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ORIGINAL ARTICLE Head neck imaging

Comparison of the neck image quality of two different CT scanners

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ABSTRACT

Purpose: To compare image quality and radiation dose of two different CT scanners in neck imaging.

Material and Method: 200 consecutive patients with neck CT were selected from 2014 to 2017: 100 patients underwent a CT exam at CT1 and 100 patients were examined with CT2. At both CTs, a tube voltage of 120 kVp, and a reconstructed slice thickness of 3 mm with a soft Kernel were applied. The signal-to-noise ratio (SNR) and the contrast-to–noise ratio (CNR) were calculated from the level of the vertebral body C5. SNR and CNR were standardized to the dose level used (√(CTDI)). Mann Whitney test was applied to compare the image technical/quality parameters and the figure of merit (FOM) of both CTs.

Results: Significantly less radiation dose was administered at CT1 (9.5±6.7 mGy) compared to CT2 (11.6±2.1 mGy; $P < 0.001$). In contrast, the image parameters were superior on CT2: SNR and CNR were 8.1±3.0 and 11.0±6.1 at CT1 and 11.5±7.3 and 12.2±5.0 at CT2 ($P < 0.001$ and 0.024 respectively). The standardized SNR, CNR, and FOM were not significantly different between the two scanners.

Conclusion: The presented results suggest that CT2 provides significantly better image quality delineated by SNR and CNR when compared to CT1. However, the dose adjusted image quality and the figure of merit of the two CTs were the same.

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KEY WORDS
Computed-Tomography; radiation dose; image quality; signal to noise; contrast to noise; figure of merit

Introduction

Computed tomography (CT) is the most used imaging method for the head and neck region due to its wide availability, low cost, and scan time compared to MR imaging. However, the use of this modality is well known to contribute to the central part of the cumulative radiation dose in most countries (about 67% in the US and 68% in the UK), and the number of CT scanners and CT exams keeps rising [1]. This relatively high radiation dose has resulted in many scientific publications concerning dedicated scanning dosimetry [2-5], an issue widely reported in radiological protection reports [6,7].

It is proven that compared to the other X-ray-based imaging techniques, CT scans deliver a significantly bigger dose to the patients and the environment. For example, a routine one-phase chest CT scan provides about 100 times (or higher) the amount relative to conventional chest radiography [8, 9].

It should be noted that calculating doses received by patients during a CT scanning are a rather complicated procedure [10]. The computed tomography dose index (CTDI) indicates the dose transferred to the standard phantoms, and the dose length product (DLP) is defined as the CTDI multiplied by the scan length. Although CTDI and DLP are commonly used dose descriptors, they have some disadvantages. More specifically, it has been observed that CTDI is independent of the patient's body size and shape. The DLP, on the other hand, depends on the scan length; it is not dependent on the scan region, and radio sensitivity of organs/tissues is not considered [11, 12], but it is readily captured directly from the CT dose report.

Regarding the scan region, various local and national studies on CT dosimetry indicate that head and neck CT procedures are the most broadly used [13] and comprise about 25 to 30% of all CT scans. Pantos et al. [15] reviewed 42 CT dosimetry publications and indicated the wide range (0.3-8.2 mSv) of the radiation dose from head CT examinations. Smith-Bindman et al. reported that this value for a routine head scan ranged from 0.27 to 5.8 mSv and for a routine neck scan from 0.78 to 8.8 mSv. They proved that the radiation dose from head and neck examinations varies depending on scan parameters and scanner specifications.

The objective goal of this study was to compare image quality and radiation doses of neck CT examinations in the daily clinical routine of two different CT scanners in two tertiary care centers with similar numbers of detector rows.

Material and Methods

Due to an irreversible anonymization process of the patient identifiers, the IRB could be waived (2017).

Patient recruitment

We retrospectively recruited 200 consecutive patients with neck CT scans from 2014 to 2017: 100 patients underwent the CT exam with a 128-row CT scanner of Canon (CT1, Aquillion CXL, Tokyo, Japan), and 100 patients underwent the exam at a 128-row CT scanner of Siemens (CT2, SOMATOM Definition Flash, Siemens Healthcare, Erlangen, Germany). Patients were excluded when they did not match the standard routine CT scan parameters (below) or demonstrated pathologies that complicated signal measurements (especially at the level of C5, described below).

Image acquisition

At both CT scanners, a tube voltage of 120 kVp was applied, and 100 ml Iomeron® was administered with CT Exprès® (both Bracco Imaging SPA, Milan, Italy) with a flow of 3 ml/s without saline flush. Images were acquired with a delay of 70 s with the arms positioned along the body (not over the head). The protocol at the Aquilion (CT1) comprised collimation of 0.5 mm, and a slice thickness of 3 mm was reconstructed using a soft Kernel (FC08). Automated tube current modulation AIDR3D STD was applied, and the reference noise level was set to medium. CT specifications of SOMATOM Definition Flash (CT2) were as follows: automated tube current modulation was applied with 160 reference mAs, collimation 128 mm × 0.6 mm, pitch 0.6, slice thickness

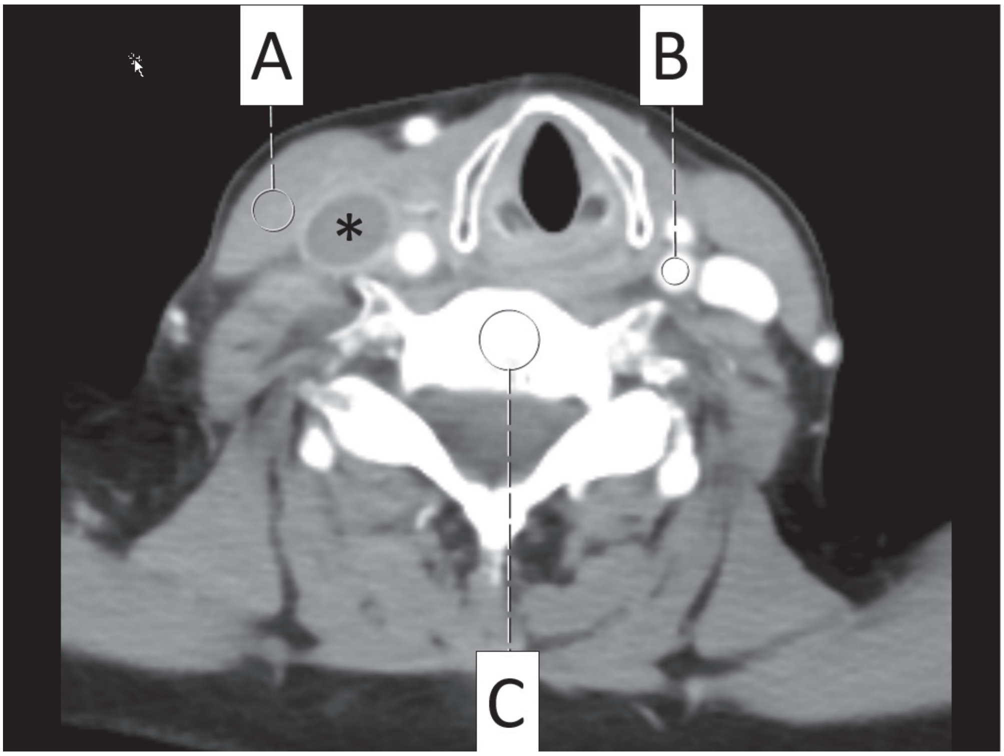

***Fig. 1** – Axial CT image of the Siemens Definition Flash scanner (CT2, 3mm slice thickness) at C5 level. A) Soft tissue density measured 85±5.9 HU (sternocleidomastoid muscle), B) common carotid artery demonstrated 191±5.1 HU, C) vertebral body C5 measured 227 HU±30.4, leading to a signal to noise-ratio of 7.5 and 37.5 for bone and carotid artery. Contrast-to- noise ratio and CT figure of merit were 7.8 and 5.2. Note the thrombosis of the internal jugular vein (asterisk, lemierre syndrome).*

3 mm, soft tissue kernel (I30), and an iterative reconstruction level 3 was used.

Image analysis

Two radiologists with 25 and 15 years of experience in neck imaging read the 200 cases individually. For the objective image quality assessment, they were advised to place regions of interest (ROI) in the neck muscles (sternocleidomastoid muscle), the common carotid artery, and the vertebral body on the level of C5 (Fig. 1 and 2). ROI was programmed to be as large as possible, excluding the fat adjacent to the muscles, the artery wall, or the vertebral cortex (0.5 to 2cm). The structure densities in Hounsfield units (HU) and the noise levels as the standard deviation of the density measurement were captured on the 3 mm thick slices. In addition, the scan range in mm, CTDI and DLP, tube current, and scan time from the image output of the scanner were documented. Furthermore, the subjective image quality for each reader was recorded on a 5-point Likert scale: 1) non-diagnostic; 2) poor, diagnostic confidence significantly reduced; 3) moderate, but sufficient for diagnostics; 4) good; 5) excellent.

Statistical analysis:

The CT scanners used reference mAs and the other one mA with a noise index; we converted the tube current to mAs. The mean measurements of both radiologists were used to calculate the signal-to-noise ratio (SR) and the contrast to noise ratio (CNR) as follows:

$$SNR=HU/Noise$$

$$CNR1=(HU(bone)-HU(st)) \,/\, \frac{Noise(bone)+Noise(st)}{2}$$

$$CNR2=(HU(carotid)-HU(st)) \,/\, \frac{Noise(carotid)+Noise(st)}{2}$$

HU = Hounsfield unit, st = soft tissue

The Figure of Merit (FOM) of a CT is defined as $FOM=CNR^2/CTDI$[15].

Standardized SNR and CNR were calculated by dividing the calculated ratios through the square root of

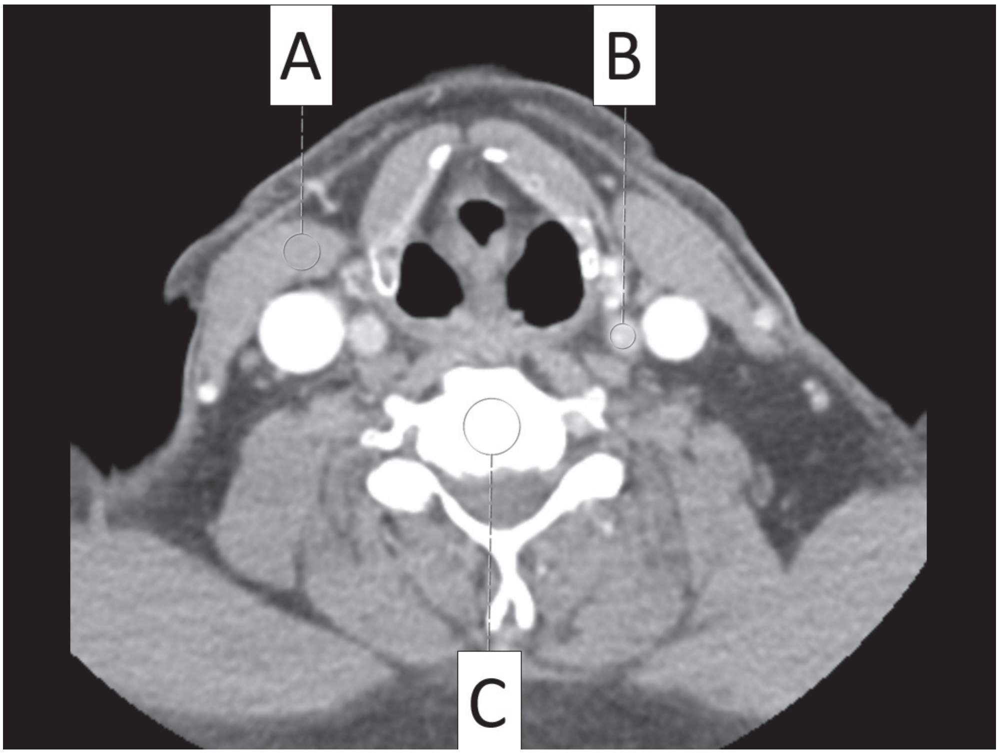

Fig. 2 *– Axial 3 mm CT image of the Aquilion scanner (CT1) at the level of the vertebral body C5. A) Soft tissue density measured 69.9±6.0 HU (sternocleidomastoid muscle), B) common carotid artery demonstrated 126±14.8 HU, C) vertebral body C5 measured 312±64.7 HU, leading to signal-to-noise ratios of 4.8 and 8.5 for bone and carotid artery. Contrast-to-noise ratios of bone and CT figure of merit were 6.8 and 6.0.*

the CTDI. Two-tailed unpaired Mann Whitney test was applied to compare the technical and quality parameters of the images, including the figure of merit (FOM) of both CTs.

Pearson's correlation coefficient between the two readers was calculated for inter-reader correlation with 95%-confidence intervals and p-values. The correlation of HU measurements for soft tissue, bone, and carotid artery was analyzed separately.

Statistical analysis was done on MedCalc, version 7.6.0.0 (MedCalc Software, Mariakerke, Belgium), using a significance level of <0.05.

Results

Image quality and radiation dose

On average, significantly less radiation dose was administered at CT1 (9.5±6.7 mGy) compared to CT2 (11.6±2.1 mGy; p-value < 0.0001). The scan range, standardized tube current and DLP were significantly higher at CT2 (Table 1).

On that account, the image parameters were often superior on CT2 images (Table 2): the signal of the bone was significantly higher with 427±101 HU compared to 390±169 HU on CT1 images (P = 0.0008), and the image noise of neck muscles was substantially lower: 7.0±2.9 HU compared to 8.6±2.6 HU on CT1 (P < 0.0001). SNR and CNR were 8.1±3.0 and 11.0±6.1 at CT1 and 11.5±7.3 and 12.2±5.0 at CT2 (P = 0.0003 and 0.0238, respectively). The standardized SNR, CNR, and FOM were insignificantly better for CT2 (Table 3).

Interreader agreement and subjective image quality

The readers, correlation coefficient (CC) for soft tissue HU was 0.35 (95%-CI: 0.22-0.47; p-value < 0.0001) and the CC for bone and carotid density measurements were 0.47 (95%-CI: 0.36-0.57; p-value < 0.0001) and 0.70 (95%-CI: 0.58-0.79; p-value < 0.0001). Both readers rated the subjective image quality the same for both CT scanners (5=excellent).

Discussion - Conclusions

The presented results suggest that CT2 provides signif-

TABLE 1. - (Adjusted) tube current and radiation dose for both CTs at 120 kV tube voltage.

	CT1		CT2		
	mean	**SD**	**mean**	**SD**	
Adj. Tube current (mAs)	142.6	43.2	176.4	22.1	P < 0.0001
Scan range (mm)	283.0	275.3	325.7	60.7	P < 0.0001
CTDI (mGy)	9.5	6.7	11.6	2.1	P < 0.0001
DLP (mGycm)	244.6	105.3	309.2	66.8	P < 0.0001

CTDI: CT Dose Index, DLP: Dose Length Product; SD: Standard Deviation

TABLE 2. Density measurements for both CTs as means of two readers

		CT1		CT2		
		HU	**SD**	**HU**	**SD**	
soft tissue	density	63.2	9.5	64.3	11.1	P = 0.2678
	noise	8.6	2.6	7.0	2.9	P < 0.0001
bone	density	390.0	169.4	427.3	101.4	P = 0.0008
	noise	58.2	26.5	64.5	41.0	P = 0.8060
CCA	density	163.0	78.6	140.5	49.4	P = 0.0364
	noise	9.3	4.0	6.8	4.1	P < 0.0001

HU: Hounsfield Units; SD: Standard Deviation; CCA: Common Carotid Artery

icantly better image quality delineated by SNR and CNR when compared to CT1. However, at CT2, significantly more tube current and hence radiation dose (CTDI) was used. The dose-adjusted image quality (SNR and CNR) and the figure of merit of the two CTs were invariable. Image quality is known not only based on the signal but also characterized by contrast and noise, so a loss of subjective image quality is less impressive than an increase in noise level. Hence, radiologists rate image quality higher than the measured noise would suggest. Perception of image quality is influenced by spatial resolution, individual vision, and pattern recognition, not all of which can be precisely measured. In this study, subjective image quality was not rated below as the noise level would predict.

Figuring out the necessary image quality and dose to provide adequate task-based performance is challenging [16]. Phantom acquisitions can objectively assess image performance of conventional image metrics, such as spatial resolution, contrast resolution, image uniformity, and CT number accuracy. These metrics are generally helpful for characterizing system performance but are rarely used for selecting patient protocols. Due to this range of tasks, it is challenging to determine a fixed set of acquisition parameters that will provide "necessary" image quality. Moreover, the mini-

TABLE 3. Image quality (SNR, CNR) and figure of merit (FOM) for both CTs

	CT1		CT2		
	mean	**SD**	**mean**	**SD**	
SNR soft tissue	8.1	3.0	11.5	7.3	P = 0.0003
SNR bone	7.9	4.4	8.5	4.3	P = 0.1586
SNR CCA	21.3	14.1	27.9	17.7	P = 0.0105
Adj. SNR soft tissue	3.0	1.5	3.5	2.3	P = 0.2823
Adj. SNR bone	2.8	1.5	2.6	1.9	P = 0.1782
Adj. SNR CCA	7.7	5.7	8.6	6.0	P = 0.3117
CNR bone-soft tissue	11.0	6.1	12.2	5.0	P = 0.0238
CNR CCA-soft tissue	11.8	8.9	14.0	12.1	P = 0.5094
Adj. CNR bone-soft	3.9	2.0	3.8	2.5	P = 0.4576
Adj. CNR CCA-soft	4.3	3.6	4.3	4.0	P = 0.5994
FOM CNR bone-soft tissue	19.1	22.8	20.2	56.2	P = 0.4576
FOM CNR CCA-soft tissue	31.2	60.6	34.8	62.8	P = 0.5926

SNR: Signal to Noise Ratio; CNR: Contrast to Noise Ratio, CCA: common carotid artery; FOM: Figure of Merit; SD: Standard Deviation

mum required image quality varies because of the wide range of clinician preferences and experience. There are no standard protocols for these reasons, but they are developed based on local expertise.

CT protocols, like the other multiple processes at an institution, should be reviewed in specific periods to ensure that the combination of image quality and the dose is the best, as baseline level can be considered the preexisting CT protocols if the referring doctors believe that these have sufficient image quality. Phantom acquisitions with successive image quality or dose reductions should be compared with these baseline protocols. Minimal dose reductions of approximately 5–15% often increase acceptable noise [17]. Calculating these noise increases during phantom scans may lead to minor dose reductions in particular patient studies (e.g., testing little reductions in some follow-up studies may be more appropriate than attempting drops in initial staging studies). These new protocols might be applied to all patients if the contractions were correct in a smaller group of studies. This graduated systematic evaluation of dose reductions and image quality acceptance may result in a more dose-conscious CT use. It should be stressed that over- and under-dosing could be considered medical errors; therefore, dose reductions and dose increases may be appropriate while optimizing CT protocols.

In this study, we used CTDI and DLP, which were captured directly from the CT dose report. However, some controversial results in the literature need to be considered. The first problem with this use is the lack of correction concerning the patient size. For example, Monte Carlo simulations of absorbed patient dose that consider patient size have shown that the adequate amount increases much slower than the CTDI or DLP value. To achieve similar image quality, the scanner output (CTDI) should be increased by about a factor of two as patient size changes from a typical adult abdomen (lateral dimension, 35–40 cm) to an obese adult abdomen (lateral width, 45–50 cm) [18-20]. Even though the scanner output increases by a factor of two, the dose for many of the radiosensitive internal organs used in calculating the effective dose does not increase by the same amount owing to the attenuation of the additional fatty tissue. More precisely, the increase of CTDI by a factor of two in larger patients results in a net gain of the effective dose of only approximately 20%–30% [21,22]. An important implication of taking patient size into account when estimating patient dose and prescribing the correct scanner output settings is that considerable variation in CTDI-based dose metrics can be expected. Facilities that adjust their CT technique appropriately for patient size, whether with the use of manual technique charts or automatic exposure control, will prescribe a wide range of scanner output (CTDI) values [23-27]; reflecting the facility's conscientiousness in "right-sizing" the dose settings based on specific patient body habitus. Furthermore, the different diagnostic tasks and clinical applications introduce variability not only in the image quality criteria but also in the scanner output settings, even for patients of the same size. Thus, radiation management in CT requires choosing the correct settings for scanner output for both patient size and the imaging task. **R**

IRB

Due to an irreversible anonymization process of the patient identifiers the IRB could be waived (2017).

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