RESEARCH

Optimization of exposure parameters for cone beam computed tomography sialography

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Objectives: The assessment of image quality is a crucial step in the development of a new imaging protocol. Having proposed and reported on a preliminary protocol for sialography using cone beam CT (CBCT), the purpose of this study was to further optimize this protocol by maximizing the image signal difference-to-noise ratio (SDNR) and to relate these new data to previously published dosimetric data for CBCT sialography.

Methods: An imaging phantom was constructed using samples with different concentrations of iodine and a water-immersed mandible. The CB MercuRay (Hitachi Medical Systems, Tokyo, Japan) was used to image the phantom using different peak kilovoltage (kVp) and milliamperage (mA) settings. SDNR was then calculated using the raw images based on mean pixel values (MPV) measured in selected regions of interest (ROI). Finally, a figure of merit (FOM) was calculated to examine the trade-off between image SDNR and effective radiation dose.

Results: The SDNR demonstrated an expected increase as the kVp increased from 60 to 120. Also, images made with the higher mA setting (15) had greater SDNR. The iodine concentration also influenced the image quality such that SDNR increased with increased amounts of iodine. The calculated FOM was greatest for the technique using 80 kVp, with equivalent results for 10 mA and 15 mA.

Conclusion: An optimized protocol for CBCT sialography using CB MercuRay entails a 6 inch field of view with 80 kVp and 10 mA.

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Introduction

In a previous study we demonstrated the wide range of radiation doses that can be delivered to the patient from cone beam CT (CBCT) scans simply by varying three exposure parameters: field of view (FOV), peak kilovoltage (kVp) and milliamperage (mA).¹ Based on the results of the dosimetry study, we proposed a preliminary protocol for sialography using CBCT. The recommended protocol entailed centring the gland of interest in a 6 inch FOV and using 80 kVp with 10 mA.¹ However, our previous study was based primarily on minimization of dose and did not consider the effect of the variation of exposure parameters on image quality. The purpose of the present study was to further optimize this protocol by maximizing the image

quality while delivering radiation doses to the patient that are as low as reasonably achievable.

Radiographic image quality is defined as the amount of information within the image that allows the radiologist to make a diagnostic decision with a particular level of certainty.² It is therefore believed that increasing the image quality increases the certainty of the diagnostic decision; most often this can be achieved simply by increasing the exposure technique such as choosing greater kVp or mA settings. However, higher exposure techniques come at the expense of greater radiation dose to the patient.³ Therefore, it is crucial when developing a new imaging protocol to consider both radiation dose and image quality and balance the trade-off between them to produce images that are diagnostically adequate while minimizing radiation exposure to the patient.

Recently there has been interest in addressing the issue of image quality for CBCT. In an *in vitro* study, Liang et al compared the image quality and visibility of anatomical structures in a dry mandible between five CBCT units

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and one multislice CT (MSCT) system using a subjective image classification system.⁴ 13 different scan protocols were used and 5 independent dentists were asked to rate the visibility of 11 anatomic structures of the mandible on a 5 point scale.⁴ Although they found great variability among the different CBCT systems, they concluded that the image quality was comparable with MSCT even though the MSCT had less image noise.⁴ This study did not investigate the trade-off between image quality and patient dose and no recommendations were made for a specific CBCT protocol.

Similarly, in 2008 Kwong *et al* subjectively evaluated the quality of images of cadaver heads and dry skulls made with the CB MercuRay (Hitachi Medical Systems, Tokyo, Japan) taken at different exposure settings.⁵ 3 FOVs were examined (12 inch, 9 inch and 6 inch) in addition to 2 kVp settings (100 and 120) and 4 mA settings (2, 5, 10 and 15). They also examined the effect of filtering the X-ray beam with copper.⁵ Image quality was defined as “the ability of the image to answer our [the authors’] diagnostic question”; this was determined by having 30 observers from various dental disciplines fill out a diagnostic quality questionnaire.⁵ For the 6 inch FOV images we are interested in, Kwong *et al*’s study showed that the image quality was affected by altering the kVp and mA setting but not by adding a copper filter.⁵ Images were considered to be of superior diagnostic quality when greater kVp and mA settings were chosen.⁵ Although the subjective findings for the 6 inch FOV images are consistent with our objective findings, once again the issue of patient radiation dose was not addressed.

In this study we are interested in maximizing the detection of the difference between the signal from an iodinated contrast agent injected into a salivary gland duct and the signal from surrounding normal tissue. This signal detection task can be quantified by the metric signal difference-to-noise ratio (SDNR), which describes the relative magnitude of useful information to that of the noise that impairs its detection.

A common method employed to optimize a radiographic imaging technique is to maximize the ratio of image SDNR to the radiation dose to the patient while maintaining some required level of SDNR.⁶ We used this approach in the present study to optimize CBCT sialography based on the results of an exploration of how SDNR varies with kVp and mA. We also report on an evaluation of the sensitivity of image signals to the iodine concentration in the contrast agent.

Materials and methods

An imaging phantom was constructed using a round (20 cm diameter) water-filled Plexiglas container. A dry, human mandible was immersed alongside four iodine contrast phantoms containing variable concentrations of iodine within this container. The iodine phantoms (1 × 1 × 4 cm) are solid radiographic test objects composed of iodobenzene embedded in epoxy resin.⁷ When

exposed to the X-ray energies used in this study, the epoxy resin has an X-ray attenuation that is the same as water and this was confirmed by imaging a phantom containing only epoxy (0 mg ml⁻¹ iodine) immersed in water for each imaging technique. The iodine phantoms imaged with the mandible contained concentrations of 180, 160, 140 and 120 mg ml⁻¹ iodine. Figure 1a is a cropped photograph of the imaging phantom and Figure 1b is the corresponding axial CBCT image of the same area of the phantom acquired at 80 kVp and 10 mA.

The CB MercuRay was used to image the phantom. All the images were acquired in a 6 inch FOV but the kVp and mA settings were varied in the technique combinations outlined in Table 1. Sample images produced by different technical parameters are displayed in Figure 2. For each technique combination, three scans were made on three separate occasions to account for CBCT system variability. Then the raw unprocessed DICOM (digital imaging and communications in medicine) images were uploaded into ImageJ (National Institutes of Health, Bethesda, MD) for analysis. Using the same axial slice from each scan, equal sized regions of interest (ROIs), 3.5 mm in diameter, were chosen in the centre of each iodine phantom and in an adjacent area that contained only background water while avoiding areas with image artefacts as demonstrated in Figure 3.

The signal difference (SD) was calculated by subtracting the mean pixel value (MPV) recorded in an adjacent ROI with only background water (ROI_W) from that recorded in an iodine phantom (ROI_I).⁶

$$SD = MPV \text{ in ROI}_I - MPV \text{ in adjacent ROI}_W \quad (1)$$

Noise for each iodine sample measurement was defined as the standard deviation (σ) of the pixel values recorded in the adjacent ROI_W.⁶

$$\text{Noise} = \sigma(\text{ROI}_W) \quad (2)$$

Then the pixel SDNR (SDNR_p) was calculated as the ratio of Equation 1 to Equation 2.⁶

$$SDNR_p = SD/\text{noise}$$

Finally a figure of merit (FOM), which is a numerical expression representing the efficiency of a given system or procedure, was calculated to relate the image quality to the radiation dose risk.⁸ It was computed as the SDNR_p² divided by the effective radiation doses (*E*) for the parotid. The effective doses were either calculated in our previous dosimetry study or calculated in the same manner for the technique combinations used in this study and are listed in Table 2.¹ The FOM values are reported in inverse microsieverts (μSv)⁻¹.

$$FOM = SDNR_p^2/E$$

The reason the SDNR was squared in the calculation of FOM is because the SDNR is proportional to the

Figure 1 Photograph (a) and axial cone beam CT image (b) of the imaging phantom showing the immersed mandible and the iodine cubes arranged adjacent to one another. In the photograph the mandible is seen resting on an object with wells that functioned to support the mandible and hold the iodine cubes in place. The iodine cubes are solid radiographic test objects composed of iodobenzene embedded in epoxy resin. They contained iodine concentrations of 180 mg ml⁻¹, 160 mg ml⁻¹, 140 mg ml⁻¹ and 120 mg ml⁻¹. The image was acquired using 80 kVp and 10 mA and edge artefacts are noted along the edges of the square-shaped iodine phantoms

square root of the dose, or in other words the SDNR squared is proportional to the dose.

Statistical analysis was performed using the SAS software version 9.1 (SAS Institute Inc., Cary, NC). The paired *t*-test was used to compare the SDNR and FOM for all the combinations of exposure techniques. The *P*-value was set at 0.05.

Results

The SDNR_p was calculated for the iodine concentration routinely used in our clinic (180 mg ml⁻¹) and was found to range between 2.3 for the technique using 60 kVp and 10 mA and 13.3 for the highest exposure using 120 kVp and 15 mA. By applying the paired *t*-test, a statistically significant difference was noted between the images that were made with 60 kVp and the images that were made with 80 kVp, 100 kVp and 120 kVp. This difference was noted regardless of the mA setting. In addition, a statistically significant difference in SDNR_p was found between the images generated with 10 mA and 15 mA, but this was only true for the 60 kVp and 120 kVp techniques. These results are outlined in Figures 4 and 5.

Table 1 The different combinations of operational parameters that were used for the cone beam CT scans

	10 mA	15 mA
60 kVp	60 kVp 10 mA	60 kVp 15 mA
80 kVp	80 kVp 10 mA	80 kVp 15 mA
100 kVp	100 kVp 10 mA	100 kVp 15 mA
120 kVp	120 kVp 10 mA	120 kVp 15 mA

The effect of the iodine concentration on the image SDNR_p is demonstrated in Figure 5. The SDNR_p ranged from 5.8 for the 120 mg I ml⁻¹ concentration to 9.0 for the 180 mg I ml⁻¹ concentration when 80 kVp was chosen with 10 mA. No statistically significant difference was noted among the concentrations when the paired *t*-test was applied.

The calculated FOM for the different technique combinations are plotted in Figure 6. SDNR_p was used to calculate the FOM. The technique with the greatest FOM value is the one that employs 80 kVp and 15 mA.

Discussion

Production of diagnostic images involves a complex interplay of many different parameters. A solid understanding of these parameters, how they interact and how they influence both image quality and radiation dose is crucial in the process of developing new imaging applications. Therefore, the protocol we propose for three-dimensional (3D) sialography using CBCT is based on an extensive evaluation of the parameters that can be adjusted by the operator and their effect on both radiation dose and image quality.

The calculated SDNR increased in proportion to kVp when the mA was held constant. This was expected because as the kVp increases, so does the X-ray energy, the flux emitted from the tube, the X-ray penetrability and thus the number of X-ray photons reaching the detector. With greater numbers of X-ray photons per unit area of image receptor, the noise decreases throughout the image,

Figure 2 Four example cone beam CT images made with the same 10 mA setting but different kVp settings that varied from 60 to 120. The images demonstrate the effect of changing the kVp on the perceived image noise with the image on the far right made with the highest kVp setting displaying the least noise

which in turn increases the SDNR. However, in our experimental data the positive relationship between SDNR and kVp is exaggerated by another source of image noise—spatial non-uniformity. An inverse cupping artefact was present in our images, which is sometimes known as “capping”.⁹ This artefact occurs when there is overcompensation for a cupping artefact, an image artefact caused by beam hardening. The capping was more pronounced in the images acquired at lower kVp, where the beam is less penetrating or “softer”. Therefore, the SDNR could be improved at the lower peak kVp by appropriate corrections for spatial non-uniformity (*i.e.* flat-fielding). A positive relationship was also noted between increases in mA and SDNR, where, within the observed range of SDNR variability, the SDNR increased by a factor of $\sqrt{1.5}$ when mA was increased by a factor of

1.5. This relationship indicates that quantum noise is likely to be the dominant factor affecting image performance for each technique combination considered here. These quantitative results are consistent with the trends observed in the subjective image quality ratings published by Kwong *et al.*⁵

FOM values were influenced by the choice of kVp and mA. At the lower kVp settings of 60 and 80, a direct relationship was noted between FOM and mA but at the higher kVp settings of 100 and 120, an inverse relationship was demonstrated because the dose increased significantly more than the SDNR. The only FOM values that were significantly different than the others were those at 60 kVp. This indicates that among the other examined techniques, where image quality was adequate, the one that should be adopted as a clinical protocol for sialography is the one that imparts the lowest effective radiation dose. This technique is the one that utilizes 80 kVp with 10 mA. These technique factors are optimized specifically for the CB MercuRay unit, which has a limited range of choices of kVp (60, 80, 100 and 120) and of mA (10 and 15). Further refinement of the sialography protocol could be made following the same approach used in this work for other CBCT systems with different imaging technique parameters.

Suomalainen *et al* published an extensive study that evaluated the radiation dose and image quality of four CBCT scanners and compared them with two MSCT scanners.¹⁰ Using appropriate phantoms, radiation doses and contrast values were measured.

Figure 3 Cone beam CT image showing the region of interest (ROI) in an iodine phantom (black outline) and a ROI in an adjacent area with only background water (white outline)

Table 2 The effective radiation doses that were used to calculate the figure of merit for the different technique combinations

Technique	Effective dose (μSv)
60 kVp, 10 mA	27
60 kVp, 15 mA	33
80 kVp, 10 mA	60
80 kVp, 15 mA	71
100 kVp, 10 mA	97
100 kVp, 15 mA	145
120 kVp, 10 mA	120
120 kVp, 15 mA	167

Figure 4 Variation of signal difference-to-noise ratio (SDNR_p) with different peak kilovoltage (kVp) and milliamperage (mA) settings. The error bars represent the maximum and minimum values for each individual technique. The graph highlights the statistically significant differences that were found among the various techniques (*)

Radiation doses were then calculated and reported as effective doses.¹⁰ The contrast-to-noise ratio (CNR) was used as an indicator of image quality, and the modulation transfer function (MTF) quantity was used to determine the resolving power of each system.¹⁰ The authors concluded that despite the large variation in results, CBCT scanners provide adequate image quality with smaller effective doses compared with MSCT.¹⁰

However, the authors were quick to emphasize the importance of optimizing imaging factors for the different examinations.¹⁰ It is our opinion that this recommendation can be accomplished only by optimizing these parameters for specific clinical applications of CBCT and by objectively measuring the trade-off between image quality and radiation dose as we have done in this study for CBCT sialography.

Figure 5 Variation of signal difference-to-noise ratio (SDNR_p) relative to iodine concentration in the phantom. Four concentrations are shown: 120 mg I ml⁻¹, 140 mg I ml⁻¹, 160 mg I ml⁻¹ and 180 mg I ml⁻¹; these were imaged using 80 kVp and 10 mA. The error bars represent the maximum and minimum values of SDNR_p for each concentration. A linear fit to the data is shown by the solid line

Figure 6 Calculated figure of merit (FOM) for eight different combination techniques using 60, 80, 100 and 120 kVp and two milliamperage (mA) settings (15 and 10). The error bars represent the maximum and minimum values for each individual technique

Omnipaque® (Iohexol injection 39%, General Electric Healthcare Canada Inc., Mississauga, ON) is a low osmolar, non-ionic iodinated contrast agent that is produced in 5 concentrations: 140 mg I ml⁻¹, 180 mg I ml⁻¹, 240 mg I ml⁻¹, 300 mg I ml⁻¹ and 350 mg I ml⁻¹ for different indications of use. At our institution and for the indication of sialography, the 180 mg ml⁻¹ iodine concentration is used. Therefore, in this study we examined the effect of the 2 smaller commercially available concentrations (140 mg I ml⁻¹ and 180 mg I ml⁻¹) in addition to 2 intermediate concentrations (120 mg I ml⁻¹ and 160 mg I ml⁻¹) in order to fully understand the impact of the iodine concentration on SDNR for CBCT sialography. Greater concentrations of the contrast agent were deemed unnecessary because the injection is locally confined to the ductal structures of the salivary glands as opposed to vascular injection, where the agent quickly becomes diluted by distribution throughout the whole body. An increase in SDNR was noted with the increase in iodine concentration, but there was no statistically significant difference among the different concentrations for the technique using 80 kVp and 10 mA. This relationship is owing to the substantial signal difference provided by the attenuation of iodine in the salivary ducts.¹¹

To our knowledge there are no other studies in the English-language literature that examine the effect of varying the iodine concentration in contrast agents on the quality of the resultant images for sialography studies. However, a similarity can be drawn to studies of CT angiography. Unfortunately, these studies vary in their results, from demonstrating no significant effect of the

iodine concentration¹² to showing a statistically significant difference^{13,14} on vascular enhancement. The results of these studies are complicated by many other variables, such as imaging techniques and the injection parameters.¹⁴ One particular study by Wang *et al* examined the effects of two iodine concentrations of the same contrast agent on image quality and side effects for dynamic CT of the neck.¹⁵ They found that both concentrations produced images of excellent quality but the lower concentration resulted in fewer immediate minor complications.¹⁵ In the current study, we demonstrated a positive linear trend between the iodine concentration and SDNR_p, as indicated by the linear fit to the data in Figure 5. This relationship is expected because the signal difference will increase proportionally to the amount of iodine present. However, no statistical differences in SDNR_p were noted among the 4 concentrations examined, the highest of which (180 mg I ml⁻¹) is the concentration currently used at our institution for sialography. The likelihood of complications from contrast media in relation to sialography procedures is minimal because of the localization of the contrast material in the salivary ducts; nevertheless, it would be more reasonable to use the smaller, commercially available concentration of the contrast agent, which is 140 mg I ml⁻¹.

In conclusion, we have developed an optimized protocol for CBCT sialography using the CB MercuRay unit. The protocol involves centring the salivary gland of interest in a 6 inch FOV and choosing a kVp of 80 with 10 mA. Our future goal is to examine the diagnostic efficacy of this optimized CBCT sialography technique in a prospective clinical trial where it will be compared with the standard of practice which is two-dimensional plain film sialography.

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