RESEARCH ARTICLE

The effective dose of different scanning protocols using the Sirona GALILEOS® comfort CBCT scanner

¹D Chambers, ²R Bohay, ³L Kaci, ^{3,4,5}R Barnett and ^{3,4,5}J Battista

¹Division of Graduate Orthodontics, Schulich School of Medicine & Dentistry, Western University, London, ON, Canada;

²Division of Oral Medicine and Radiology, Schulich School of Medicine & Dentistry, Western University, London, ON, Canada;

³Department of Physics and Engineering, London Regional Cancer Program, London, ON, Canada; ⁴Department of Oncology, Western University, London, ON, Canada; ⁵Department of Medical Biophysics, Western University, London, ON, Canada

Objectives: To determine the effective dose and CT dose index (CTDI) for a range of imaging protocols using the Sirona GALILEOS® Comfort CBCT scanner (Sirona Dental Systems GmbH, Bensheim, Germany).

Methods: Calibrated optically stimulated luminescence dosimeters were placed at 26 sites in the head and neck of a modified RANDO® phantom (The Phantom Laboratory, Greenwich, NY). Effective dose was calculated for 12 different scanning protocols. CTDI measurements were also performed to determine the dose-length product (DLP) and the ratio of effective dose to DLP for each scanning protocol.

Results: The effective dose for a full maxillomandibular scan at 42 mAs was $102 \pm 1 \mu\text{Sv}$ and remained unchanged with varying contrast and resolution settings. This compares with $71 \mu\text{Sv}$ for a maxillary scan and $76 \mu\text{Sv}$ for a mandibular scan with identical milliampere-seconds (mAs) at high contrast and resolution settings.

Conclusions: Changes to mAs and beam collimation have a significant influence on effective dose. Effective dose and DLP vary linearly with mAs. A collimated maxillary or mandibular scan decreases effective dose by approximately 29% and 24%, respectively, as compared with a full maxillomandibular scan. Changes to contrast and resolution settings have little influence on effective dose. This study provides data for setting individualized patient exposure protocols to minimize patient dose from ionizing radiation used for diagnostic or treatment planning tasks in dentistry.

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Introduction

Dental radiography provides essential information in diagnosis and subsequent treatment planning. While two-dimensional analogue techniques continue to be employed in routine dental practice, digital and three-

dimensional (3D) modalities are becoming increasingly common. Digital radiography offers improved compatibility with practice management and imaging software, reduces storage needs, eliminates chemical, silver and lead foil waste, facilitates transfer of images between practitioners and may offer reduced radiation exposure.¹ 3D imaging techniques provide a further means for interpreting the spatial relationship of structures otherwise superimposed on standard two-dimensional images.

Correspondence to: Dr David Chambers. E-mail: drchambers@gmail.com

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Figure 1 The Therapax HF 150 unit (Pantak, East Haven, CT) is operated at 85 kVp with aluminium-copper filter in place and a Farmer Chamber (Capintec, Ramsay, NJ) at a depth of 6 cm in a polymethyl methacrylate phantom.

This has important implications in dental implantology and other surgical procedures where detailed assessment of structures adjacent to the surgical site is essential. In clinical situations where multiple two-dimensional images are indicated, 3D techniques offer an efficient method of reconstructing the same group of images while limiting image acquisition to a convenient and consistent single exposure.²

While 3D images offer abundant clinical information, it is not sufficient to prescribe them simply on the basis of convenience. As with any imaging study that uses ionizing radiation, the underlying principles of justification and optimization of radiation levels must be applied. The benefits of the examination should exceed the risks incurred by exposure.³⁻⁵ A 3D scan obtained with maximum field of view (FOV) and resolution may provide the most complete set of imaging data for the patient. However, it is completed at the expense of increased radiation dose. Determination of the appropriate scanning protocol must be considered in the context

of the patient's history, clinical findings, previous imaging results and radiation exposures, differential diagnosis and treatment plan. The scanning protocol should also take into consideration the age and size of the patient as well as the ability of the scan to demonstrate the anatomy of greatest interest. It becomes obvious that image protocol optimization is not a simple task and prescription requires a thorough knowledge of both diagnostic imaging requirements and the radiobiological risks associated with ionizing radiation.

While the number of dosimetry studies relating to CBCT units used in dentistry continues to increase, few have reported findings specifically associated with the Sirona GALILEOS® Comfort scanner (Sirona Dental Systems GmbH, Bensheim, Germany). A review of the literature identified three such studies in which only a restricted number of protocols were investigated.⁶⁻⁸ All protocols described were of full maxillomandibular scans. No single published study was identified comparing the effective dose and CT dose index (CTDI) for a range of milliamperes-seconds (mAs), FOV, contrast and resolution settings for the Sirona GALILEOS Comfort scanner.

The purpose of this comprehensive study is to determine and compare the effective dose and CTDI associated with a representative set of the scanning protocols available. This information should assist in determining the optimal scanning protocol for each patient when combined with knowledge of respective effects on image quality. In addition, ratios of effective dose to dose-length product (ED/DLP) provide an efficient means of estimating effective dose for non-standard protocols and for comparing results with other scanner designs.^{9,10}

Methods and materials

Absorbed dose was measured at established sites using InLight® nanoDot™ optically stimulated luminescence dosimeters (OSLDs; Landauer, Glenwood, IL) placed within a single, modified, male tissue-equivalent Alderson RANDO® phantom (The Phantom Laboratory, Greenwich, NY). OSLDs were used because they provide excellent linear response, radiosensitivity and reproducibility over a wide range of operating energy, with efficient non-destructive readout allowing re-analysis.¹¹ For kilovoltage radiation, however, precautions must be taken to calibrate the OSLDs for the tissues of interest (soft tissue or bone) and the beam conditions of interest. The aluminium content of the OSLDs causes a strong energy and orientation dependence at kilovoltage energies.^{11,12}

In total, 26 OSLDs were used in the RANDO phantom throughout the study. All dosimeters were individually calibrated using a Therapax HF150 superficial radiation therapy X-ray unit (Pantak, East Haven, CT) operated at 85 kVp with a composite filter [1.1 mm of aluminium (Al) and 0.3 mm of copper]

Figure 2 Optically stimulated luminescence dosimeters are placed in the square holes positioned within a polymethyl methacrylate intersection template.

(Figure 1). This filter produced a beam quality comparable to that of the GALILEOS Comfort system in terms of the first half value layer (7.3 mm Al; effective energy, 52 keV) in polymethyl methacrylate (PMMA), commonly used as a tissue-equivalent material in this energy range.¹³ The Therapax unit allows user control over tube potential (*i.e.* kVp), mA, exposure time and source-to-object distance. Groups of OSLDs were positioned flat within the radiation field on a PMMA block 5 cm in thickness and overlaid with a second block 6 cm in thickness (Figure 1). These platforms approximated the typical depth of internal dosimeters placed within the RANDO phantom and accounted for the presence of similar levels of primary and Compton-scattered radiation. The OSLDs were exposed at fixed dose increments with the cumulative count during calibration exceeding that during phantom measurements. Each OSLD was read using a Microstar OSLD reader (Landauer), and the reading for each exposure was taken as the average of three consecutive readings.

For calibration, each exposure was also measured under identical conditions using a Farmer Chamber (0.6 cm³) and Capintec 192 Digital exposure meter (Capintec Inc., Ramsay, NJ) with calibration traceable to the Canadian national standard. The calibration procedure complied with an established protocol used in radiation therapy.¹⁴

Linear calibration curves were established to characterize each individual OSLD response in terms of dose to either soft tissue (muscle assumed) or to compact bone, using exposure-to-dose conversion factors (cGy R⁻¹) of 0.95 and 3.40, respectively.¹⁵ Note, however, that all exposures used PMMA for backscattering and thus enhanced backscatter¹⁴ from the bone¹⁶ was not accounted for in the calibration.

The head and neck segment of the RANDO phantom consists of 10 axial sections, each with pre-drilled holes designed to accommodate thermoluminescent dosimeters

(TLDs). To substitute OSLDs, five PMMA templates (Figure 2) 2.15 mm thick were fabricated and placed superior to sections labelled as 3, 4, 6, 7 and 9 (Figure 3). Square holes were established in each PMMA template to accommodate OSLDs at anatomical landmarks without allowing movement.

26 sites were measured corresponding to the 24 anatomical positions described previously by Ludlow *et al*¹⁶ (Table 1). Two sites were established for the left and right mandibular body. Owing to difficulty in approximating the centre of the body, one site was positioned superior to and another inferior to the mandibular body with the average absorbed dose chosen to represent that of the mandibular body.

Each phantom is constructed of a natural human skeleton cast inside a standard mould composed of a material radiologically simulating soft tissue. Initial images of the phantom revealed discrepancies between the soft-tissue surface contours of the head and neck and the underlying skeleton (Figure 4). Dosimeter sites for skeletal and internal soft-tissue landmarks were therefore established in relation to the skeleton, while

Figure 3 The RANDO® phantom (The Phantom Laboratory, Greenwich, NY) is reassembled with the polymethyl methacrylate templates in place.

Table 1 The list of anatomic sites for dosimeter placement

Anatomic location	OSLD number	OSLD identification
Calvarium anterior	1	34151D
Calvarium left	2	42477T
Calvarium posterior	3	42098X
Mid-brain	4	44986I
Pituitary	5	46848I
Right orbit	6	29436W
Left orbit	7	48908I
Right lens of eye	8	41606Z
Left lens of eye	9	46757L
Right cheek	10	409410
Right parotid	11	27592Y
Left parotid	12	37841X
Right ramus	13	26177Z
Left ramus	14	32758P
Centre cervical spine	15	31835W
Left back of neck	16	44131D
Right mandible body upper	17	45032C
Right mandible body lower	18	35299O
Left mandible body upper	19	32817T
Left mandible body lower	20	48629K
Right submandibular gland	21	31657U
Left submandibular gland	22	32871Z
Centre sublingual gland	23	28791V
Midline thyroid	24	46970B
Thyroid surface—left	25	294126
Oesophagus	26	35015A

OSLD, optically stimulated luminescence dosimeter.

surface landmarks were established for soft-tissue sites. Surface landmarks were marked with tape over which OSLDs were centred for each protocol.

With the PMMA templates in place, lead foil from intraoral radiograph packets was placed in each internal OSLD location and a 3D scan completed. The reconstructed image was examined to confirm acceptable location of each internal landmark (Figure 5).

The RANDO phantom was oriented with the occlusal plane parallel to the scan plane, the outline of the soft-tissue chin and nose visible and both inferior and posterior borders of the mandible superimposed. After an acceptable phantom set-up was obtained, the volume control head mount supplied for the Sirona GALILEOS Comfort was modified with three wires extending from custom acrylic attachments (Figure 6). The tips of each wire corresponded to surface fiducial points on the phantom marked with tape. This allowed for reproducibility of the set-up between protocols. The image from each protocol was examined and deemed acceptable if the “tilt” and “rotation” settings required to establish optimal superimposition of the right and left borders of the mandible did not exceed ± 1 unit of correction (Figure 7).

The Sirona GALILEOS Comfort CBCT tube voltage is fixed at 85 kV, with a current of 7 mA (reduced to 5 mA for 10 mAs setting) and number of basis projection images set at 200. The user can establish multiple protocols only by varying the mAs, FOV and using high-contrast (HC) or high-resolution (VO1) modes (voxel size, 0.15 or 0.30 mm). The mAs options include

10, 14, 21, 28, 35 and 42 mAs. The full FOV is a spherical volume 15 cm in diameter, while the “upper jaw” and “lower jaw” settings collimate the beam to an approximate height of 8.5 cm and therefore a reduced volume of approximately $8.5 \times 15 \times 15$ cm.¹⁷

With 3 FOV, 6 mAs and 2 contrast and resolution settings available, a total permutation of 72 protocols is possible with the Sirona Galileos Comfort CBCT scanner. Of these 72 protocols, a total of 12 were investigated, chosen as a representative sample of those available. The protocols selected are summarised in Table 2. Protocol 3 (full FOV, 42 mAs; VO1HC) was repeated on three separate occasions to investigate experimental variability in the radiation dosimetry results.

Each scanning protocol was repeated three times prior to removing the OSLDs from the phantom to minimize the contribution of noise. Integrated absorbed dose was thus obtained by dividing the measured dose by three. A delay of at least 15 min was used between scanning and reading of the dosimeters as suggested by the manufacturer. This allows low-level, non-dosimetric electron traps to stabilize.¹¹

The dosimeters were read using a Microstar OSLD reader taking the mean value of three consecutive readings. The standard deviation in readings was used to estimate uncertainty in effective dose through error propagation (Table 3). Dosimeters were annealed using a high-intensity light source after no more than three subsequent protocol exposures to minimize uncertainty associated with excess dose accumulation.

Dosimeter counts were entered into a spreadsheet and the absorbed dose determined by applying individual

Figure 4 The relationship between hard and soft tissues in the RANDO® phantom (The Phantom Laboratory, Greenwich, NY) are illustrated by an image reconstructed from a three-dimensional scan using the Sirona GALILEOS® Comfort scanner (Sirona Dental Systems GmbH, Bensheim, Germany).

Figure 5 Internal optically stimulated luminescence dosimeter locations are marked with lead foil and a three-dimensional scan is completed to verify acceptable positioning.

calibration curves based on the ion chamber calibration measurements (Figure 1). Effective dose (E) was then calculated using the mean absorbed dose (H_T) to the individual tissues and organs, applying the corresponding 2007 International Commission on Radiological Protection (ICRP)¹⁸ tissue weighting factors (W_T) and correcting for the fractional volume of the tissue irradiated (f_T). The fraction of tissue irradiated was determined by following that outlined by Ludlow and Ivanovic.⁷ Each effective dose was therefore calculated using the following equation:

$$\text{Effective Dose } (E) = \sum f_T \times W_T \times H_T$$

Table 4 lists the dosimeters used to sample each tissue, respective fraction of tissue irradiated and tissue weighting factors following that described by Ludlow *et al.*¹⁶ For those tissues located outside of the head and neck region, absorbed dose was assumed to be negligible.

In addition, the volume CT dose index (CTDI_{vol}) was measured for each protocol using a PMMA head phantom (Capintec Inc.) 16 cm in diameter and the Farmer Chamber and Capintec 192 digital exposure meter described previously. With a cone beam geometry, we opted to use the small ion chamber rather than a long pencil chamber typically used for fan beam CT scanners. Pencil chambers can be underexposed at their extremities, as is the case for maxillary and mandibular protocols. This clipping must then be accounted for in interpreting the ionization signal.²⁰

Prior to each protocol, room temperature and pressure were measured using a Traceable Digital Workstation Barometer (VWR International, Mississauga, ON) to correct the ionization chamber readings. The chamber was placed into central and peripheral holes of the phantom. The DLP for each protocol was determined by multiplying the CTDI_{vol} value by the superior-inferior field length (8.5 or 15 cm) and adjusting for the different mAs settings linearly. Ratios of effective dose to DLP were also calculated for each protocol (Table 3) as described by Huda *et al.*^{9,10}

Results

A summary of the effective dose associated with each protocol is presented in Table 3. The effective dose ranged from 23 μSv for a maxillary scan at 14 mAs to 103 μSv for a full maxillomandibular scan at 42 mAs. This is equivalent to between 3 and 12 days natural background radiation (8.5 μSv per day) per capita¹⁹ and significantly less than the effective dose for a typical

Figure 6 The volume control head mount is modified to reproduce the set-up between protocols.

Figure 7 The “tilt” and “rotation” settings are examined after superimposing the right (R) and left (L) borders of the mandible to confirm acceptable set-up reproduction.

conventional diagnostic CT scan of the head with reports ranging between 429.7 and 1410 μ Sv.^{21–23}

The full maxillomandibular scan at 42 mAs, HC and VO1 was repeated on three separate occasions (Protocol 3). The effective dose calculated for each was 102, 103 and 103 μ Sv. Minimal variability (<1%) in effective dose was associated with dosimeter placement and phantom orientation between protocols. The ED/DLP ratios for full scans are well within the range reported for diagnostic CT scanners,¹⁰ as might be expected since a similar range of anatomy is scanned.

Protocols 3–6 represent the effective dose for protocols where contrast and resolution were changed while FOV and mAs were kept at full maxillomandibular and 42, respectively. Effective dose was within 2 μ Sv for these protocols suggesting that changes in contrast and/or resolution appear related only to software and have little, if any, impact on the effective dose.

Collimating for a maxillary or mandibular scan resulted in a reduced effective dose relative to a full maxillomandibular scan. The effective dose associated with a full maxillomandibular scan is reduced on average by 29% when collimated for a maxillary scan and 24% when collimated for a mandibular scan. The ED/DLP ratios for partial volume scans are higher than those for full scans as might be expected with a higher effective dose owing to exposure of organs with a greater ICRP weighting factor. This may also be owing to the use of a fixed fraction of irradiated tissue for all protocols whether full or partial (Table 4).

Figure 8 depicts the changes in effective dose with increasing mAs for maxillary, mandibular and full FOVs. As expected, an increase in mAs results in a linear increase in effective dose for all FOVs. All scans were at high resolution and contrast. Both maxillary and

mandibular scans show a similar effective dose reduction relative to a full scan throughout the range of mAs settings.

The mean absorbed dose to individual organs and tissues is summarised in Table 5. Table 6 compares the individual organ/tissue absorbed doses for maxillary scans as a percentage of the doses for mandibular scans, taking the mean of that calculated for 14, 28 and 42 mAs. Although the effective doses for maxillary and mandibular scans are similar, the distribution of absorbed doses to the tissues differs. For a maxillary scan, the absorbed dose to the brain is on average 298% of that for a mandibular scan and, similarly, 146% for the skin and 140% for the bone marrow.

Discussion

The effective dose calculated for the maximum exposure (maxillomandibular scan, 42 mAs, HC, VO1) averaged 103 μ Sv, with 1% precision. This compares with previous findings of Ludlow and Ivanovic⁷ who reported 128 μ Sv, while Rottke et al⁶ tabulated 95 μ Sv for the “highest exposition” parameters. The effective dose calculated for a maxillomandibular scan at 28 mAs (HC, VO1) in this study was 68 μ Sv compared with 84 μ Sv reported by Pauwels et al.⁸ All reports are based on 2007 ICRP¹⁸ tissue weighting factors.

The discrepancies across these studies are likely due in part to differences in the dosimeter calibration protocols. During this study, calibration curves for each OSLD were initially investigated by two methods. The final method chosen simulated scattering conditions of internally positioned dosimeters, whereby OSLDs were placed between two PMMA blocks (in PMMA). The other method used a calibration with dosimeters placed on the surface of a 14-cm thick block of PMMA (in air). Greater response of the OSLDs (*i.e.* readings increased by approximately 30%) was observed when calibrated at a depth in a phantom *vs* a surface exposure. We speculate that this enhanced response is owing to changes in overall beam energy spectrum and angular incidence caused by Compton scattering in-PMMA at depth. A

Table 2 The list of protocols investigated

Protocol number	Field of view	mAs	Contrast	Resolution
1	Full	14	High	VO1
2	Full	28	High	VO1
3	Full	42	High	VO1
4	Full	42	Normal	VO1
5	Full	42	High	VO2
6	Full	42	Normal	VO2
7	Maxillary	14	High	VO1
8	Maxillary	28	High	VO1
9	Maxillary	42	High	VO1
10	Mandibular	14	High	VO1
11	Mandibular	28	High	VO1
12	Mandibular	42	High	VO1

mAs, milliamperere-seconds; VO1, high resolution setting; VO2, standard resolution setting.

Table 3 The effective dose and ratio to dose–length product (DLP) is determined for each scanning protocol

Protocol	Field of view	mAs	Contrast	Resolution	Effective dose ^a (μSv)	Standard deviation (μSv)	Equivalent days background ^b	DLP ($\text{mGy} \cdot \text{cm}^{-1}$)	ED/DLP [$\mu\text{Sv} (\text{mGy cm})^{-1}$]
1	Full	14	High	High	30.77	0.13	4	16.77	1.83
2	Full	28	High	High	67.52	0.30	8	33.55	2.01
3	Full	42	High	High	103.37	0.35	12	50.32	2.05
3	Full	42	High	High	101.89	0.64	12	50.32	2.02
3	Full	42	High	High	102.70	1.24	12	50.32	2.04
4	Full	42	High	Standard	103.29	0.32	12	45.80	2.26
5	Full	42	Normal	High	102.94	0.87	12	50.32	2.05
6	Full	42	Normal	Standard	101.66	0.10	12	50.32	2.02
7	Maxillary	14	High	High	22.48	0.04	3	8.66	2.60
8	Maxillary	28	High	High	47.92	0.11	6	17.31	2.77
9	Maxillary	42	High	High	70.50	0.45	8	25.97	2.71
10	Mandibular	14	High	High	23.80	0.13	3	8.66	2.75
11	Mandibular	28	High	High	51.55	0.24	6	17.31	2.98
12	Mandibular	42	High	High	76.17	0.40	9	25.97	2.93

mAs, milliamperes-seconds.

^a2007 International Commission on Radiological Protection¹⁸ tissue weighting factors.^bThis assumes an average of 3100 μSv per year.¹⁹

6-cm PMMA layer between the source and dosimeters was therefore chosen for our calibration, because it models the beam “seen” by the majority of dosimeters positioned within the RANDO phantom. Calculations using “free in air” calibration curves instead of “in PMMA” calibration resulted in a higher effective dose of 132 μSv for a full scan at 42 mAs.

Ludlow and Ivanovic⁷ reported the use of TLDs that were pre-calibrated and analysed by the supplier Lindauer (Glenwood, IL). It is unclear under what conditions that the respective calibration curves were established. Pauwels *et al*⁸ reported the use of two types of TLDs, one type calibrated in air and the other by internal calibration dosimeters for each experiment. Rottke *et al*⁶ calibrated each TLD to obtain individual

calibration factors. It is again unclear whether calibration was completed free in air or under different “in phantom” conditions. For OSLDs, it is especially important to reproduce the expected conditions of exposures including a match to the initial beam energy and quality (kVp, half value layer) and beam hardening/softening occurring within tissue. This emphasizes the need to specify calibration procedures much more explicitly in dental CT dosimetry.

Another possible calibration issue when using “flat”-shaped dosimeters is that their response could vary with edge-on orientation of the beam during CT exposures. OSLDs are more dependent on incident beam angle than are TLDs.¹² While this effect is readily observed with a fixed beam, it is less pronounced in CT exposure

Table 4 The percentage of tissue irradiated, tissue weighting factor and optically stimulated luminescence dosimeter (OSLD) samples for each tissue/organ is summarised

Tissue/organ	Identification of OSLDs used to sample tissue	Fraction of tissue irradiated (%)	Tissue weighting factor (W_T)
Red bone marrow		16.5	0.12
Mandible	13, 14, 17–20	1.3	
Calvaria	1, 2, 3	11.8	
Cervical spine	15	3.4	
Oesophagus	26	10	0.04
Thyroid	24, 25	100	0.04
Skin	8, 9, 10, 16	5	0.01
Bone surface		16.5	0.01
Mandible	13, 14, 17–20	1.3	
Calvaria	1, 2, 3	11.8	
Cervical spine	15	3.4	
Brain	4, 5	100	0.01
Salivary glands	11, 17–23, 12, 21, 22, 23	100	0.01
Remainder			0.12
Lymphatic nodes	11–15, 17–24, 26	6	
Muscle	11–15, 17–24, 26	6	
Extrathoracic airway	6, 7, 11–15, 17–24, 26	100	
Oral mucosa	11–14, 17–23	100	
Gonads		0	0.08
Colon		0	0.12
Lung		0	0.12
Stomach		0	0.12
Bladder		0	0.04
Breast		0	0.12
Liver		0	0.04

Figure 8 The effective dose is determined for each field of view at 14, 28 and 42 mAs. mAs, milliamperere-seconds.

with a range of incident angles, further amplified by internal Compton scattering. Al-Senan and Hatab¹² report an effect of <10% for a wide angular range of CT exposures at 80 kVp. If we assume that all of these angles occur internally in the RANDO phantom, this uncertainty would only account for a variation of 5% relative to the calibration exposure of OSLDs laid flat.

This study aimed to identify the effect of protocol parameters on effective dose. A reproducible set-up is essential so that valid comparisons can be made between different protocols. Once an acceptable position of the phantom was obtained, a modified volume control head mount was used to reproduce positioning based on three surface fiducial points. After each protocol, the “tilt” and “rotation” settings were modified to obtain optimal superimposition of the posterior and inferior borders of the mandible. A set-up was deemed adequately

reproduced if these settings did not deviate more than ± 1 unit of tilt or rotation correction unit. This range was established arbitrarily prior to the study. It is unclear if the tilt and rotation settings correspond to units of angular degrees, but it was observed that modifying the tilt or rotation ± 1 corresponded to relatively little movement of the image. Using the modified head mount proved successful with every protocol adequately reproducing the original. To investigate the reproducibility of effective dose measurement, a maxillomandibular protocol with maximum exposure (42 mAs, HC, VO1) was repeated on three separate occasions. The effective dose was within 1% (102, 103 and 103 μSv).

TLDs have been used with success for many decades and account for the majority of dosimeters used in organ dose studies in dentistry. As a result, most phantoms in use are prepared with pre-drilled holes to accommodate TLDs. These are often cylindrical and smaller in size than the InLight nanoDot OSLDs used in this study. OSLDs were selected for this study. They provide a more efficient and less technique sensitive reading than do TLDs and are not subject to the same degree of fading.^{11,24} OSLDs offer high sensitivity and precision with a potential for high accuracy when calibrated under the intended exposure conditions.^{11,24} A logistical advantage of OSLDs is the potential for incremental use, where, unlike TLDs, a minimal charge is lost during analysis (0.04–0.25%).²⁵ The manufacturer discourages continued incremental use since a supralinear

Table 5 The absorbed dose (μSv) to the individual tissues and organs is compared for each scanning protocol investigated

Protocol	Bone marrow	Thyroid	Oesophagus	Skin	Bone surface	Salivary glands	Brain	Remainder tissues			
								Lymphatic nodes	Extrathoracic region	Muscle	Oral mucosa
Full 14 mAs VO1HC	73	112	4	36	202	1104	447	889	856	889	1020
Full 28 mAs VO1HC	139	233	19	67	382	2225	776	1793	1719	1793	2042
Full 42 mAs VO1HC-1	208	335	32	98	573	3372	1150	2683	2582	2683	3058
Full 42 mAs VO1HC-2	206	351	30	100	568	3296	1108	2638	2536	2638	3005
Full 42 mAs VO1HC-3	211	327	33	102	583	3328	1144	2665	2557	2665	3025
Full 42 mAs VO1NC	201	359	31	102	555	3418	1109	2659	2559	2659	3028
Full 42 mAs VO2HC	211	322	30	105	582	3363	1157	2686	2584	2686	3065
Full 42 mAs VO2NC	206	326	29	105	567	3320	1123	2646	2546	2646	3021
Max 14 mAs VO1HC	69	45	0	29	189	707	442	632	630	632	710
Max 28 mAs VO1HC	131	97	5	62	361	1399	785	1242	1239	1242	1383
Max 42 mAs VO1HC	185	145	10	93	511	2016	1090	1793	1794	1793	1989
Mand 14 mAs VO1HC	51	90	3	25	139	956	196	782	701	782	909
Mand 28 mAs VO1HC	94	204	13	41	260	1966	257	1588	1413	1588	1830
Mand 42 mAs VO1HC	129	299	24	56	355	2917	301	2327	2068	2327	2687

HC, high contrast; Mand, mandibular; Max, maxillary; NC, normal contrast; VO1, high resolution setting; VO2, standard resolution setting.

Table 6 The mean absorbed organ and tissue dose for a maxillary scan is represented as a percentage of the dose for a mandibular scan

<i>Bone marrow</i>	<i>Thyroid</i>	<i>Oesophagus</i>	<i>Skin</i>	<i>Bone surface</i>	<i>Salivary glands</i>	<i>Lymphatic</i>		<i>Extrathoracic</i>		
						<i>Brain</i>	<i>nodes</i>	<i>region</i>	<i>Muscle</i>	<i>Oral mucosa</i>
140%	49%	27%	146%	140%	71%	298%	79%	88%	79%	76%

dose response is noted when accumulated dose exceeds 300 cGy, and the uncertainty of measurements can increase from 2% to as much as 4%. A 100 cGy dose corresponds to an approximate signal of 130,000 counts with the OSLD used.¹¹ For logistical reasons, OSLDs were used for incremental measures with up to three protocols measured prior to annealing for reuse. The accumulated signal did not exceed 100,000 counts and was therefore well within the range of a linear dose response.

In this study, a RANDO phantom previously prepared for use with TLDs was modified to accommodate OSLDs using five 2.15-mm thick PMMA templates designed to fit between the phantom sections. This effectively increases the overall dimensions of the phantom by approximately 11 mm and could therefore affect absorbed doses by displacing OSLDs located more superiorly and inferiorly further from the incident primary beams.

Initial imaging of the RANDO phantom revealed that the skull was not accurately related to the soft-tissue surface contours with the hard tissue chin displaced superiorly (Figure 4). Each phantom is composed of a natural human skull cast inside a standard soft-tissue mould, and thus each phantom is unique. Size differences in the natural skulls used to fabricate each phantom may produce variability in the phantom section containing a given anatomical landmark as well as the relationship of tissues and structures to the primary beam. In this study, we chose to place internal dosimeters according to hard tissue references and external dosimeters according to soft tissue landmarks. Accuracy in the spatial relationship between internal and external dosimeters was limited by the phantom anatomy. The phantom was oriented with the goal of maximizing exposure of hard-tissue structures and soft-tissue contours of interest. It is possible that superior and inferior structures were further displaced from the primary beam owing to the anatomy of the phantom *vs* that of humans. An image obtained with lead foil in place of dosimeters revealed reasonable approximation to the tissues and structures of interest. Although the phantom anatomy and need for modification to accommodate OSLDs does present some limitations, comparisons between protocols within this study remain valid.

One comparison of interest is the effect of resolution and contrast settings on effective dose. With the Sirona GALILEOS Comfort CBCT scanner, the user has the options of high or standard contrast and resolution settings. A full maxillomandibular scan was completed at 42 mAs for all four resolution and contrast combinations. There was little difference in effective dose between the four protocols. The manufacturer reports that both high- and standard-resolution options produce a data volume set with $512 \times 512 \times 512$ voxels.

The VO1 secondary reconstruction can be generated by the software only if the VO1 setting is selected. Specific information regarding the difference between high and standard settings for resolution and contrast proved difficult to obtain. Limited information available suggested that pixel binning occurs if the VO1 setting is selected, thus transferring data with a voxel size of 0.3 mm but maintaining the capability of reconstructing an image with 0.15 mm voxel size post acquisition. The standard resolution does not maintain this capability but may as a result allow for faster data transfer. It is likely that the same projection set is acquired regardless of the contrast and resolution software settings selected prior to image acquisition.

This study quantifies the substantial reduction in dose that is obtained when the FOV is minimized. The results of this study show that collimation to a maxillary or mandibular scan results in a reduction in effective dose of approximately 29% and 24%, respectively; this exceeds the approximate 15% reported by the manufacturer.¹⁷ It should be noted that no modifications were made to the percentage of tissue irradiated when calculating the effective dose for partial scans. It could be argued that reductions should be made, which could result in additional reduction in effective dose when partial scans are selected. A review of previous literature revealed only one study where the fraction of tissue irradiated for partial scans was adjusted based on the opinion of the radiologist.²⁶ No modification to the fraction of tissue irradiated was elected, as it was observed that all dosimeters registered a significant dose even for partial scans. This approach reduces the risk of under reporting the effective dose for partial scans but may as a result neglect to report the full benefit in dose reduction when selecting a collimated beam protocol.

The primary aim of this study was to investigate the impact different parameters had on effective dose. Ideally, each of the 72 potential protocols would be investigated individually. This would allow more robust statistical analysis of linearity when comparing effective dose with changes in mAs. In addition, changes in contrast and resolution settings were investigated only for full maxillomandibular scans at 42 mAs. While a similar data acquisition process is likely for the smaller FOVs and mAs settings, this cannot not be concluded with certainty.

The highest mean effective dose calculated was for a full maxillomandibular scan at maximum (42) mAs and was 103 μ Sv based on the 2007 ICRP¹⁸ tissue weighting factors. The lowest mean effective dose was 23 μ Sv for a maxillary scan at 14 mAs. These effective doses remain significantly less than that associated with conventional diagnostic CT scanners and the annual exposure owing to background radiation.^{19,21–23} Unfortunately, these effective

dose values were difficult to reconcile with values reported by other investigators of the GALILEOS Comfort scanner. Adoption of standardized dose calibration protocols for dental CBCT would greatly reduce this disparity.¹⁴

Effective dose is important when determining the optimum scanning protocol for an individual patient and clinical question. The information provided by this study is useful in understanding the relative effects that each parameter has on the effective dose. It is important to understand that establishing an optimum protocol must also consider image quality. Collimation reduces the amount of structures visible on the image while changes in mAs, contrast and resolution can and often do impact the image quality. Such changes may or may not impact diagnosis. Future study of the impact of the same parameters on image quality (*e.g.* standard tests of spatial and contrast resolution) would provide valuable information in determining optimum scanning protocols using the Sirona GALILEOS Comfort CBCT scanner.

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