RESEARCH ARTICLE

Investigation of an optimized scanning protocol for the dentomaxillofacial region using 320-slice multidetector computed tomography

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Objectives: To propose an imaging protocol that provides satisfactory image quality for oral examination while minimizing radiation dosage using 320-slice multidetector CT (MDCT).

Methods: An anthropomorphic head phantom was scanned using 320 MDCT with protocols combining different scanning modes: volume scanning (whole or local) vs helical scanning (80- or 64-slice detectors); tube voltage settings (80 kVp, 120 kVp and 135 kVp); and tube current settings (60 mA, 80 mA, 100 mA and 120 mA). A total of six anatomical bone structures and three anatomical soft-tissue structures were assessed using quantitative and qualitative analysis in the three orthographic planes (axial, sagittal and coronal). A figure of merit (FOM) was used to determine the optimal imaging protocol in terms of tube voltage, tube current and scanning mode.

Results: The 80-kVp setting had the worst quantitative and qualitative results (both $p < 0.001$) compared with the 135-kVp and 120-kVp settings, especially for soft-tissue structures. A significant difference was noted for the scores obtained using a tube current between 120 mA and 60 mA by quantitative analysis, but not by qualitative analysis. Volume scans using either whole or local modes had a significantly higher FOM than helical scanning of 80 or 64 slices.

Conclusions: In 320 MDCT, a protocol using 135 kVp, 80 mA and the volume-scanning mode (whole or local) offers adequate visualization of both soft-tissue and bone structures while keeping the radiation dose as low as possible. This may therefore be considered a first choice among a wide selection of scanning protocols for dentomaxillofacial CT.

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Introduction

Three-dimensional imaging is starting to play an important role in oral medicine for screening purposes,

periodontal evaluation, orthodontic treatment planning, oral surgery and implant planning.¹ This is especially the case for CBCT and multidetector CT (MDCT). There is some evidence that the panoramic images reformatted from source axial CT images offer equivalent diagnostic ability to conventional digital panoramic radiography.²⁻⁴ The development of 320 MDCT allows a scan length of up to 16 cm at the z-axis with a single rotation, which enables coverage of the

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entire maxillofacial region.^{5–8} The time required with 0.5 s for one rotation by 320 MDCT is much faster than that by CBCT (10–70 s per rotation); as a consequence, 320 MDCT has the potential for generating much smaller motion artefacts. Tube voltage and tube current greatly influence radiation exposure and are very important parameters that should be taken into account when adhering to the “as low as reasonably achievable” principles.⁹ There are no previous studies exploring the optimal protocol design in terms of selecting tube voltage and tube current settings for 320-slice MDCT in dental imaging. In order to design an optimized scanning setting, a balance between diagnostic image quality and the radiation dose should be achieved, particularly for relatively high-contrast structures. Our study aims to propose a feasible protocol that provides satisfactory image quality for diagnosis while minimizing radiation dosage.

Methods and materials

CT imaging

Our study employed a single anthropomorphic head phantom consisting of bone and soft-tissue structures at the maxillofacial area, skull and paranasal sinuses (PH-47 Dental Radiography Head Phantom; Kyoto Kagaku, Kyoto, Japan). This phantom was primarily composed of polyurethane, and the elemental composition was easily adjusted using additives to create electron densities of 1.234 g cm⁻³, 1.305 g cm⁻³, 1.502 g cm⁻³ and 1.060 g cm⁻³ for the bone crest, enamel, dentin and soft-tissue structures, respectively. All images were obtained using a 320-slice MDCT scanner (Aquilion ONE™; Toshiba Medical Systems, Otawara, Japan) with 320 × 0.5-mm detectors, tube voltage settings of 80 kVp, 120 kVp and 135 kVp and tube-current settings of 60 mA, 80 mA, 100 mA and 120 mA. This modality offered two volume-scanning modes (whole and local) for oral diagnostic examination. The whole-volume mode simultaneously scanned both the maxilla and the mandible. For the local-volume mode, the range of scanning was collimated to cover either the maxillary or the mandibular region alone. Helical scanning modes were also studied, using 80-row (80S) and 64-row (64S) detectors, with a pitch of 0.896. The 80S protocol was the default recommended by the manufacturer, while 64S represented the protocol commonly used for dental examination.^{10,11} All raw data sets were then reconstructed into images at 1-mm slice thickness, in 1-mm increments, under both bone and soft-tissue reconstruction algorithms. By using multiplanar reformation, the coronal and sagittal images were reconstructed from the axial images for all scanning protocols.

The effective dose was estimated from the CT dose index provided by the scanner. The scanner was routinely calibrated, the dosimetry was measured and the report was reviewed regularly under a standardized

quality assurance program run by the government. The dose-length product was defined as the CT dose index multiplied by the scan length and was an indicator of the integrated radiation dose of an entire examination protocol. An approximation of the effective dose was obtained by multiplying the dose-length product by a conversion factor, k (equal to 0.0019 mSv mGy⁻¹ cm⁻¹ for the head).¹²

Quantitative analysis

The images were reviewed by two board-certified radiologists using the same viewing workstation. The radiologists specialized in head and neck imaging and each had more than 8 years' experience. They were asked to grade the image quality for bone structures (delineation of outer cortical and marginal bone plates, enamel and dentin) and soft-tissue structures (inferior border of the maxillary sinus and the pulp cavity). The anatomical positions of the enamel, dentin and pulp cavity were evaluated for both the maxillary and mandibular regions. The observers determined the subjective image quality without knowledge of the settings used for tube voltage (peak kilovoltage), tube current (milliampere) or scanning mode. The images were displayed at a fixed setting: window centre of 600 and window width of 3200 for bone structures, and window centre of 50 and window width of 350 for soft-tissue structures. The anatomic structures were evaluated using a 4-point scoring system for image quality: a score of 3 refers to sharp definition and equalled excellent image quality; a score of 2 refers to minimal blurring and equalled good image quality; a score of 1 refers to blurring, but with definable margins and equalled fair image quality; and a score of 0 refers to marked blurring without definable margins and non-diagnostic (Figure 1, bone structures; Figure 2, soft-tissue structures).

We considered a score of 2 as sufficient for diagnostic purposes for both bone and soft-tissue structures.¹³ The averaged scores were calculated for each landmark at various setting parameters.

Qualitative analysis

In order to evaluate qualitative image quality, the fuzzy C-means clustering algorithm¹⁴ was performed on segmentation of bone structures (enamel and dentin) and soft-tissue structures (bilateral maxillary sinuses and pulp cavities) for each image. Signal difference-to-noise ratio (SdNR) measurements from these regions of interest (ROIs), which are considered a valuable tool in the optimization of exposure parameters,¹⁵ were calculated as:

$$\text{SdNR} = \frac{\text{mean (ROI)} - \text{mean (background)}}{\sqrt{\frac{\text{SD (ROI)}^2 + \text{SD (background)}^2}{2}}}$$

where SD indicates the standard deviation and the background was an ROI of approximately 1200 pixels surrounding the phantom.

Optimization of tube voltage, tube current and scanning mode was based on the figure of merit (FOM), which was defined as follows:

$$\text{FOM} = \frac{\text{averaged score} \times \text{SdNR}^2}{\text{effective dose}}.$$

Statistical analysis

Agreement in quantitative evaluation was analyzed using interobserver and intraobserver κ statistics computed from the 4-grade categorical scores. κ described the agreement between categorical results of paired diagnostic ratings, taking into account only agreement beyond that expected by chance.^{16,17} The equation used was:

$$\kappa = \frac{P_0 - P_C}{1 - P_C}$$

where P_0 and P_C represented the proportion of observed agreement and the proportion of agreement expected by chance, respectively.

We used the standards for κ statistic strengths proposed by Landis and Koch:¹⁸ $\kappa \leq 0$ indicated poor agreement; $0.01 \leq \kappa \leq 0.20$, slight agreement; $0.21 \leq \kappa \leq 0.40$, fair agreement; $0.41 \leq \kappa \leq 0.60$, moderate agreement; $0.61 \leq \kappa \leq 0.80$, substantial agreement; and $0.81 \leq \kappa \leq 1.00$, almost perfect agreement.

Between any arbitrary pair of parameter settings (tube voltage or tube current), two-way ANOVA and Student's *t*-test were used to assess the mean results of quantitative and qualitative analyses for both bone and soft-tissue structures. A *p*-value of <0.025 ($0.05/2$) was considered to indicate a statistically significant difference.¹⁹

Results

Quantitative analysis

The intraobserver variability of the readers reached the range of substantial agreement (weighted κ , 0.63–0.72). The interobserver variability also reached substantial agreement, with a weighted κ value of 0.67–0.83 for Reader 1 and 0.77–0.79 for Reader 2.

There was no significant difference observed between the four scanning modes for either bone or soft-tissue structures. There were significant differences in the diagnostic quality scores among all three tube voltage settings, with a *p*-value <0.0001 for both bone and soft-tissue structures. Significant differences were noted for the tube current settings (Table 1). For bone structure, the scores between 120 mA and 60 mA were statistically different for whole-volume scanning ($p = 0.021$), local-volume scanning ($p = 0.019$), 80S scanning ($p = 0.001$) and 64S scanning ($p = 0.015$); and the scores between 100 mA and 60 mA were significantly different for 80S scanning ($p = 0.012$). For soft-tissue structures, statistically

significant differences were found between all pairs of settings: 120 mA *vs* 60 mA for whole-volume scanning ($p = 0.016$), 80S scanning ($p = 0.003$) and 64S scanning ($p = 0.001$); 120 mA *vs* 80 mA for 80S scanning ($p = 0.022$); and 100 mA *vs* 60 mA for 64S scanning ($p = 0.008$).

Qualitative analysis

For tube current settings, there was no significant difference in SdNR at bone structure or soft-tissue structure regions. The tube voltage setting of 80 kVp had the worst qualitative results compared with 135 kVp or 120 kVp at the soft-tissue structures (all $p < 0.001$). On qualitative analysis, significant differences were observed under different scanning modes: for bone structures, 80S *vs* whole-volume mode ($p = 0.0005$) and 80S *vs* local-volume mode ($p = 0.0002$); 64S *vs* whole-volume mode ($p = 0.007$) and 64S *vs* local-volume mode ($p = 0.002$). For soft-tissue structures, 80S *vs* local-volume mode was statistically different ($p = 0.003$). This indicates that either the 80S or 64S mode with a pitch of 0.896 has the potential capability to improve image quality; however, radiation doses (range, 112–1045 μSv) were approximately two times higher than those generated from using the whole- or local-volume modes (61–501 μSv).

For both bone and soft-tissue structures, the FOM values under different tube voltage settings were not significantly different (Figure 3). The FOM values using 60 mA or 80 mA for bone structures were significantly higher than those using 100 mA or 120 mA (all $p < 0.005$); for soft-tissue structures, 60 mA showed relatively better performance than 120 mA ($p = 0.02$). Under both helical modes, the FOM performance was inferior to that seen under either whole-volume or local-volume modes ($p < 0.001$) for both bone and soft-tissue structures.

Discussion

There are several published studies that compare the accuracy and radiation dosage between MDCT and CBCT examination; however, some report the results of head-to-head comparisons between a single technology or apparatus and another with respect to dosimetry or image quality in a laboratory environment, without further reference to the clinical utility of the resulting data. In addition, most studies use only quantitative information to assess image quality, omitting an objective qualitative evaluation. Our study appears to be among the first to successfully examine the optimal protocols in a clinical setting for 320 MDCT, incorporating radiation doses with quantitative and qualitative assessments.

In our study, subjective visualization suggested that bone crest delineations have a minimal effect on image quality, with scores ranging from 1.75 to 2.30 (fair to good), consistent with the results of a previous study by

Figure 1 Quantitative interpretation as (a) Score 0, (b) Score 1, (c) Score 2 and (d) Score 3 for bone structures (enamel regions).

Koizumi et al.²⁰ The visualization of dentin or enamel was worst (Table 2), indicating that they are inherently low-contrast structures (the difference between dentin and enamel regions was <200 HU)²¹ and susceptible to changes in exposure setting, especially for tube voltage settings. The visualization of mandibular structures was slightly better than that of maxillary structures, similar to the results presented by Pittayapat et al.²² This might be because most mandibular structures are well corticated. Presumably, the maxilla contains more bone mass or much denser bone than the maxilla; therefore, even if the contrast and sharpness of the images is limited, the landmarks are clear enough for clinical diagnosis.

Brem et al.¹³ report that the settings of 67 mAs and 134 mAs with a fixed tube voltage of 120 kVp could provide sufficient diagnostic quality at a minimum radiation dose for both bone structures and soft-tissue

structures. The study by Marmolya et al.²³ suggests that a tube current as little as 23 mAs is sufficient for diagnostic quality when only sinusitis is of concern. Hagtvedt et al.²⁴ conclude that a setting of 40 mAs enables diagnosis of sinusitis but is unsatisfactory if detailed information on the paranasal sinuses is required, and Sohaib et al.²⁵ show that 50 mAs is sufficient for bone structures. Our findings describe tube current settings of 60 mA (equivalent to 30 mAs for volume mode and to 34 mAs for helical mode) with significantly worse results for the diagnosis of bone and soft-tissue structures compared with 120 mA (equivalent to 60 mAs for volume mode and to 67 mAs for helical mode) in quantitative evaluation. However, our qualitative results disagree: 60 mA fares better than 120 mA in the tests of FOM performances for both bone and soft-tissue structures, when radiation doses are taken into account. In addition, the higher tube voltage settings in

Figure 2 An example of quantitative interpretation for soft-tissue structures (asterisks) surrounding the maxillary sinus (white triangles): each image examined in this study can be scored from 1–3: (a) Score 1 (blurring, but with definable margins, fair image quality), (b) Score 2 (minimal blurring of soft-tissue structures, good image quality) or (c) Score 3 (sharp delineation of soft tissues, excellent image quality).

Table 1 Significant differences were presented for different tube current settings in either bone or soft-tissue structures

Scan mode	120 mA vs 60 mA	120 mA vs 80 mA	100 mA vs 60 mA
Bone structures			
Whole volume	0.021		
Local volume	0.019		
80S	0.001		0.012
64S	0.015		
Soft-tissue structures			
Whole volume	0.016		
Local volume			
80S	0.003	0.022	
64S	0.001		0.008

Data indicated as *p*-values.

our study were found to be directly related to better image interpretation capability. We therefore suggest that a protocol using 135 kVp, 80 mA and volume-scanning mode (whole or local) safely enables adequate visualization of both soft-tissue and bone structures using 320 MDCT.

The typical effective dose of a head CT is about 2 mSv.²⁶ Based on the International Commission of Radiation Protection (ICRP) Publication 60, the effective dose in MDCT of maxillofacial bone studies is 453 μ Sv at 120 kVp and 90 mA,¹¹ and in implant studies it is 596 μ Sv at 120 kVp and 100 mA.²⁷ Suomalainen *et al*²⁸ found that 64 MDCT acquires the highest dose, at 742 μ Sv, while CBCT yields a dose of 289 μ Sv. When using ICRP Publication 103, tissue-weighting factors increase the effective dose by a factor of 1.9–2.6 compared with ICRP Publication 60. Rottke *et al*²⁹ performed radiation dose measurements with three different CBCT devices (3D Accuitomo FP[®], ProMax 3D[®] and Master3Ds[®]) and revealed that the typical radiation dose for a standard acquisition protocol (upper and lower jaw) had to be located between 314 μ Sv and 396 μ Sv. The radiation dose measured in the optimized protocol was 301–334 μ Sv in our study, which is potentially comparable with or lower than the radiation dose used in CBCT devices. Compared with MDCT, measured radiation dose of our study is also apparently lower than the original dose used in regular scanning protocols but is still slightly higher than the results

described by Koizumi *et al*²⁰ (stating that the radiation dose could be reduced to 298 μ Sv), who demonstrate that a radiation dose as low as 28 mAs could be used for pre-surgical implant planning. Our study covered regions such as the enamel, dentin and pulp cavity for potential periapical diagnosis, as mentioned in Table 2. On quantitative evaluation, we found the lowest scores were obtained in the regions of the enamel and dentin; so, the tube current can potentially be adjusted to 60 mA for implant planning and the radiation dose would accordingly decrease to 226 μ Sv.

Some researchers have investigated the image artefacts generated by dental implants on CBCT and MDCT. The results show fewer artefacts with MDCT than with CBCT; the images of the implant on MDCT are correctly produced in all axial and coronal cross-sections when compared with the main implant. Reportedly, 16% of implants on MDCT generate artefacts, while >75% of CBCT images have artefacts. Moreover, there are more artefacts in the canine area than in the molar area because of the position of the canine in the arch that makes it vulnerable to generating artefacts.^{30,31} On both CBCT and MDCT, the artefacts can be reduced when the peak kilovoltage is increased; however, no reduction in artefacts is found with increased tube currents. Fortunately, the recommended protocol suggested by our study uses 135 kVp and 80 mA, a setting that is able to provide adequate image capability while keeping the radiation at a lower level.

Figure 3 Figure of merit (FOM) performance at different tube voltages and tube currents in the four scanning modes for (a) bone structures and (b) soft-tissue structures.

Table 2 Quantitative scores were averaged by two readers at upper (maxillary) and lower (mandible) jaw for volume and helical scan modes, respectively

Quantitative scores		Volume mode		Helical mode	
		Whole	Local	80S	64S
Maxillary	Maxillary sinus	1.47 ± 0.50	1.58 ± 0.55	1.69 ± 0.70	1.69 ± 0.66
	Outer cortical bone crest	1.86 ± 0.63	1.91 ± 0.60	2.00 ± 0.75	1.75 ± 0.64
	Enamel	1.19 ± 1.06	1.55 ± 1.27	1.38 ± 1.17	1.36 ± 1.17
	Dentin	1.27 ± 1.08	1.30 ± 1.19	1.30 ± 1.16	1.08 ± 0.99
	Pulp cavity	1.86 ± 0.79	1.88 ± 0.85	2.00 ± 0.82	1.83 ± 0.87
Mandible	Marginal bone crest	2.30 ± 0.46	2.16 ± 0.50	2.13 ± 0.42	2.16 ± 0.48
	Enamel	1.41 ± 1.18	1.33 ± 1.09	1.02 ± 0.94	1.27 ± 1.16
	Dentin	1.11 ± 1.00	1.02 ± 0.94	1.00 ± 0.95	0.88 ± 0.82
	Pulp cavity	1.80 ± 0.82	1.72 ± 0.84	1.41 ± 0.64	1.52 ± 0.69

This protocol might potentially, if successfully conducted, directly reduce beam-hardening artefacts.

There are some limitations to our study. Mainly, the anthropomorphic phantom used in this study lacks information on the mandibular canal and mental foramen, where important nervous structures pass through. Such structures are of great clinical value, for they are vulnerable and easily injured during clinical procedures such as implant placement surgery or osteotomy.³² In addition, the presence of anatomical variations such as bone density and thickness in humans should be considered, since the

use of MDCT is increasing in popularity. More evidence is required to validate the feasibility of our proposed optimized protocol in terms of image quality and diagnostic value; larger pre-clinical studies and human subject studies are warranted.

In conclusion, the tube voltage, rather than the tube current, is the key determinant of radiation dose in the dentomaxillofacial region during 320 MDCT. Fine tuning of the tube voltage followed by the tube current in our proposed parameters (135 kVp and 80 mA) provides sufficient reduction of the radiation dose without much loss of image quality.

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