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The impact of integrated circuit detector technology in routine MDCT: Improved image quality compared to conventional detector design for constant SSDE

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ABSTRACT

Purpose: The aim of this study was to compare the image quality and radiation dose of routine CT scans acquired with a third-generation scanner built with integrated circuit detector technology (Stellar®-detector) to those acquired with a conventional solid-state detector (UFC-ultra fast ceramic solid-state detector) equipped with a discrete system using the same protocol setting on the same 128-MDCT scanner.

Material and Methods: 262 routine CT examinations of 240 patients (140 male, 100 female) with a mean age of 61.7 years (range: 20-89 years) were reviewed retrospectively. 131 examinations were acquired with a UFC solid-state detector and 131 with the new digital Stellar®-detector (23 CT chest and abdomen, 50 CT chest and 58 CT abdomen). The following parameters were recorded: BMI, scan length, kVp, CT dose index ($CTDI_{vol}$) and dose-length-product (DLP). $CTDI_{vol}$ multiplied with a conversion factor depending on the patient size determined the size-specific dose estimates (SSDE). The image noise was defined as standard deviation of mean attenuation values in Hounsfield units and was quantitatively assessed by circular region of interest at predefined size of 31.4 mm^2 (20 mm diameter) manually placed in the center of the left liver lobe. Iterative reconstruction (SAFIRE, level 3) was used.

Results: Changing the detector unit from a UFC solid-state detector with distributed electronics (conventional) to a fully-digital Stellar® detector leads to a significant decrease of image noise by 7.5% in the liver (mean ± SD: 25.5 ± 5.4 HU *vs.* 23.6 ± 5.8 HU; p=0.0019) for all CT scans included in this study. In the subgroup CT chest (n=100) the average noise reduction was 6% (mean ± SD: 28.2 ± 5.7 HU *vs.* 26.5 ± 7.1 HU; p=0.038) and 6,4% for the subgroup CT abdomen (n=116) (mean ± SD: 23.2 ± 3.7 HU *vs.* 21.9 ± 3.7 HU; p=0.022). kVp, mAs and total scan length were the strongest predictors for effective dose (ED, p<0.0001) whereas SSDE shows

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a strong positive correlation with kV, mAs and effective diameter (*p*<0.0001). The liver noise correlates negatively with both ED as well as SSDE (*p*=0.0002 and *p*<0.0001, respectively).
Conclusion: For the same applied radiation level the implementation of an integrated circuit digital detector reduces image noise compared to detectors having distributed electronics. Using a detector with integrated circuit detector technology provides great potential to reduce radiation dose. Further measures such as changing the CT-image noise level must be performed to achieve dose reduction.

KEY WORDS
tomography, x-ray computed/instrumentation; multidetector; radiation dosage; effective dose; size specific dose estimate

1. Introduction

In the past decade, the advances in computer tomography (CT) technology and applications have increased the clinical utilization of CT, creating concerns about individual and population doses of ionizing radiation. In the United States, the number of CT studies performed has increased more than twentyfold in the past 25 years [1] and the number of CT scans continues to increase by approximately ten percent per year [2, 3]. Consequently, radiation exposure is increasing and the U.S. per-capita annual effective dose from medical procedures has increased about sixfold from 0.5 mSv in 1980 to 3.0 mSv in 2006 [3], with essentially all of the increase deriving from medical imaging, especially CT [4]. Recently, it has been reported that new cancer cases in the United States attributed to CT scanning are estimative between 0.7% - 2.0% per year [5, 6].

Although there is no marker that identifies a cancer as being radiation induced, it is universally accepted that exposure to ionizing radiation increases cancer risk [4, 7, 8]. Concerns about the risk associated with exposure to ionizing radiation lead to the principle of "*as low as reasonably achievable*" (ALARA), which has also driven the need to better quantify the effect of noise [9,10].

The doses used for CT examinations are highly variable-across patients and institutions, and are higher than needed for diagnosis. Scanner manufacturers have subsequently implemented several options to appropriately manage or reduce the radiation dose from CT. An essential step in the process of dose reduction is the development of hardware. New detectors are essential for improving the Detective Quantum Efficiency (DQE) of the device if diagnostic image quality is to be maintained when less photons are contributed to the imaging process. A new generation of detectors, integrated circuit detectors (IC detectors), combines the photodiode and the analogue to digital converter (ADC) in one application specific integrated circuit (ASCI). It is designed to reduce electronic noise, power consumption and heat dissipation by reducing the path of the signal [11].

Electronic noise is dependent of the number of photons reaching the detector. At higher doses electronic noise is insignificant. However, in low dose examinations electronic noise becomes dominant and degrades image quality. Reduction in electronic noise becomes more important as CT practice moves toward lower dose scanning. As the number of obese patients continues to increase, electronic noise could have a significant impact on image quality and could become a primary constraint to the ability to reduce patient dose [12]. In low dose scans such as in low-kV datasets produced e.g. with CARE kV effective designed detectors with low electronic noise levels and a high dynamic range are of benefit.

The purpose of this study was to compare the image quality and radiation dose of routine CT scans acquired with a third-generation scanner built with integrated circuit detector technology to those acquired with a conventional solid-state detector equipped with a discrete system.

2. Material and methods

The study was approved by the institutional review board. Informed consent was waived by the institutional review board due to the retrospective nature of this study. The study was performed according to the ethical standards as described by the Declaration of Helsinki.

262 routine CT examinations of 240 patients (140 male, 100 female) with a mean age of 61.6 years of age

Table 1. Patients demographics. Data expressed as mean ± SD. ICD=Integrated Circuit Detector. CD=Conventional Detector. BMI=Body Mass Index. ED=Effective Dose. SSDE=Size-specific Dose Estimate

Detector	Examination	Age [years]	BMI [kg/m²]	eff. Diameter [cm]	Tube Voltage [kV]	Tube Current [mAs]	Noise liver	ED [mSv]	SSDE [mGy]
ICD	chest CT (*n*=50)	60.7 ± 12.1	28.2 ± 4.3	30.4 ± 4.8	100.8 ± 4.0	134.2 ± 46.1	26.5 ± 7.1	2.6 ± 1.1	6.3 ± 2.3
CD	chest CT (*n*=50)	58.5 ± 14.2	25.2 ± 4.9	30.1 ± 2.8	101.2 ± 7.5	122.9 ± 38.2	28.2 ± 5.7	2.5 ± 0.9	6.4 ± 1.8
p-value		>0.05	>0.05	>0.05	>0.05	>0.05	**0.038**	>0.05	>0.05
ICD	abdomen CT (*n*=58)	63.5 ± 12.3	25.5 ± 6.1	29.5 ± 6.2	106.9 ± 14.8	199.3 ± 72.1	21.9 ± 3.7	6.7 ± 3.3	11.0 ± 3.4
CD	abdomen CT (*n*=58)	61.6 ± 16.1	25.0 ± 5.7	29.6 ± 5.0	106.3 ± 12.6	195.5 ± 70.5	23.4 ± 3.7	7.1 ± 4.7	11.3 ± 3.9
p-value		>0.05	>0.05	>0.05	>0.05	>0.05	**0.022**	>0.05	>0.05
ICD	all CT (*n*=131)	62.4 ± 13.3	26.2 ± 6.1	30.4 ± 5.6	106.4 ± 13.4	169.1 ± 67.3	23.6 ± 5.8	5.7 ± 4.0	9.3 ± 3.8
CD	all CT (*n*=131)	61.0 ± 15.8	25.1 ± 5.0	29.8 ± 3.9	104.3 ± 11.4	162.5 ± 67.1	25.5 ± 5.5	5.4 ± 4.1	9.1 ± 3.8
p-value		>0.05	>0.05	>0.05	>0.05	>0.05	**0.0019**	>0.05	>0.05

(range: 20-89 years) were reviewed retrospectively and patients were included continuously for the time period before and after September 2012 until we reached the same patients number in both groups. All CT studies were clinically indicated and were performed on a dual source 128-multi-detector CT (Somatom Definition Flash; Siemens Healthcare, Erlangen, Germany): 131 examinations with a UFC solid-state detector (scans acquired before September 2012) and 131 with the digital Stellar®-detector (scans acquired after September 2012). The following scanning parameters were used: Pitch 1.0; rotation time 0.5 sec, carekV 100-140 kVp and 140 reference mAs (caredose), scanning delay 70 seconds for the portal venous phase. Iterative reconstruction level 3 was used for image generation. All measurements were performed on soft tissue kernel images (B20). They were reviewed on a PACS-workstation (Picture Archiving and Communication System IDS7; Sectra, Linkoping, Sweden) at a soft tissue window (level: 35 HU, width: 250 HU). The scans included either the entire torso (chest, abdomen and pelvis) or the subregions CT chest and CT abdomen.

The following parameters were recorded: BMI, scan

Table 2. Multimple regression analysis for ED and SSDE as dependent variables. kVp, mAs and total scan length were the strongest predictors for ED ($p<0.0001$), whereas SSDE shows a strong positive correlation with kV, mAs and effective diameter ($p<0.0001$). For both ED and SSDE the liver noise correlates negatively ($p=0.0002$ and $p=0.0002$ and $p<0.0001$ respectively)

Independent variable	Dependent variable	Zero order correlation	*p*-value
BMI	ED SSDE	0.481 0.392	0.2034 0.3614
Noise liver	ED SSDE	-0.163 -0.278	0.0002 <0.0001
kV	ED SSDE	0.510 0.364	<0.0001 <0.0001
mAs	ED SSDE	0.664 0.778	<0.0001 <0.0001
Stellar	ED SSDE	0.055 0.061	<0.1383 <0.0075
Scan length	ED	0.622	<0.0001
Effective diameter	SSDE	0.362	<0.0001

length, kVp, CT dose index ($CTDI_{vol}$) and dose-length-product (DLP). Patient dimensions such as maximal antero-posterior (AP) thickness and lateral (LAT) width were determined on an axial CT image after the scan. The effective radiation dose (ED) was estimated from the DLP with an organ weighting factor (k): ED[mSv] ≈ k × DLP[mGy×cm]. The SSDE, which considers patients size and scanner output (CTDIvol), was determined for each patient by calculating the square root of the summation of the AP and LAT representing the patient size defined as effective diameter ($\sqrt{AP + LAT}$). The AAPM Report 204 [14] provides tables based on effective diameters to find a conversion factor (f_{size}), that when multiplied by $CTDI_{vol}$, yields the SSDE (SSDE=$f_{size} \times CTDI_{vol}$ [mGy]) [13, 14].

Patients' records were retrieved from the clinical information system (KIS), integrated patient dossier (i-pdos), CompuGroup medical (CGM, Phoenix, version 7.8.0.1.5), Koblenz, Germany. CTDI was recorded from the scanner console. The image noise was defined as standard deviation (SD) of mean attenuation values in Hounsfield units and was quantitatively assessed by circular region of interest at predefined size of 31.4 mm^2 (20 mm diameter) manually placed in the center of the left liver lobe at the level of the falciform ligament. For all included patients same protocol setting was used for the exams acquired on the same scanner.

The Mann - Whithney U Test for non-normally distributed independent samples was used to calculate the significance level for noise reduction between the two different scanners in the subgroups (CT chest and CT abdomen) and for all CT scans (CT chest and abdomen, CT chest and CT abdomen). Multiple regression analysis was used to identify unique predictors of estimated dose and SSDE. In addition, we analysed the patients with a BMI over 25 kg/m^2 because the proposed effect is supposed to be stronger in these patients. Statistical significance was defined as $p<0.05$. Statistical analysis were performed on MedCalc® Version 7.6.0.0 (MedCalc Software, Mariakerke, Belgium).

3. Results

Between the randomly selected patients included in the two groups scanned with the integrated circuit detector (ICD) *vs.* conventional detector (CD) the patients' mean age (62.4 ± 13.3 *vs.* 61.0 ± 15.8 years; $p>0.5$), mean BMI (26.2 ± 6.1 *vs.* 25.1 ± 5.0; $p=0.38$) and effective diameter (30.4 ± 5.6 *vs.* 29.8 ± 3.9; $p=0.27$) were not different (**Table 1**).

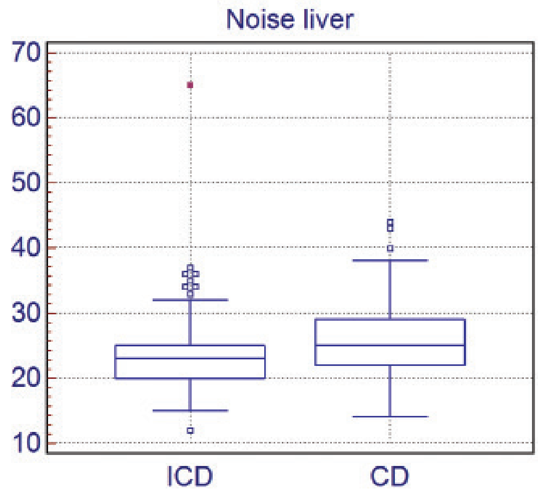

Fig. 1. *Boxplots show effect of a conventional detector with discrete circuit (CD) and of a third generation detector with integrated circuit technology (ICD) on image noise in routine CT scans on a dual source MDCT. Image noise in CT scans of the torso was significantly reduced by 7.5% in the liver (p=0.0019) for the same scanner settings and output level*

Image noise was significantly reduced by 7.5% in the liver (mean ± SD: 25.5 ± 5.4 HU *vs.* 23.6 ± 5.8 HU; $p=0.0019$) for all CT scans included in this study ($n=262$) (**Fig.1** and **2**). In the subgroup CT chest ($n=100$) the average noise reduction was 6% (mean ± SD: 28.2 ± 5.7 HU *vs.* 26.5 ± 7.1 HU; $p=0.038$) and 6.4% for the subgroup CT abdomen ($n=116$) (mean ± SD: 23.2 ± 3.7 HU *vs.* 21.9 ± 3.7 HU; $p=0.022$). The ED did not show significant difference before and after changing the detector unit (5.4 ± 4.1 *vs.* 5.7 ± 4.0; $p>0.05$). The SSDE didn't differ significantly in both groups (9.3 ± 3.8 *vs.* 9.1 ± 3.8; $p=0.3$) (**Table 1**). Multiple regression analysis for ED and SSDE as dependent variables showed that kVp, mAs and total scan length were the strongest predictors for ED ($p<0.0001$) whereas SSDE showed a strong positive correlation with kV, mAs and effective diameter ($p<0.0001$). The liver noise correlates negatively with both ED as well as SSDE ($p=0.0002$ and $p<0.0001$ respectively) (**Table 2**).

Patients with a BMI over 25 kg/m2 did not show a significant difference between the two groups scanned in the mean BMI (29.3 ± 3.8 *vs.* 29.9 ± 5.3; $p>0.05$).

Image noise was not significantly reduced by 7,8% in the liver (mean ± SD: 27.4 ± 5.1 HU *vs.* 25.3 ± 7.0 HU; $p=0.054$) for all CT scans included in this study ($n=129$). In the subgroup CT chest ($n=46$) the average noise reduc-

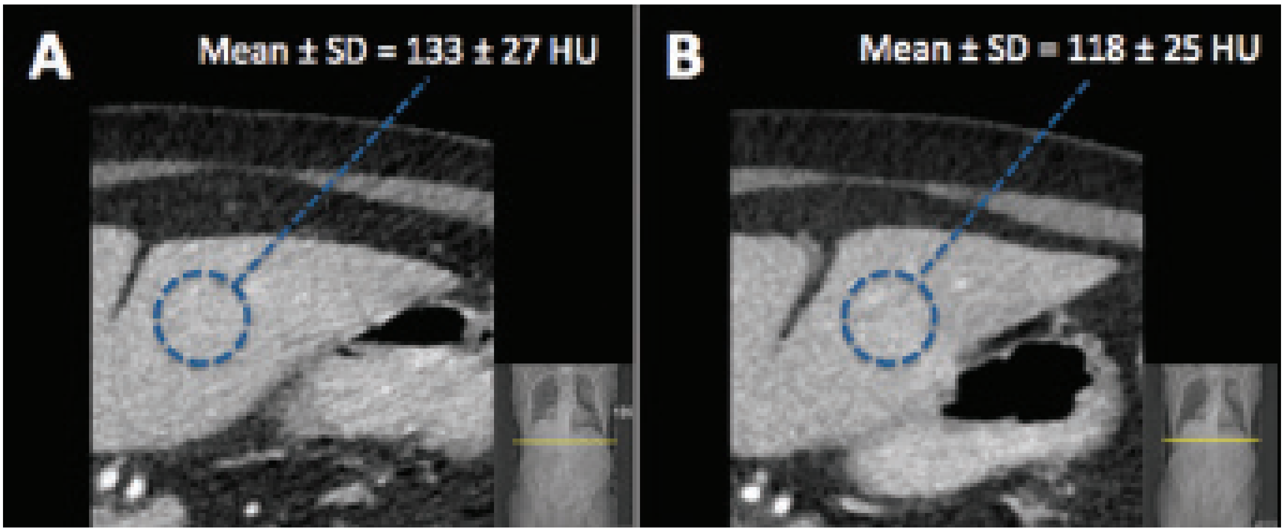

Fig. 2. *78 y/o patient with body mass index of 25.7 kg/m². Transverse CT scans of the torso (level 50 HU, width 450 HU) obtained with a conventional detector* ***(A)*** *and in the same patient with a third generation detector with integrated circuit* ***(B).*** *Images acquired with integrated detector technology are associated with lower noise levels (27 vs. 25 HU, noise reduction of 7,4 %)*

tion was 5.3% (mean ± SD: 31.0 ± 5.7 HU *vs.* 29.3 ± 8.5 HU; $p>0.05$) and 8.3% for the subgroup CT abdomen ($n=57$) (mean ± SD: 25.4 ± 3.3 HU *vs.* 23.3 ± 4.1 HU; $p>0.05$). The ED did not show significant difference before and after changing the detector unit (6.5 ± 4.7 *vs.* 7.1 ± 4.5; $p>0.05$). The SSDE didn't differ significantly in both groups (10.0 ± 4.2 *vs.* 10.6 ± 4.0; $p=0.3$).

4. Discussion

The overall goal in CT scans is to achieve high quality images with the lowest possible dose which means finding the accurate noise level for a specific indication. By reducing radiation dose image noise increases and therefore image quality decreases.

The newly launched IC detector showed significant decrease of electronic noise by 7.5 % in the left liver lobe for the same scanner output level for all CT scans included in this study. In the subgroup CT chest the average noise reduction was 6% and 6.4% for the subgroup CT abdomen. This effect is most probably based on the reduced electronic noise of the integrated system in the third generation scanner. Duan et al. reported a substantially reduced level of electronic noise, resulting in reduction in image noise and artifacts in a phantom study compared with detectors having distributed electronics supporting our data [12]. Christe et al. stated a reduction of noise for air at a constant dose level of 31% and an increase of contrast to noise for soft tissue of 25%. Trading image quality for dose reduction, these results meant a dose reduction of 54% and 70% for soft tissue and air at a constant noise level [15]. In settings with low photon counts where improved image quality is mandatory, such as in low dose CT examinations or in obese patients with high signal absorption, improved signal quality can be achieved with the integrated detector at same radiation level. Low dose examinations profit directly from the new integrated detector [11, 12]. Other authors demonstrated that low-dose CT images acquired with IC detectors provide better depiction of fine osseous structures of temporal bone in comparison with conventional detector [16]. In addition, added value of IC detector was analyzed in low-dose head CT showing superior objective and subjective image quality [17].

Moreover, decreased noise level by same dose exposure can be combined to other dose reduction techniques and improve image quality such as in iterative reconstructions [18, 19].

In the subgroup of patients with a BMI over 25 kg/m², the reduction of the noise was even higher for the abdominal studies. In the chest, no additional effect was observed. This basically means that the superiority of the IC detector is even more pronounced in body areas were less photons reach the detector.

For quantitative measurement of radiation dose delivered by CT scanning there are various methods. Among the most relevant are the effective dose, expressed in Sievert (Sv), and the volume CT dose index ($CTDI_{vol}$), expressed in milliGray (mGy). The latter allows the direct comparison of the radiation dose from different scanner parameter settings and even between scanners [20]. The effective dose on the other hand reflects radiation detriment averaged over gender, age and risk of population. It conveys the relative potential for harm from the radiological examination and reflects the stochastic risk (e.g. cancer induction) from an exposure to ionizing radiation [20-22]. Another X-ray tube output metric is the dose-length product (DLP), which corresponds to CTDIvol multiplied by the length of the scan (cm), meaning that a longer anatomical coverage delivers more radiation for a given $CTDI_{vol}$ [20]. Both parameters, CTDIvol and DLP are heavily dependent on scan parameters such as the peak kilovoltage (kVp) and the tube current-time product mAs=mA x rotation time / pitch, where mA is the tube current in milliAmperes [23]. In our retrospective analysis of routine CT scans of the whole body the ED did not show significant differences before and after changing the detector. Analysis of CT colonography acquired with IC detector showed similar results with significantly lower image noise, whereas $CTDI_{vol}$ was unchanged [24].

The American Association of Physicists in Medicine (AAPM) Report 204 introduced in 2011 a novel method to estimate patient dose. While $CTDI_{vol}$ is a measurement only of scanner output it does not appreciate patient size [13, 25]. Conversion from $CTDI_{vol}$ to the SSDE requires the use of the tabulated size-depended conversion factors (f_{size}) combined with a measurement of patient size ($SSDE = f_{size} \times CTDI_{vol}$ [mGy]) [14].

In this retrospective analysis neither SSDE nor ED differ between the groups. However, in both groups kVp, BMI and total scan length were the strongest predictors for ED, whereas SSDE showed a strong positive correlation with effective diameter, kV and mAs ($p<0.0001$). The liver noise correlates negatively with both ED as well as SSDE. Therefore, a dose reduction can be achieved using lower tube parameters. Further measures such as changing the CT image noise level must be performed to achieve dose reduction by decreasing radiation level. Brodoefel H. et al. already showed an aggressive dose reduction of almost 50 % in low-dose head CT through combination of iterative reconstruction with IC detector design [18].

While other research groups included a small number of patients (Morsbach et al. [11]: 20 patients, He J. et al. [16]: 35 patients, Korn A. et al. [17]: 25 patients) we investigated a quite equal patient group consisting of a significantly larger collection of 262 patients.

The retrospective analysis of the data is one of the limitations of the study. However, a prospective analysis is ethically not feasible, as it would require investigation of patients with an outdated technique. While measuring the pixel standard deviation is a simple tool to measure the noise level it provides no information about noise spatial characteristics introduced into the noise by the reconstruction algorithm. Furthermore, we did not modulate the radiation level settings as an important tool for changing the image noise level leading to potential dose reduction. Last, we did not study the effect of the detector on the contrast to noise ratio. This was beyond the scope of this study, as it would have required a phantom study.

In summary, for the same applied radiation level the implementation of an integrated circuit digital detector reduces image noise compared to detectors having distributed electronics and therefore leads to superior image quality for the integrated detector system. Using a detector with integrated circuit detector technology provides great potential to reduce radiation dose. Further measures such as keeping the CT-image noise level constant to achieve dose reduction must be performed. R

Conflict of interest:
The authors declared no conflicts of interests.

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