RECENT ADVANCES IN COMPUTED TOMOGRAPHY RADIATION DOSIMETRY

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ABSTRACT: Computer tomography (CT) has proved fundamental in image evaluation throughout the past three decades. By combining rapid scanning with high-quality data sets, multi-detector technology continues to influence practice patterns. This has led to new applications and improved use in conventional applications. However, the increased use of CT has generated significant concern regarding the high radiation doses received by patients during CT scans compared to traditional radiography examinations. Many studies have been undertaken on minimizing patient dose and adhering to the as low as reasonably achievable (ALARA) principle. A total of 40 articles from PubMed, Science Direct and Google Scholar were systematically summarized in this review paper to introduce the growth of CT scan from single-slice to multi-slice technology from 2000 until December 2020 as well as dual-energy and multi-detector CT technologies. The important role of utilizing CT radiation dosimeters for CT dose measurement is defined included CT dose reduction techniques.

KEYWORDS*: CT dose reduction techniques; CT radiation dosimetry; Dual-energy CT; Multi-detector CT; Multi-slice CT*

1.0 INTRODUCTION

1.1 CT Generation from Single-Slice to Multi-Slice Scanner

CT scans have been in clinical use for about 30 years, and they are now used in almost every hospital environment. CT imaging technology has advanced from scanning a single slice to spiral CT, and then to a multi-slice scanner. In the first conventional CT scanners, the tube and a row of detectors are located on opposing sides of a spinning ring around the patient. Since the tube is connected to the power cables, it cannot rotate continuously. The scanner stops and rotates in the opposite direction after each rotation. It takes one rotation to acquire an axial image, with a thickness of 1 cm, and the process takes about one second per rotation [1]. The movement of the patient through the scanner is made by moving the table between each slice. Hence, there are some drawbacks to conventional scanners, scan time is long to run, and this results in noise due to movement or breathing. In addition, scanners have an irregular ability to reformat in different planes, dynamic contrast analyses are also extremely difficult, and even small tumors can be missed between slices [2].

Since conventional CT scanners are considered as time-consuming. Hence, great efforts were made in the late 1980s to increase scanning volumes in less time [3]. This concept led to the creation of a new technique that was utilized by moving the table while the x-ray tube and detectors rotate for several times to perform scans of tissue. As a result, the beam travels in a circle around the patient. This is technically known as spiral CT, however others refer to it as helical CT [4]. In 1998, the introduction of a new generation of CT scanners was made at the Radiological Society of North America (RSNA) meeting in Chicago [5]. These scanners are known as multislice CT (MSCT) scanners. Multislice CT was developed by adapting single-slice CT (stop-and-go, slice-by-slice data acquisition), which was more flexible in dealing with the limitations of conventional CT. In doing so, multislice CT scanners could reduce the time for data acquisition, thus speeding up the volume coverage. A "turbocharged" spiral scanner is another name for a MSCT scanner [6]. The single row of detectors used by spiral and conventional scanners to detect the x-ray beam once it has passed through the patient. On the other hand, up to eight active detector rows can be used with a multi-slice scanner. Coverage of a given volume of tissue is increased because of the increased detector and tube rotation times that take less than a second to complete [7], [8]. In newer multi-slice scanners, faster computer software is also included, making the processing of reconstructed images and post-processing faster. A multi-slice scanner with four detectors theoretically will reduce scan time by a quarter compared to a single slice spiral scanner [1], [2].

Multi-slice scanners have the ability to scan images two to three times faster than single-slice scanners in practice. As seen in Figure 1, the number of slices each revolution has increased steadily over time.

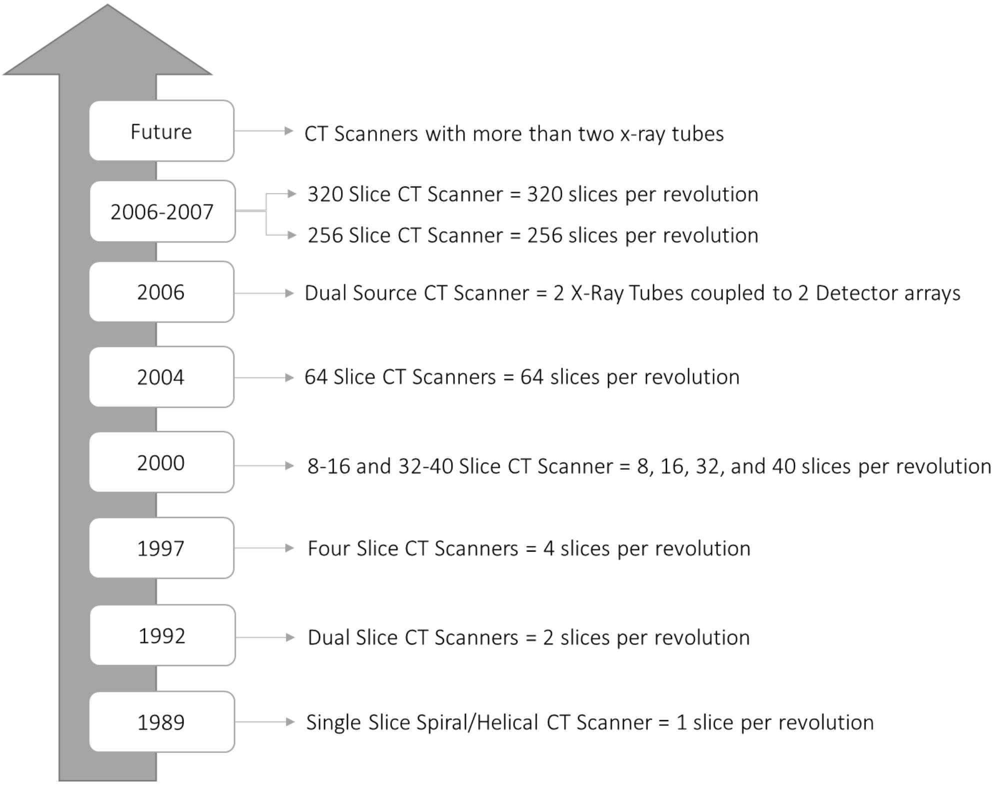

Figure.1 The evolution of MSCT scanners, including the DSCT scanner [2]

1.2 Overview of Multi-detector CT Technology

There have been considerable developments in computer tomography technologies in recent years such as cone beam, extreme multi-detector, dual-energy, portable, and phase contrast in CT technology. The primary difference between single-slice CTs and multi-slice CTs is a technology for detectors. The "multi-detector-row" type of MDCT scanners refers to multiple arrays (rows) in the z-direction [9]. Now available MDCT scanning systems utilize CT geometry of the third generation, which rotates the range of detectors together with the x-ray tube(s). Due to their improvements, these scanners offer enormous versatility not only in detector technology, but also in the data acquisition systems (DAS), X-ray tubes, and other subsystems. One example is that whereas MDCT scanners feature multiple rows of detectors, the collected data from multiple rows of detectors can be merged as if collected from one single detector [10], [11].

One of the most significant components of the CT imaging chain is the detector because it detects radiation transmitted by the body and transforms it into electrical signals that are then digitized and forwarded to the computer for processing and image creation. Currently, two types of detectors detect and transfer radiation into digital data [12]. Scintillation and photon-counting are examples of these detectors [13]. Figure 2 illustrates the major components of two types of scintillation detectors: traditional energy integrating and dual-layer. Scintillation crystals in the MSCT include cadmium tungstate ($CdWO_4$); high purity ceramic material, rare earth oxides, based on rare-earth-doped compounds as yttria; and gadolinium oxysulfide ultrafast ceramic. GE Healthcare has implemented gemstone spectral imaging, the world's first garnet scintillator for use in computed tomography. Additionally, Philips Healthcare also utilizes zinc selenide triggered with tellurium in their dual-layer scintillator detectors [14]. The photon-counting detector is a new technology that is being investigated in prototype scanners such as Siemens SOMATOM Definition Flash [15]. Semiconductors such as cadmium telluride (CdTe) and cadmium zinc telluride (CZT) are used because they can immediately convert x-ray photons into pairs of electron-hole (electric charge).

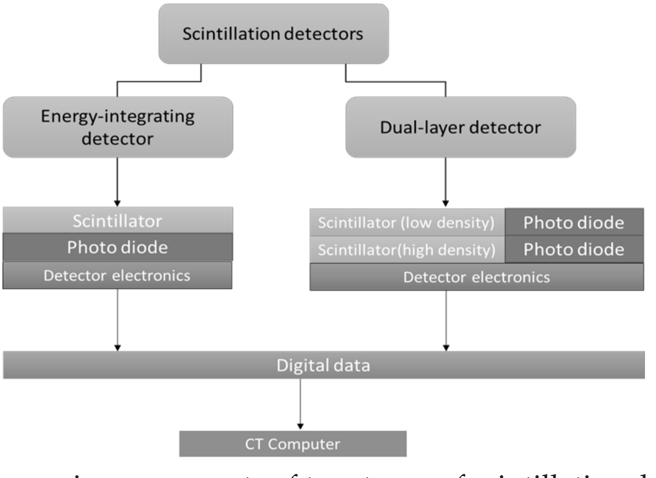

Figure 2. The main components of two types of scintillation detectors [12, 13,14, 15]

1.3 Dual-Energy Computed Tomography

Advancements in multidetector technology have made imaging possible for dual-energy computed tomography (CT). In 2006, a CT scanner was presented at the RSNA meeting mainly for cardiac imaging and other applications [16]. The latest improvements in multidetector CT (MDCT) capabilities include two-source and single-source designs, allow almost simultaneous image acquisition at mainly two kVps for dual-energy imaging, allowing the study of different tissue and material components inside a CT image's voxels [3]. The existing dual-energy CT system utilizes a higher kVp of 140 (labeled A) and a lower one of either 80 or 100 kVp (labeled B), each tube has a corresponding detector positioned on a rotating gantry opposite the tube. The radiation beams of the 2 tubes are parallel to each other at a 90-degree angle at the isocentre. The A tube has been associated with a maximum field of view (FOV) of 50 cm because of its detector size while the B tube has a maximum FOV of 26 cm in the first generation of the dual-source MDCT scanner.

CT imaging of the heart using MSCT scanners began in 1999, and due to the constant motion of the heart (heartbeat), temporal resolution is critical to avoid motion artifacts [17]. Additionally, it is critical to cover the entire heart with a single breath-hold when performing cardiac imaging with CT. Despite four-slice CT scanners have offered satisfactory results, issues with long breath-holding, artifacts motion, higher heart rates, and poor spatial resolution still persisted [18].

One of the issues with CT cardiac imaging is the elimination of the necessity for cardiac rate control, hence efforts are necessary to increase time resolution below 100 ms at any heart rate. Other scanners are needed to solve these issues, such as the electron beam CT. While this scanner has some advantages in imaging, it does not offer an acceptable signal-to-noise ratio in large patients. This scanner is therefore not currently deemed suitable for state-of-the-art cardiac CT imaging. To further increase the time resolution required by a factor of two another CT scanner for cardiac CT imaging, the DSCT scanner for cardiac imaging (developed by Siemens Medical Solutions), was termed the definition [14].

This review paper presents the history of CT scanner in terms of CT generations from single-slice to multi-slice scanners, with emphasis on the development of Multi-detector CT technology. The second section of this paper focus on the current different types of CT radiation dosimeters used for CT dose measurement including the pros and drawbacks of each CT radiation dosimeter. Besides, this article also reviews the CT radiation dose to patients and outlines the certain units and terminology that are relevant to the CT, and strategies for minimizing CT radiation dose.

2.0 METHODOLOGY

2.1 CT RADIATION DOSIMETRY

Several types of CT radiation dosimeters have been used previously for dose measurement in CT. These include standard ionization chambers, solid-state semiconductor detectors, radiographic film dosimeters, and thermoluminescent dosimeters (TLDs). In 1981, the Center for Devices and Radiological Health (CDRH) introduced the important step towards measuring the CT dose [19].

Conventional ionization chambers cannot be used to measure the absorbed dose as large beamwidths in CT scanners with multiple beam apertures are present. When a typical CT chamber (10cm) is used to determine the absorbed dose improperly, the dose profile for wide beams is likely to be underestimated [20], [21]. These patient dose problems can be easily addressed with a solid-state dosimetry CT dose profiler made by RTI Group Electronic, Sweden. However, to choose the sort of ionization chamber appropriate to measure CT exposure, two approaches must be considered. The first method is to use a very small volume ionization chamber and to measure a cross-section similar to that used in the TLD measurement [22]. This method has two significant drawbacks: it requires a large number of scans to produce a dose profile and the poor sensitivity of a small volume ionization chamber. The second method involves the use of a long pencil chamber to determine the cross-section of the x-ray field. On the other hand, a RaySafe X2 CT sensor (Unfors RaySafe, Billdal, Sweden) which can be considered as an ionization chamber has been used by many researchers [23].

This type of sensor has been proven to be a successful tool for determining the CT dose. It may be simply inserted into a head/body CT phantom or placed free-in-air. Pressure and temperature are not regulated manually because the sensor contains a method for correctly managing these two factors [5]. However, unlike other dosimeters, the RaySafe X2 CT sensor has the drawback of being unable to identify the CT dose profile [24]. In this situation, the radiographic film is typically utilized to determine the CT dose profile. Some advantages of the film include inexpensive cost, dose profile indication, reasonably quick working, and availability. Certain disadvantages of the film are qualitative measures, limited dynamic range, densitometry, and limited surface measuring use only [9]. To sum up, the features of an ionization chamber are its high sensitivity and dynamic range, quantitative measurement capabilities, the ability to detect both internal and external exposures, and immediate results. The disadvantages of an ionization chamber are that there is not enough information about the beam profile and the need for a particular chamber because of the X-ray field configuration [25].

Where CTDI is determined by utilizing a CT dose profile probe, the traditional five axial scans are replaced with a single helical scan within the central hole of the phantom. The CT dose profiler has replaced the conventional TLD and OSL methods or radiographic films for measuring the dose profile [26]. The CT dose profiler was designed to be used with a computer system with the Ocean Professional 2014 software. Besides, a narrow beamwidth (<10 mm) is used, the scattering dose of radiation is high beyond 100 mm. [27] studied the scattering index of CT dose when various input parameters were used. The scatter index values were found to be significantly affected by the size of the CTDI phantom and just minimally affected by the voltage applied [27], [28].

Thermoluminescence and semiconductor detector dosimeters are considered in this section. TLDs are available in a variety of forms (powder, chips, rods, and ribbons) and are composed of a variety of materials. While lithium fluoride doped with magnesium and titanium (LiF:Mg,Ti) is the most often used material in medical applications, additional materials such as LiF:Mg,Cu,P; Li2B4O7:Mn; CaSO4:Dy, and CaF2:Mn have also been utilized [28]. High sensitivity and dynamic range, quantitative measurement are the benefits of TLD as well as can be used to measure dose profile. The drawbacks of TLD are the need for an external system for heat stimulation, consuming the time for readout process, and high cost [29]. Certain clinical applications require solid-state semiconductor detectors.

The diode is the basic semiconducting detector, as it is based on a p–n junction between the semiconductor's p- and n-type components. Semiconductor dosimeters enable practical real-time measurements;

the small size of thermoluminescence dosimeters (TLDs) enables their use in patient measurements. Generally, the main limitation of these detectors has been their response energy dependence, which is significantly different from that of ionization chambers. These dosimeters have a variety of purposes, including postal audits and routine clinical assessments in hospitals [30]. The following Table 1.0 shows the comparison between CT radiation dosimetry systems.

Table 1. The comparison between CT radiation dosimeters [9], [20], [21], [24], [29], [30]

CT Radiation Dosimeter	Dosimetry Type	Principle of Operation	Limitation
Conventional Ionization Chamber	Real-time	Ion chamber	1- Cannot be used to measure the absorbed dose as large beamwidths in CT scanners with multiple beam apertures 2- Requires a large number of scans to produce a dose profile and the poor sensitivity of a small volume ionization chamber
CT Sensor from RaySafe X2	Real-time	Ion chamber	1- Unable to measure the width of CT dose profile
Radiographic Film	Passive	Radiophotoluminescence (RPL)	1- Qualitative measures 2- Limited dynamic range 3- Densitometry, and limited surface measuring use only
CT Dose Profile from Black Piranha	Real-time	Solid-state / Semiconductor	1- Ocean Professional Software License required more cost
Thermoluminescence Dosimeters	Passive	Thermoluminescence (TL)	1- Data erased during readout 2- Easy to lose reading 3- Accurate results require care 4- Readout and calibration time consuming

3.0 RESULTS AND DISCUSSION

3.1 CT RADIATION DOSE

CT scans are used in the medical industry, with several CT scans increasing from 3 million in 1980 to 62 million in 2007, alone in the USA. In 1992, the National Radiological Protection Board indicated patients with inappropriate examinations using computed tomography the possibility for high doses [31]. However, the fact that the patient is exposed to radiation by performing CT scans cannot be ignored. Research in the United States in 2009 showed that 75.4% of effective radiation doses from CT scans are around 7 times higher than X-rays. Lately, the CT was utilized in China to diagnose the discovered patient with the coronavirus because of the COVID-19 epidemic and it has been stated that the CT is highly sensitive and needs additional studies carefully [9].

It is necessary to understand certain units and terminology that are relevant to the CT include absorbed dose, effective dose equivalent, and CT dose index (CTDI), and dose length product (DLP) [11]. The absorbed dose is the actual concentration of the radiation dose to a certain organ or tissue measured in Grey (Gy).

The effective dose equivalent enables the conversion of a localized dose to a whole-body equivalent in terms of radiation consequences such as cancer risk. The effective dose equivalent can be calculated mathematically or using anthropomorphic phantoms, or utilizing CTDI or DLP and conversion factors [12], [13].

The CT dose index is measured in a single slice using cylindrical acrylic phantoms of a standard length. For the weighted CTDI (CTDI-w), dosimeters are inserted in the phantom center and periphery holes and the sum of them (weighted dose) is expressed in mGy [8], [31]. This value does not accurately represent the dose contribution of all parameters in helical scanning. For some of this contribution, the volume CTDI or CTDI-vol has lately become more acceptable and considers the contribution from the pitch to CTDI-w. The higher the pitch, the lower the CTDI-vol if all other parameters are constant. The dose length product in a unit (mGy.cm), is calculated as the product of CTDI and scan length, and hence increases proportionally with scan distance covered [1].

3.2 CT DOSE OPTIMIZATION

CT scanner design and performance advancements have contributed to the increased usage of CT in clinical medicine. Numerous studies indicate that CT delivers greater doses to patients than other modalities [32]. For instance, CT doses to individual organs within the image field generally range between 5 and 50 mGy. As of 2011, CT accounted for the most cumulative quantity of medical radiation exposure in the United States, exceeding all other imaging modalities [33]. Paediatric CT examinations with high doses have drawn attention to the cancer risks involved with CT scanning. There has been an increased emphasis on minimizing the dose to the patient without reducing the image quality required for diagnosis [9], [31], [34]. The International Commission on Radiation Protection (ICRP) defines optimization as reducing radiation doses "as low as reasonably achievable" (ALARA) without affecting an image's diagnostic quality. Hence, optimization considers both radiation dose and image quality.

Several parameters are affecting the dose of the CT patient. These include the scan parameters such as exposure technique factors (mAs and kVp), patient cantering, automatic tube current modulation, collimation, pitch, number of detectors, over-ranging, and iterative image reconstruction [35]. To lower the dose successfully and achieve the required image quality, users need to have systematic procedures or techniques to optimize CT dose. Several researchers have reported different ways of dose optimization in CT, particularly in multi-slice CT [36]–[38]. [31] reviewed optimizing the CT dose and summarised the important items to be considered.

Dose optimization in paediatric CT has gained increasing attention. For instance, [39], [40] provided great overviews on this subject, not only analysing trends and patterns of CT use, CT radiation issues, and how radiologists might control CT dose. They also concentrated on technical aspects of dose management, such as the adjustment of tube current and voltage, the impact of gantry cycle time, and the selection of pitch and detector width. [39] highlighted the significance of the radiologists to confirm that all requested examinations are justified and to ensure that communication is essential as a first step to the reduction of the CT dose between the requesting physicians and radiologists. In addition, an important study in CT pulmonary angiography has shown that the patient's dose is substantially decreased by changing the peak kilovoltage from 120 kVp to 100 kVp with no loss of objective or subjective quality of the image [17].

4.0 CONCLUSION

This review article presented an overview of CT generations from single-slice to multi-slice scanners, with emphasis on the development of Multi-detector CT technology. Many previous research papers are summarised in this review article and mainly focused on the various types of CT radiation dosimeters for CT dose measurement including the advantages and disadvantages of each CT radiation dosimeter. Furthermore, this article reviews the subject of CT radiation dose to patients and outlined the certain units and terminology that are relevant to the CT, and strategies for minimizing CT radiation dose.

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