 \\ y_1 - y_0 \end{bmatrix} + \begin{bmatrix} x_0 \\ y_0 \end{bmatrix} \quad (2)$$

where (x_2, y_2) is the rotated coordinate of the original coordinate of (x_1, y_1) on the projection image. (x_0, y_0) is the point of rotation about the z-axis.

3 Results

Firstly, we made experiments in the coordinate of the bench-top micro-CT with the motion waveforms shown in Fig. 3. That is, the motions were introduced only in one direction, either parallel to the detector plane or against the detector plane. Figure 6 shows the difference images between the motionless image and the motion-affected images when the step-like motion waveform was applied about the z-axis, i.e., parallel to the detector plane. The amplitudes of the step-like motion were 0.3°, 0.6°, 0.9° and 1.2° for Fig. 6a, b, c and d, respectively. Even with a small motion angle of 0.3°, the effect of edge spreading can be seen clearly in the difference image shown in Fig. 6a. As the motion angle increases, the edge spreading also increases. In the respective image in Fig. 6, the MSEs in CT-number, between the motionless image and the motion-affected image in the region-of-interest (ROI) indicated by the dashed line in Fig. 6a is shown. The MSEs also increase as the motion angle increases. Figure 6 suggests that a small head motion parallel to the detector plane may introduce big artifacts in the CT images. Figure 7 shows the difference images between the motionless image and the motion-affected images when the step-like waveforms were applied about the y-axis, i.e., against the detector plane. The amplitudes of the step-like motion were 3° and 6° for Fig. 7a and b, respectively. With small amplitudes of

Fig. 5 **a** The phantom to observe the motion effects. **b** A CT image of the phantom. The ROI indicated by the *solid line* is for observing edge profiles

Fig. 6 Difference images between the motionless image (Fig. 5b) and the motion-affected images. The motion was about the z -axis in the coordinate of the bench-top micro-CT and the motion waveforms shown in Fig. 3 were applied. The amplitudes of the motion were **a** 0.3° , **b** 0.6° , **c** 0.9° and **d** 1.2°

0.3° , 0.6° , 0.9° and 1.2° , the motion effects were not noticeable when the motions were about the y -axis. This suggests that an object motion against the detector plane makes a much smaller change in the projection image than the motion parallel to the detector plane. A small rotation of the object against the detector plane would make little change in the projection image, particularly when the

object has a cylindrical shape. In this case too, the MSEs increase as the motion angle increases, but with a slower slope than the previous case.

To validate that the motion was accurately applied as designed, we corrected the projection data of the phantom acquired with the motion introduced in Fig. 6. With the motion waveform known a priori, we simply rotated the

Fig. 7 Difference images between the motionless image (Fig. 5b) and the motion-affected images. The motion was about the y-axis in the coordinate of the bench-top micro-CT and the motion waveforms shown in Fig. 3 were applied. The amplitudes of the motion were **a** 3° and **b** 6°

projection data in the reverse direction to correct the projection data. From the motion-corrected projection data, we reconstructed images of the phantom and we computed difference images from the motionless image. Figure 8 shows the difference images between the motionless image and the corrected images. The difference images show the noise patterns with some residual ring artifacts suggesting that all the motion effects have been corrected.

After identifying the motion effects in the coordinate of the bench-top micro-CT, we converted the step-like motions to the ones in the coordinate of a dental CT shown in Fig. 4. Figure 9 shows the difference images between the motionless image and the motion-affected images in the coordinate of a dental CT. The amplitudes of the motion were 0.3°, 0.6°, 0.9° and 1.2° in Fig. 9a, b, c and d, respectively. As expected, the motion effects are smaller than the ones in Fig. 6, but bigger than the ones in Fig. 7 in terms of edge spreading and MSE.

We corrected the motion effects in the projection data for the cases of Fig. 9. The edge profiles were computed at the ROI indicated by the rectangle in Fig. 5b. Figure 10a, b

show the edge profiles before and after the motion correction, respectively. As can be seen from Fig. 10, all the edge profiles are very close to that of the motionless image after the motion correction.

4 Discussion

With the bench-top micro-CT, head motions in a dental-CT configuration have been physically simulated by two goniometers during the scan. A physical simulation of the patient motion during the CT scan has some advantages over a computational simulation. For an exact computational simulation of patient motions, the computational model of a patient must have such a high resolution that a tiny rotation of the model in the computational space would not introduce any significant errors in computing projection data. A rotation of a computational model needs a spatial interpolation, hence, the interpolation may introduce significant errors to the projection data if the computation model has a low spatial resolution, i.e., a limited matrix

Fig. 8 Difference images between the motionless image and the motion-corrected image. The motions were the same as in Fig. 6. The amplitudes of the motion were **a** 0.3° and **b** 1.2°

Fig. 9 Difference images between the motionless image and the motion-affected images. The motions were applied with the waveforms shown in Fig. 4 with the amplitudes of **a** 0.3° , **b** 0.6° , **c** 0.9° and **d** 1.2°

Fig. 10 The edge profiles at the ROI shown in Fig. 5b when the motions were applied as in Fig. 9. **a** Before the motion correction. **b** After the motion correction

size. In addition to being free from the interpolation errors, the physical simulation is more realistic than the computational simulation in that the projection data acquired in the physical simulation are affected by many physical factors involved in the real scan, such as photon scattering, x-ray beam hardening, photon noise, and non-zero detector element size, etc.

To physically simulate the patient motion, we may use a robotic arm to hold a phantom inside a CT scanner and

control the robotic arm to apply a pre-defined motion to the phantom. However, installing a robotic arm inside a CT scanner is very difficult due to the limited space apart from the high cost of a robotic arm. On the other hand, the proposed system can simulate any pattern of motion with an angular resolution of 0.05° which is expected to be useful for high-resolution dental imaging.

Our experimental results suggest that a motion parallel to the detector plane introduce much bigger motion

artifacts than a motion against the detector plane. A rotational motion parallel to the detector plane introduces a rotation of the projection data on the 2D detector plane, thereby causing backprojection of the projection data to wrong positions in the CT image space. In a dental-CT scan in which a patient is sitting on a chair or standing, both head tilting and head nodding may occur even though the patient's head is fixed by a bandage or a fixture. We expect the bench-top micro-CT can be greatly used in developing motion artifact correction methods for a high-resolution dental-CT. With the real projection data with some motion effects, it will be difficult to evaluate the motion artifact correction methods particularly when the patient's motion is very small. If the images without any motion artifacts are available, evaluation of the corrected images can be easily made by investigating the difference between the corrected images and the artifact-free images. When it comes to the case that a tiny degradation of spatial resolution could be a great concern, such as the case of direct prototyping from 3D CT images using a 3D printer, the proposed system will have great benefits since machining or printing with a precise dimension is of utmost importance in prototyping dental devices. In our future studies, we plan to develop efficient motion artifact correction methods for a high-resolution dental-CT utilizing the bench-top micro-CT. With the flexibility in physical motion simulations, projection data can be acquired with many types of motion artifacts by the proposed system.

5 Conclusions

With the bench-top micro-CT, head tilting and nodding motions can be precisely simulated in a predetermined way to evaluate the motion effects on dental-CT images. We expect that the projection data acquired with the motion effects can be greatly used in developing motion artifact correction methods for a high-resolution dental-CT.

Acknowledgements This work was supported by the National Research Foundation of Korea (NRF-2015M2A2A7A0 3043177), the Korean government (MEST).

Compliance with ethical standards

Conflict of interest The authors have no COI to report.

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