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CTDI AND DLP MEASUREMENTS IN ESTONIAN COMPUTED TOMOGRAPHY CABINETS

Master Thesis in Applied Physics

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1. CONTENTS

2. TERMS

CT – Computed Tomography

$CTDI_w$ – CT Dose Index, it is the approximation of average dose over a single slice in the standard head or body CT dosimetry phantom, expressed in terms of absorbed dose to air (mGy)

DLP – Dose Length Product, characterizes exposure for a complete examination in relation to linear integration of the dose to the standard head or body CT dosimetry phantom on the basis of absorbed dose to air (mGy cm)

DRL – Diagnostic Reference Dose values are indicated for two dose descriptors *weighted CTDI* ($CTDI_w$) and *dose-length product* (DLP) on the basis of absorbed dose to air, in relation to technique for a standard-sized patient.

DSCT – Double-Source Computed Tomography

EBCT – Electron Beam Computed Tomography**,** same as EBT

EC – European Commission

EG – European Guidelines Quality Criteria for Computed Tomography', European Commission Report

FOV – Field Of View, the maximum diameter of the reconstructed image

HU – Hounsfield Units, measured values of attenuation are transformed into CT numbers using the international Hounsfield scale

MDCT – Multi-Detector-row Computed Tomography systems, same as MSCT

Monte Carlo technique – a technique for obtaining an approximate solution to certain mathematical and physical problems, characteristically involving the replacement of a probability distribution by sample values, usually performed using a computer

MSCT – Multi-Slice Computed Tomography, same as MDCT

NRPB – National Radiation Protection Board, UK

PMMA – Polymethyl Methacrylate (PMMA) or Polymethyl 2-Methylpropenoate is the synthetic polymer of methyl methacrylate. Usually, the basic compound for CTDI Head and Body phantoms

Pitch factor – in relation to helical CT, ratio of the patient couch travel in horizontal direction per rotation of the x-ray tube divided by the product of the number of tomographic sections produced by a single rotation of the x-ray tube N times the nominal tomographic slice thickness T

RDL – Reference Dose Level, basically same as **DRL**

ROI – Region Of Interest; localized part of an image defined by the operator, which is of particular interest at a given time

SSCT – Single Slice Computed Tomography; same as axial tomography

SSI – Statents Strålskyddsinstitut, Swedish Radiation Protection Authority

TLD – Thermoluminescent Dosimeter

3. INTRODUCTION

In worldwide medical diagnostic radiology practices CT is related to high radiation dose to the patient and contribution of CT examinations to collective dose from medical X-ray is continuously growing [1]. Since the introduction of computed tomography (CT) in the 1970s, it has been known that CT leads to high radiation dose to the patient. CT causes the largest amount (ie, 35%–40%) of the collective effective dose of x-rays owing to diagnostic procedures [2][3].

Medical x-ray diagnostics is an indispensable practice for patients to be taken care of in an adequate way. The benefit for the patient is normally much larger than the negative aspects as e.g. the radiation risk. This doesn't imply that the radiation risk can be neglected – the patients have the right that this risk is minimized as far as possible, i.e. that the examination is optimized such that the radiation dose and with that the radiation risk is not higher than necessary to achieve the intended diagnostic result. Since 1994 formal requirements exist in SSI's (Sweden) regulations that all examinations must be justified and optimized [4].

The International Commission on Radiological Protection recommended the establishment of investigation levels for radiology procedures; for higher levels, the cause or the implications of the result should be examined [1]. The Council Directive of June 30, 1997, requires the member states of the European Community to promote the establishment and use of *diagnostic reference levels* that are expected not to be exceeded for standard procedures [5].

The Swedish Radiation Protection Authority brings definitions both for *diagnostic reference level (RDL)* and for *diagnostic standard dose (DSD)*. RDL is a value which has the meaning of a dose level established for a certain type of examination, which, if exceeded, shall lead to an action. DSD has the meaning of a typical radiation dose for a certain type of examination, confirmed by the license holder and determined in the same way as applicable for the corresponding diagnostic reference level [6].

The European Commission suggests reference doses in which the weighted CT dose index and dose-length product are used for various CT examinations. However, these dose values are based on the results of older survey data from the late 1980s.

The technical improvements in CT, in particular use of the spiral technique, have offered new possibilities in both diagnostics and dose reduction [7][8]. The tube current time product for spiral CT usually cannot be set as high as for conventional CT owing to the limited tube heat capacity; therefore, the radiation dose should be effectively lower for spiral than for conventional CT [2][8]. The results of older surveys that were based on investigations of dose for conventional CT may not be representative of the present situation.

A number of surveys have shown that the patient doses for the same type of examination vary very much between different hospitals and examination stands. There are reasons to suspect that where the highest doses are found deficiencies exist with respect to optimization. That is where the concept of diagnostic reference levels fits in, the major purpose of which is to identify and if possible to take measures against the presence of high patient doses. In this way it is expected that the patient doses and the spread between them will be reduced on a long view [4].

When investigating high doses an analysis of these data may provide direct indications on the cause for the high dose and on suitable measures. Also when comparing different CT modalities in the same or in different hospitals these data can provide information on which possible improvements could be performed for reducing the patient dose.

Doses for children are always the matter of special consideration, especially in computed tomography, where high dose levels are usually expected. In literature reported that effective doses to very young patients (aged 0-1 years) were typically higher than corresponding for values for adults [9]. This is a very serious topic, which will be covered in the scope of our future investigations. Unfortunately, in CT cabinets that we have observed, there were a very small amount of CT examinations of young patients aged less than 16 years old. For example, during four consecutive months, in CT cabinet of Western-Tallinn Central Hospital, there have been made 2099 CT investigations with only 6 investigations made on under 16 year old patients. However, all CT scanners had specific protocols for children CT investigations and at some scanners those protocols were smartly differentiated depending on age of young patients ranging from newborn to 16 years old patient's protocol. It is expected, that measurement of doses and comparison of children CT protocols in Estonian CT cabinets

will be the next challenge for the author of this paper, and of course, along with the aim of gathering dose and CT protocols data from all CT scanners located in Estonia.

Currently, in September, 2006, there were 10 CT scanners installations in Estonia, all installations were concentrated in 4 towns – Tallinn, Tartu, Pärnu, Kohtla-Järve - and consequently among 6 healthcare institutions three of them were located in Tallinn. As on 31.03.2006 there were 1 273 961 patients registered in Estonian Health Insurance Fund [10], that means presence of 1 CT scanner per about 130 thousand persons. This is a good number in worldwide practice. For comparison, in UK, there are 471 CT scanner installations providing its services for population of over 60.2 mln people [9]. This is approximately 1 scanner per almost 130 thousand people. At the present time, in Estonia there are 2 older (with manufacture year under 1999) so called single-slice installations with classic axial technology, 8 modern scanners (manufactured in 2001-2006) are equipped with so called helical technique whereas one is an ultramodern device contains 2 x-ray sources – Dual-Source CT.

In this paper presented the results of CT radiation exposure survey performed in 2003-2006 within different CT examinations and techniques at various hospitals and practices in Estonia (see Table 3 for details). Some intermediate results with measurements made for three CT scanners, were presented by the author on International Conference - EMBEC'2005, which took place in November 2005 in Prague, Czech Republic[11].

4. HISTORY AND TECHNOLOGY OF COMPUTED TOMOGRAPHY

Definition

Computed tomography (CT), originally known as *computed axial tomography* (CAT or *CT scan*) and *body section roentgenography*, is a medical imaging method employing tomography where digital geometry processing is used to generate a three-dimensional image of the internals of an object from a large series of two-dimensional x-ray images taken around a single axis of rotation. The word "*tomography*" is derived from the Greek *tomos* (slice) and *graphia* (describing). CT produces a volume of data which can be manipulated, through a process known as *windowing*, in order to demonstrate various structures based on their ability to block the x-ray beam. Although historically (see below) the images generated were in the axial or transverse plane (orthogonal to the long axis of the body), modern scanners allow this volume of data to be reformatted in various planes or even as volumetric (3D) representations of structures [12]. Although most common in healthcare, CT is also used in other fields, e.g. nondestructive testing of materials.

Diagnostic use

Since its introduction in the 1970s, CT has become an important tool in medical imaging to supplement conventional radiography and medical ultrasonography. Although it is still quite expensive, it is the gold standard in the diagnosis of a large number of different disease entities.

Advantages of CT technology

Briefly about advantages over Projection Radiography or, in other words, over conventional x-ray technology. First, CT completely eliminates the superimposition of images of structures outside the area of interest. Second, because of the inherent high-contrast resolution of CT, differences between tissues that differ in physical density by less than 1% can be distinguished. Third, data from a single CT imaging procedure consisting of either multiple contiguous or one helical scan can be viewed as images in the axial, coronal, or sagittal planes, depending on the diagnostic task. This is referred to as multiplanar reformatted imaging [12].

Radiation exposure

CT is regarded as a moderate to high radiation diagnostic technique. While technical advances have improved radiation efficiency, there has been simultaneous pressure to increase radiation dose with higher-resolution imaging, and more complex scan techniques. The improved resolution of CT has permitted the development of new investigations, which may have advantages; e.g. compared to conventional angiography, CT angiography avoids the invasive insertion of an arterial catheter and guidewire; CT colonography may be as good as barium enema for detection of tumors, but may use a lower radiation dose [12].

The greatly increased availability of CT, together with its value for an increasing number of conditions, has been responsible for a large rise in popularity. So large has been this rise that, in the most recent comprehensive survey in the UK, CT scans constituted 7% of all radiologic examinations, but contributed 47% of the total collective dose from medical X-ray examinations in 2000/2001 [13]. Increased CT usage has led to an overall rise in the total amount of medical radiation used, despite reductions in other areas.

The radiation dose for a particular study depends on multiple factors: volume scanned, patient build, number and type of scan sequences, and desired resolution and image quality.

Method

X-ray slice data is generated using an X-ray source that rotates around the object; X-ray sensors are positioned on the opposite side of the circle from the X-ray source. Many data scans are progressively taken as the object is gradually passed through the gantry. They are combined together by the mathematical procedure known as tomographic reconstruction.

Newer machines with faster computer systems and newer software strategies can process not only individual cross sections but continuously changing cross sections as the gantry, with the object to be imaged, is slowly and smoothly slid through the X-ray circle. These are called *helical* or *spiral CT* machines. Their computer systems integrate the data of the moving individual slices to generate three dimensional volumetric information (3D-CT scan), in turn viewable from multiple different perspectives on attached CT workstation monitors.

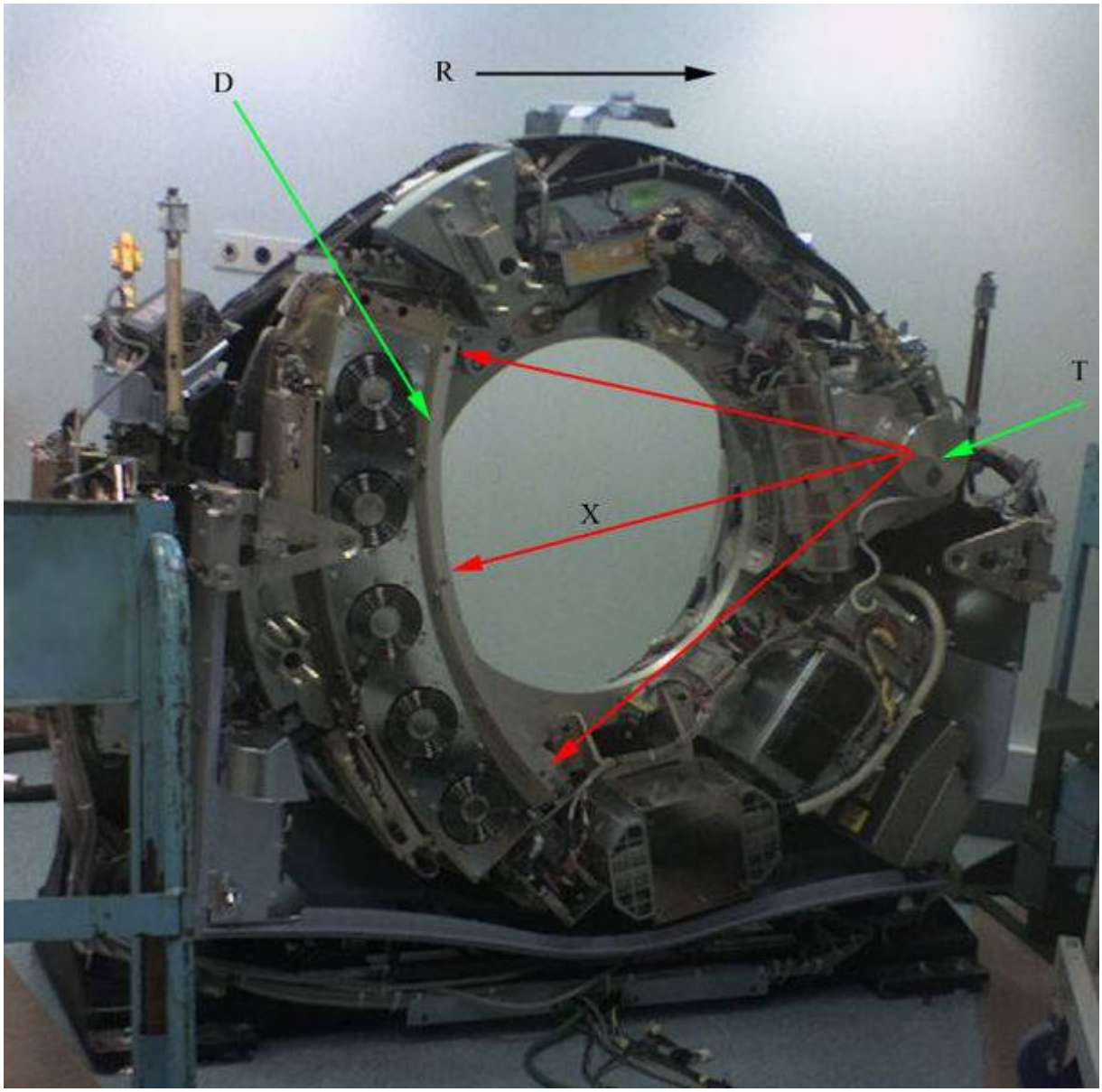


Figure 1. CT scanner with cover removed to show the principle of operation: *T* - x-ray tube, *D* - x-ray detectors, *X* - x-ray beam, *R* - Gantry rotation direction [12].

In conventional CT machines, an x-ray tube and detector are physically rotated behind a circular shroud (see the image); in the electron beam tomography (EBT) the tube is far larger and higher power to support the high temporal resolution. The electron beam is deflected in a hollow funnel shaped vacuum chamber. Xray is generated when the beam hits a stationary target. The detector is also stationary.

The data stream representing the varying radiographic intensity sensed reaching the detectors on the opposite side of the circle during each sweep is then computer

processed to calculate cross-sectional estimations of the radiographic density, expressed in Hounsfield units. Sweeps cover 360 or just over 180 degrees in conventional machines, 220 degrees in EBT [12].

CT is used in medicine as a diagnostic tool and as a guide for interventional procedures. Sometimes contrast materials such as intravenous iodinated contrast are used. This is useful to highlight structures such as blood vessels that otherwise would be difficult to delineate from their surroundings. Using contrast material can also help to obtain functional information about tissues.

Pixels in an image obtained by CT scanning are displayed in terms of relative radiodensity. The pixel itself is displayed according to the mean attenuation of the tissue(s) that it corresponds to on a scale from -1024 to +3071 on the Hounsfield scale. Pixel is a two dimensional unit based on the matrix size and the field of view. When the CT slice thickness is also factored in, the unit is known as a Voxel, which is a three dimensional unit. The phenomenon that one part of the detector can not differ between different tissues is called the *Partial Volume Effect*. That means that a big amount of cartilage and a thin layer of compact bone can cause the same attenuation in a voxel as hyperdense cartilage alone. Water has an attenuation of 0 Hounsfield units (HU) while air is -1000 HU, cancellous bone is typically +400 HU, cranial bone can reach 2000 HU or more (os temporale) and can cause artefacts. The attenuation of metallic implants depends on atomic number of the element used: Titanium usually has an amount of +1000 HU, iron steel can completely extinguish the X-ray and is therefore responsible for well-known line-artefacts in computed tomogrammes [12].

From the history of computed tomography

The first commercially viable CT system was invented by Godfrey Newbold Hounsfield in Hayes, England at THORN EMI Central Research Laboratories using X-rays. Hounsfield conceived his idea in 1967, and it was publicly announced in 1972. It is claimed that the CT scanner was "the greatest legacy" of the Beatles; the massive profits from their record sales enabled EMI to fund scientific research. Allan McLeod Cormack of Tufts University independently invented a similar process at the University of Cape Town/Groote Schuur Hospital and they shared a Nobel Prize in medicine in 1979 [12].

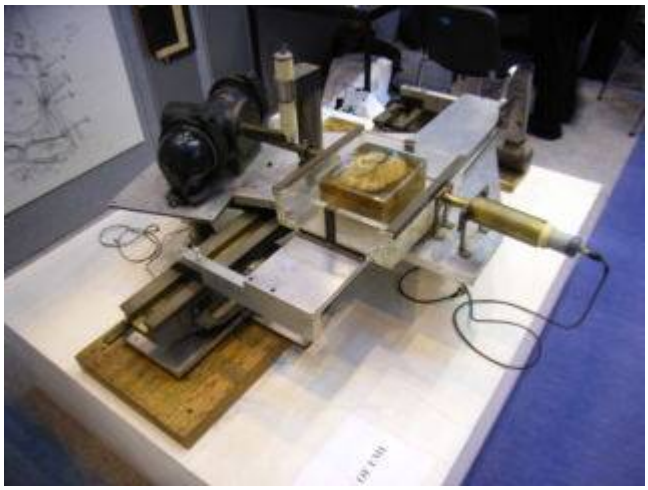


Figure 2. The very first CT scanner prototype. Invented by Hounsfield at EMI [12].

The original 1971 prototype of CT scanner took 160 parallel readings through 180 angles, each 1° apart, with each scan taking a little over five minutes. The images from these scans took 2.5 hours to be processed by algebraic reconstruction techniques on a large computer [12].

The first production of CT machine (called the EMI-Scanner) was limited to making tomographic sections of the brain, but acquired the image data in about 4 minutes (scanning two adjacent slices) and the computation time (using a Data General Nova minicomputer) was about 7 minutes per picture. This scanner required the use of a water-filled Perspex tank with a pre-shaped rubber "head-cap" at the front, which enclosed the patient's head. The water-tank was used to reduce the dynamic range of the radiation reaching the detectors (between scanning outside the head compared with scanning through the bone of the skull). The images were relatively low resolution, being composed of a matrix of only 80 x 80 pixels. The first EMI-Scanner was installed in Atkinson Morley's Hospital in Wimbledon, England, and the first patient brain-scan was made with it in 1972 [12]..

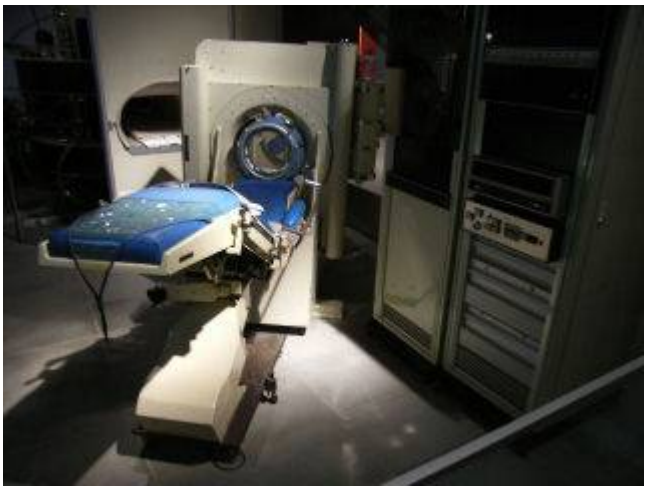


Figure 3. A historic EMI-Scanner [12]

In the US, the machine sold for about $390,000, with the first installations being at the Lahey Clinic, then Massachusetts General Hospital, and George Washington University in 1973 [12].

The first CT system that could make images of any part of the body, and did not require the "water tank" was the ACTA scanner designed by Robert S. Ledley, DDS at Georgetown University [12].

CT technology generations

Here are outlined CT technology basic achievements, presented between 1972 and 1988 years. In 1990's CT became affordable, efficient, and indispensable diagnostic tool found its place in worldwide medical practice, and a new era of CT began offering smarter acquisition solution based mainly on third generation of CT scanners – MSCT, MDCT, DSCT, Volumetric CT and others.

First generation. These CT scanners used a pencil-thin beam of radiation directed at one or two detectors. The images were acquired by a "translate-rotate" method in which the x-ray source and the detector in a fixed relative position move across the patient followed by a rotation of the x-ray source/detector combination (gantry) by one degree. In the EMI-Scanner, a pair of images was acquired in about 4 minutes with the gantry rotating a total of 180 degrees. Three detectors were used (one of these being an X-ray source reference), each detector comprising a sodium iodide scintillator and a photomultiplier tube. Some patients had unpleasant experiences within these early scanners, due to the loud sounds and vibrations from the equipment.

Second generation. This design increased the number of detectors and changed the shape of the radiation beam. The x-ray source changed from the pencil-thin beam to a fan shaped beam. The "translate-rotate" method was still used but there was a significant decrease in scanning time. Rotation was increased from one degree to thirty degrees.

Third generation. CT scanners made a dramatic change in the speed at which images could be obtained. In the third generation a fan shaped beam of x-rays is directed to an array of detectors that are fixed in position relative to the x-ray source. This eliminated the time consuming translation stage allowing scan time to be reduced, initially, to 10 seconds per slice. This advance dramatically improved the practicality of CT. Scan times became short enough to image the lungs or the abdomen; previous generations had been limited to the head, or to limbs. Patients have reported more pleasant experiences with the third and fourth generation CT scanners because of greatly reduced noise and vibration compared to earlier models.

Fourth generation. This design was introduced, roughly simultaneously with 3rd generation, and gave approximately equal performance. Instead of a row of detectors, which moved with the X-ray source, 4th generation scanners used a stationary 360 degree ring of detectors. The fan shaped x-ray beam rotated around the patient directed at detectors in a non-fixed relationship.

Bulky, expensive and fragile photomultiplier tubes gradually gave way to improved detectors. A xenon gas ionization chamber detector array was developed for third generation scanners, which provided greater resolution and sensitivity. Eventually, both of these technologies were replaced with solid-state detectors: rectangular, solid-state photodiodes, coated with a fluorescent rare earth phosphor. Solid state detectors were smaller, more sensitive and more stable, and were suitable for 3rd and 4th generation designs.

On an early 4th generation scanner, 600 photomultiplier tubes, about 12 mm in diameter, could fit in the detector ring. Three photodiode units could replace one photomultiplier tube. This change resulted in increasing both the acquisition speed, and

image resolution. The method of scanning was still slow, because the X-ray tube and control components interfaced by cable, limiting the scan frame rotation.

Initially, 4th generation scanners carried a significant advantage - the detectors could be automatically calibrated on every scan. The fixed geometry of 3rd generation scanners was especially sensitive to detector mis-calibration (causing ring artifacts). Additionally, because the detectors were subject to movement and vibration, their calibration could drift significantly.

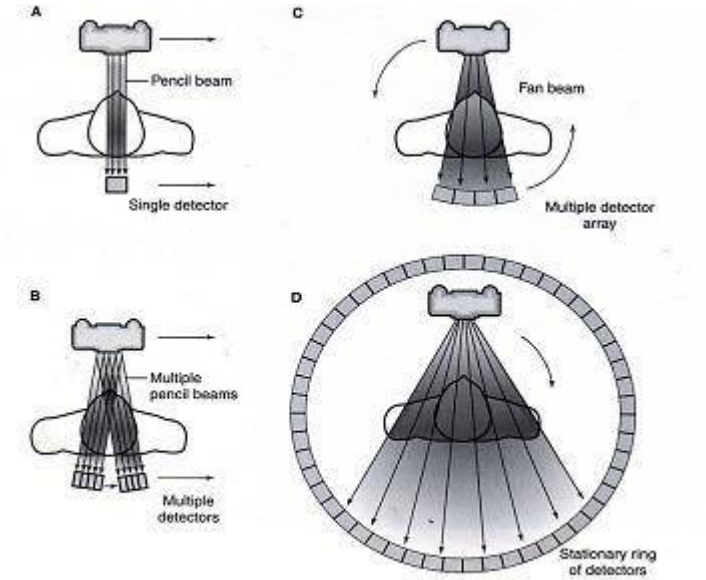


Figure 4. Four generations of CT technology [14].

All modern medical scanners are of 3rd generation design. Modern solid-state detectors are sufficiently stable that calibration for each image is no longer required. The 4th generation scanners' inefficient use of detectors made them considerably more expensive than 3rd generation scanners. Further, they were more sensitive to artifacts because the non-fixed relationship to the x-ray source made it impossible to reject scattered radiation.

Further advances

Another limiting factor in image acquisition was the X-ray tube. The need for long, high intensity exposures and very stable output placed enormous demands on both the tube and generator (power supply). Very high performance rotating anode tubes were developed to keep up with demand for faster imaging, as were the regulated 150 kV

switched mode power supplies to drive them. Modern systems have power ratings up to 100 kW.

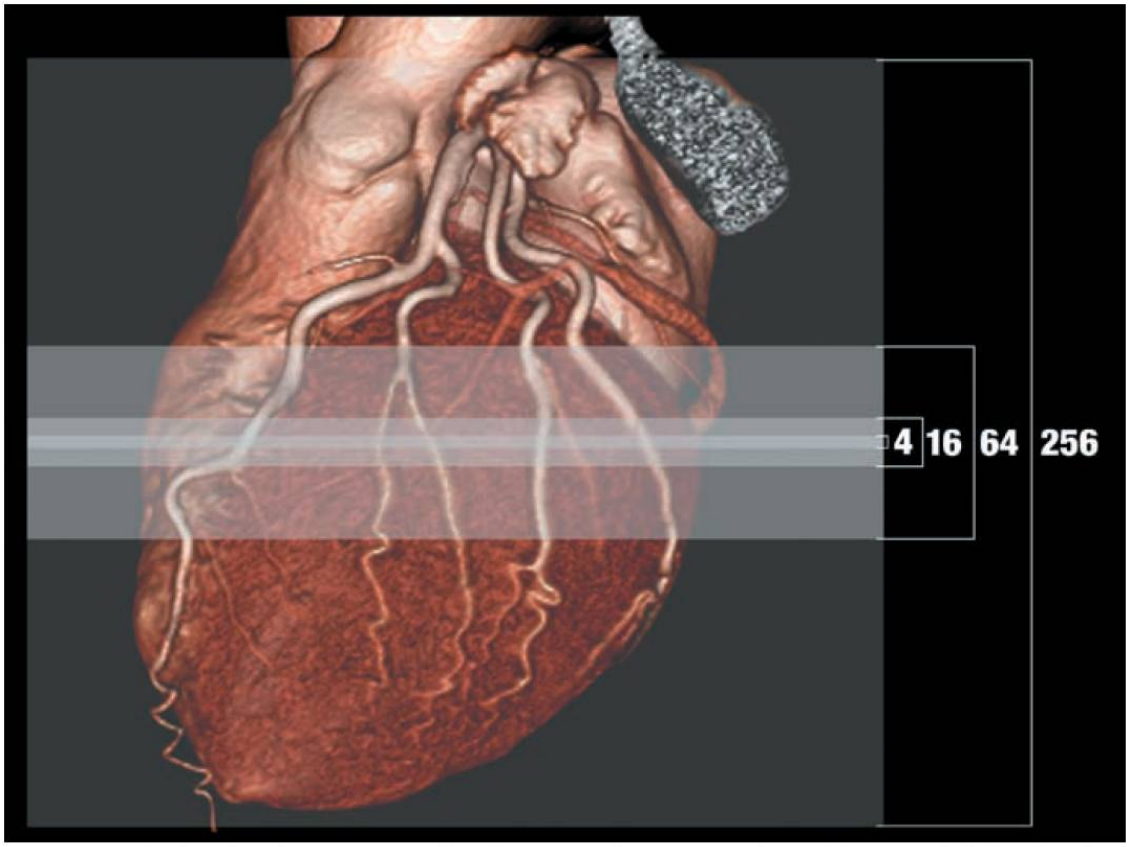
In 1989, Spiral CT Scanning, also known as Helical CT scanning or Volume CT Scanning, was introduced. Slip-ring technology replaced the spooled cable technology of older CT scanners, allowing the X-ray tube and detectors to spin continuously. When combined with the ability to move the patient continuously through the scanner this refinement is called *Volume CT*, *Spiral CT*, or, more commonly, *Helical CT*.

Multi-detector-row CT systems further accelerated scans, by allowing several images to be acquired simultaneously. Modern scanners are available with up to 64 detector rows / output channels (depends upon the technology used by the manufacturer). It is possible to complete a scan of the chest in a few seconds. An examination that required 10 separate breath-holds of 10 seconds each can now be completed in a single 10-second breath-hold. Multi-detector CT can also provide isotropic resolution, permitting cross-sectional images to be reconstructed in arbitrary planes; an ability similar to MRI. More anatomical volume coverage in less time is one of the key features of the latest generation MD CT Scanners. It is however more important to achieve better spatial resolution than only volume coverage for better reconstructed images. Latest generation MD CT scanners with flying X-ray tube focal spot in z-axis direction show better image resolution.

In 1998 the first four slice scanners were introduced. Rapid development followed with more advanced systems; 16 slice scanners in 2001, followed by 32 and 40 slice scanners and the development of 64 slice scanners in 2003 [15]. With the advances in increased number of slices, rotation speed was also improved, dropping from 1 second to approximately 0.375 seconds per rotation [15].

The advantages of MSCT can be summed up by the acronym RSVP [15]:

- Resolution: spatial resolution improved along the z-axis
- Speed: scanning time reduced for particular body regions
- Volume: increased volume able to be scanned
- Power: improved power usage (X-ray tube).



From 4-16-64-slice systems, the breath-hold time for helical scanning of the heart has been significantly reduced. However, it is only with the 256-slice system that the entire heart can be covered in a single rotation.

Figure 5. Volumetric 256-slice MSCT capabilities in comparison to 4-16-64-slice techniques [16].

Volumetric CT is an extension of multi-detector CT, currently at research stage. Current MDCT scanners sample about 4 cm wide volumes in one rotation. Volumetric CT aims to increase the scan width to 10-20 cm, with current prototypes using 256 detector-rows [16]. Potential applications include cardiac imaging (a complete 3D dataset could be acquired in the time between 2 successive beats) and 3D cine-angiography, see Figure 5.

Fifth generation CT. A different approach was used for a particular type of dedicated cardiac CT technique called electron-beam CT (also known as *ultrafast* CT or *Electron Beam Tomography - EBT*, and occasionally *fifth generation* CT). With temporal resolution of approximately 50 ms, these scanners could "freeze" cardiac and pulmonary motion providing high quality images. Only one manufacturer offered these scanners (Imatron, later GE healthcare), and few of these scanners were ever installed, primarily due to the very high cost of the equipment and their single-purpose design. Rapid development of MDCT has significantly reduced the advantage of EBCT over

conventional systems. Contemporary MDCT systems have temporal resolution approaching that of EBCT, but at lower cost and with much higher flexibility. Because of this, MDCT is usually the preferred choice for new installations [12].

Dual source CT. Improved computer technology and reconstruction algorithms have permitted faster and more accurate reconstruction. On early scanners reconstruction could take several minutes per image, a modern scanner can reconstruct a 1000 image study in under 30 seconds. Refinements to the algorithms have reduced artifacts. *Dual source* CT uses 2 x-ray sources and 2 detector arrays offset at 90 degrees. This reduces the time to acquire each image to about 0.1 seconds, making it possible to obtain high quality images of the heart without the need for heart rate lowering drugs such as beta blockers. A dual-source multi-detector row scanner can complete an entire cardiac study within a single 10 second breath hold.

Microtomography

In recent years, tomography has also been introduced on the micrometer level and is named Microtomography. But these machines are currently only fit for smaller objects or animals, and cannot yet be used on humans.

5. CT DOSE MEASUREMENT. THEORETICAL BACKGROUND

There are many ways to describe and measure radiation dose in CT [3][8]. Recently, European Guidelines (EG) on quality criteria for CT were published by the European Commission (EC) [5], where two dose descriptors, weighted computed tomography dose index CTDI and dose–length product DLP, were proposed as reference dose levels (Diagnostic Reference Levels - DRLs).

CTDI

CTDI is defined as the integral over the dose profile (for one revolution of the x-ray tube) along a line parallel with the axis of rotation divided with the nominal slice thickness d [4].

$$CTDI = \frac{1}{d}\int_{-a}^{+a} D(z)\cdot dz \qquad \text{(mGy)} \qquad (1)$$

D(z) = air kerma along a line parallel with the axis of rotation
d = nominal slice thickness

In the literature different integration limits, dose quantities and phantom materials are used [11]. Here we use the integration limit ± 50 mm, the dose quantity air kerma and a phantom of PMMA with a diameter of 160 mm and 320 mm for simulation of the head and the trunk, respectively. The direct measured dose quantity is

$$CTDI_{100} = \frac{1}{d}\int_{-50mm}^{+50mm} D(z)\cdot dz \qquad \text{(mGy)} \qquad (2)$$

The average value for $CTDI_{100}$ in a volume in the phantom with the thickness d is called the weighted CTDI or $CTDI_w$ and is calculated according to

$$CTDI_w = (\frac{1}{3}CTDI_c) + (\frac{2}{3}CTDI_p) \qquad \text{(mGy)} \qquad (3)$$

where $CTDI_c$ and $CTDI_p$ represent CT dose index measurements made with an ionization chamber at the centre and periphery of the phantom, 10 mm below the surface of the phantom, respectively [18].

The equations (2) and (3) are valid for axial scans. For the calculation of the average absorbed dose in the irradiated volume from a scan series, $CTDI_{vol}$, $CTDI_w$ and also the pitch factor is needed. The pitch factor *p* is equal with the ratio between the patient table movement between two consecutive revolutions of the x-ray tube and the nominal slice thickness, a definition that is applied here both for conventional axial and for helical technique. $CTDI_{vol}$ is calculated according to [4]:

$$CTDI_{vol} = \frac{1}{p} \cdot CTDI_w \qquad \text{(mGy)} \qquad (4)$$

DLP

The quantity dose-length product (DLP) which includes the volume of patient (phantom) irradiated in the course of complete examination, is defined as:

$$DLP = \sum_i CTDI_w \cdot T_i N_i \qquad \text{(mGy·cm)} \qquad (5)$$

where *i* represents each helical scan sequence forming part of the examination, where T_i is each different slice thickness used in the examination protocol, N_i is the number of T_i slices and $CTDI_w$ is the value of $CTDI_w$ of each particular slice thickness T_i. This translates into a similar equation for helical scanning:

$$DLP = \sum_i CTDI_w \cdot T \qquad \text{(mGy·cm)} \qquad (6)$$

where *i* now represents each helical scan sequence forming part of the examination, T is the nominal irradiated slice thickness in cm [18].

Normalized CTDI

The EC have suggested use of *normalized weighted CT dose index*, ${}_nCTDI_w$, which is expressed as absorbed integral along a line parallel to the axis of rotation z of the dose

Table 1. CEC 1998 CT quality criteria [5].

Examination	**Reference dose value**	
	$CTDI_w$ (mGy)	DLP (mGy cm)
Routine head	60	1050
Routine chest	30	650
Routine abdomen	35	800
Routine pelvis	35	600

profile D(z) of a single slice, divided by the nominal slice thickness T [5].

The actual $CTDI_w$ is obtained by multiplying with the C (mAs) value used in the hospital and provides the radiation dose from one slice at particular exposure settings:

$$CTDI_w = {}_n CTDI_w \cdot C \qquad \text{(mGy)} \qquad (7)$$

Comparison of both $CTDI_w$ and DLP values for a specific examination using different scanners and protocols will provide information on relative performance [19].

For comparison with EU standards, we used CEC 1998 quality criteria - region-specific normalized coefficients to calculate the risk of a particular examination protocol and to compare it with other CT protocols or different radiological examinations. The purpose of this study was to investigate routine examination protocols utilized in CTs in some largest hospitals of Estonia in terms of imaging technique and radiation dose and to compare results with European Commission reference dose levels (EC RDLs) [5]

Assessment of effective dose *E*

In addition to comparison of performance against reference dose values, we are going to assess effective dose for CT procedures so as, for example, to allow comparison with

other types of radiological examination like it was made by some other authors from EU countries [20][21].

The effective dose E is a quantity direct correlated with the radiation risk. For a normal population this risk is taken as 5×10^{-5} radiation induced cancer deaths for an effective dose of 1 mSv [4]. For an accurate calculation of E detailed knowledge is needed regarding the irradiation geometry, the beam quality and the patient's anatomy, which is mostly difficult, not to say impossible to achieve. The purpose here of estimating the radiation risk is more to estimate the relative risk between examinations and different techniques or procedures than to assess the absolute risk for individual patient. For this purpose it is fully satisfactory to use generalized standard calculations with established conversion factors [6].

The effective dose for a particular scanning protocol may be derived from values of DLP for an examination using appropriately normalized coefficients:

$$E = E_{DLP} \cdot DLP \qquad \text{(mSv)} \qquad (8)$$

where DLP (mGy cm) is the dose-length product as defined in Equations (2) or (3) and E_{DLP} is the region-specific normalized effective dose ($mSv \cdot mGy^{-1} \cdot cm^{-1}$).

General values of E_{DLP} appropriate to different anatomical regions of the patient (head, neck, chest, abdomen or pelvis) are given in Table 2.

Table 2. Normalized values of effective dose per dose-length product (DLP) over various body regions brought by EC , 1997 [5] and Clarke et al.[22]

	EC	Clarke et al. – Monte Carlo method
Head	0.0023	0.0021
Neck	0.0054	0.0048
Chest	0.017	0.014
Abdomen	0.015	0.012
Pelvis	0.019	0.016

Some authors (Clarke et al.,[22]) use dosimetric calculations based on organ dose data calculated using the Monte-Carlo method for a mathematical anthropomorphic phantom. In this work, however, for deriving effective dose it were decided to apply

values brought in European Guidelines on Quality Criteria for Computed Tomography (EG).

6. MATERIALS AND METHODS

In order to compare the standard doses to values shown by other authors or guidelines, the measured standard doses must be comparable with each other, i.e. must have been derived with the same methods and with sufficient accuracy. Therefore it is important to follow as far as possible the same procedures as described in existing regulations and that the (dose-)measuring instruments are calibrated in a satisfactory way. Next will be described methods and equipment used for measuring standard doses.

CTDI phantoms

The head phantom we use is a cylindrical (16 cm diameter, 14 cm length) solid tissue equivalent polyethylene (PE) phantom with five 13.1 mm diameter holes drilled parallel to its long axis, one at the axial centre and four around the perimeter, 90° apart and 1 cm from the edge. Each of the holes used to be plugged with a cylindrical solid PE rod. The body phantom is similar to the head phantom, with a 32 cm diameter, 14 cm length. After all in order to check for compliance with CT dose criteria, $CTDI_w$ and DLP values were then calculated according to equations (1), (2), (3). See results in Table 7, Table 8.

Phantoms comparison tests

A few words about preparing of CTDI Head and Body phantoms utilized in examinations. First we planned to use standard PMMA phantoms supplied by some well-known vendor (RTI, PTW, RadCal etc.), but due to circumstances, it was decided to manufacture own phantoms. There was found a plastics tools manufacture that proposed to use a polyethylene compound bulk, which should be similar to PMMA in terms of radiophysical characteristics such as x-ray attenuation and effective atomic number. Finally the phantom was manufactured but its chemical formula was still unclear.

Some authors reported that polyethylene and polyurethane resin with additives are better compounds to be used in CTDI phantoms due to its water-equivalent physical properties which make it the best choice for precise comparisons of scanner models while PMMA would be the better choice for acceptance tests and QA (Quality Assurance) [23]. This fact encouraged the authors despite of phantom's unclear chemical composition

CTDI phantoms comparison (standard PMMA phantoms compared to our polyethylene (PE) phantoms) tests were carried out in the Radiology Department of Karolinska University Hospital in Huddinge (Stockholm).

Figure 6. CTDI phantoms: a) Original reference Head and Body PMMA phantom set from RTI; b) self-made PE Body phantom and the RTI Barracuda electrometer with an ionization chamber plugged into the phantom.

All phantoms - PMMA Head, PMMA Body and respectively - PE Head and PE Body – were irradiated in the same CT scanner (Siemens Somatom, 1992.a., axial technique) under similar conditions (kVp, mAs, slice thickness etc.) within following kVp settings: 80, 120, 140 kVp. CTDI QA measurements were performed using phantoms of the both types: Head and Body phantoms CTDI value measurements repeated for three times at each point (one in centre and four points on peripheral of a phantom) to gain better statistics. Also was measured (by the CT scanner) phantom CT numbers (in HU) at each CT scan and gained the following results – for details see Table 3 and Figure 7. Here one can see that CTDIw for PMMA and PE have very close values even within various kV settings – the maximum deviation of about 6% between CTDIw value for PMMA and PE was gained when scanning body phantoms within 140 kV applied (Table 3).

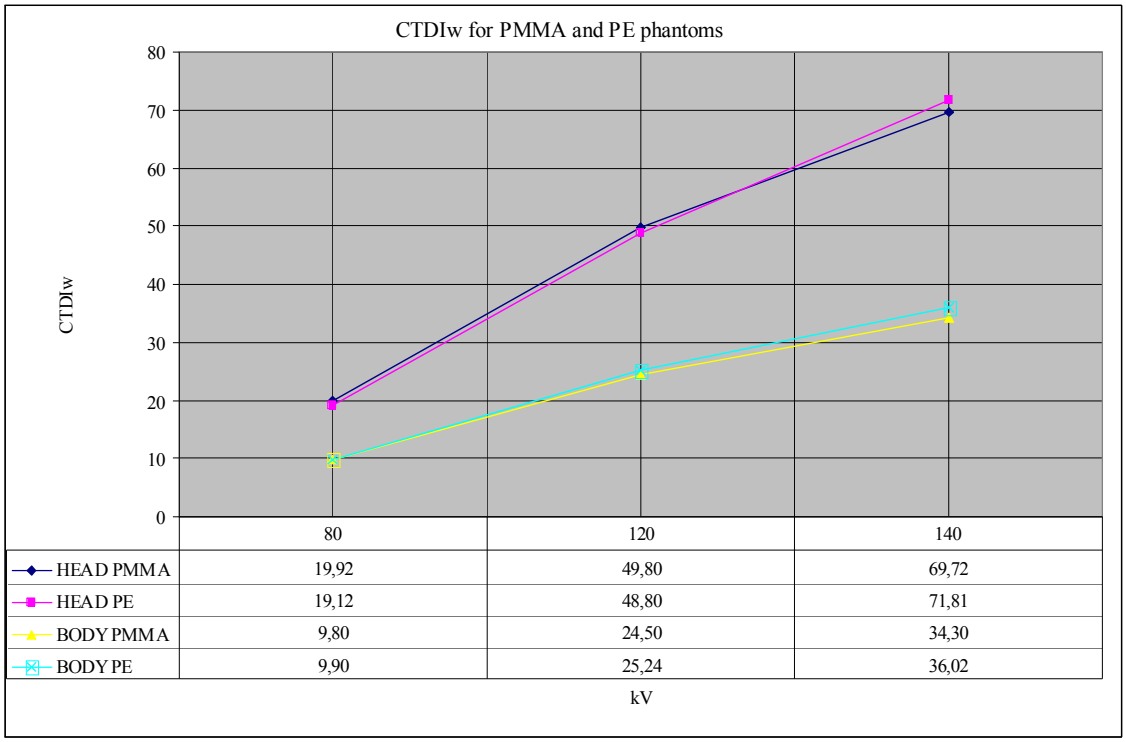


Figure 7. $CTDI_w$ measured under various kVs for PMMA and PE phantoms

As the CT number for a given tissue or a compound must remain uniform across all kV settings, the small deviation up to 5 HU (less than 5%) between CT number values for PMMA and PE phantoms is quite acceptable (Table 3) and almost within a electrometer standard error (4%). For example, in a well calibrated scanner, water has a CT number that ranges from -3 to +3 HU [from GE HiSpeed CT scanner QA manual]. An important fact was derived from the comparison – with the self-manufactured polyethylene phantom CT scanner showed smaller HU numbers that leads to a suggestion that polyethylene phantom is more water equivalent, e.g. more tissue equivalent. This make polyethylene phantom more valuable for clinical investigations.

Table 3. Comparison of $CTDI_w$ values for the reference PMMA and self-manufactured PE phantom

phantom	HEAD PMMA									
CT number, HU *)		120	÷	2						
FOV	25cm SFOV									
mA	260 mA									
scan time	1 sec scan									
scan mode	5 mm									
kV			80			120			140	
$CTDI_w$, mGy **)		19,9	÷	0,6	49,8	÷	1,5	69,7	÷	2,1

phantom	HEAD PE									
CT number, HU		116	÷	2						
FOV	25cm SFOV									
mA	260 mA									
scan time	1 sec scan									
scan mode	5 mm									
kV			80			120			140	
$CTDI_w$, mGy		19,1	÷	0,5	48,8	÷	1,4	71,8	÷	2,1

phantom	BODY PMMA									
CT number, HU		120	÷	2						
FOV	50cm SFOV									
mA	260 mA									
scan time	1 sec scan									
scan mode	5 mm									
kV			80			120			140	
$CTDI_w$, mGy		9,8		0,3	24,5	÷	0,7	34,3	÷	1,1

phantom	BODY PE									
CT number, HU		115	÷	2						
FOV	50cm SFOV									
mA	260 mA									
scan time	1 sec scan									
scan mode	5 mm									
kV			80			120			140	
$CTDI_w$, mGy		9,9	÷	0,3	25,2	÷	0,8	36,0	÷	1,2

*) Here, the uncertainty of CT number is ±1 HU

**) Here, the uncertainty of $CTDI_w$ is 4% or up to ±2.1 mGy depending on a measured value.

Phantom and chamber positioning and alignment

For chamber and phantom alignment there can occur both rotational and translational errors. A lot of time can be spent trying to get precise alignment. Fortunately, measurements are not very sensitive to over precise alignment, and quite generous tolerances on set-up are given below [23]. It is usually best to use the laser lights for approximate positioning (the use of level can also be helpful).

Alignment lights

The most useful lights are those defining the scan plane, with the other lights (sagittal and coronal planes) assisting in centering the phantom or chamber in the scan plane, see Figure 8. Light positioning accuracy can alter with the off-axis (off-isocentre) distance, or, where two or more lights are used to identify the same plane they may only match at one specified position, or they may not be properly set up.

Therefore in order for the lights to be used as the primary means of alignment for the phantom or chamber positioning they would need to be checked for both on-axis (for in air measurements), and at 8cm and 16cm off-axis (for phantom measurements).

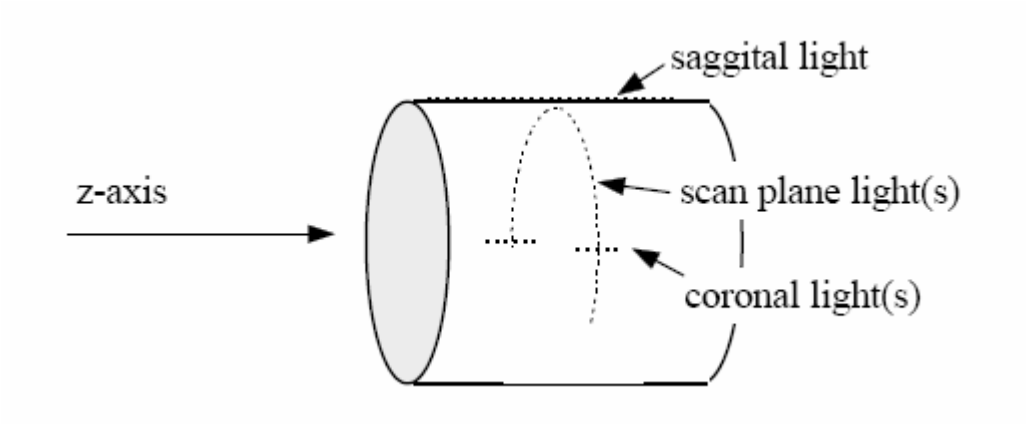


Figure 8. Scanner lights defining planes, as seen on a phantom [23]

Swivel and tilt alignment

The ‘swivel’ and ‘tilt’ alignment errors describe the misalignment of the axis of the phantom or chamber along the scanner (gantry) axis (conventionally the z-axis). ‘Swivel’ refers to the chamber or phantom swivelling left to right through the vertical plane, and ‘tilt’ is the movement from the horizontal.

The extent of the misalignment is measured either as an angle or as a divergence of the phantom or chamber ends from the orthogonal image axes, illustrated in Figure 9 and Figure 10. In both instances the alignment accuracy required is 2 degrees.

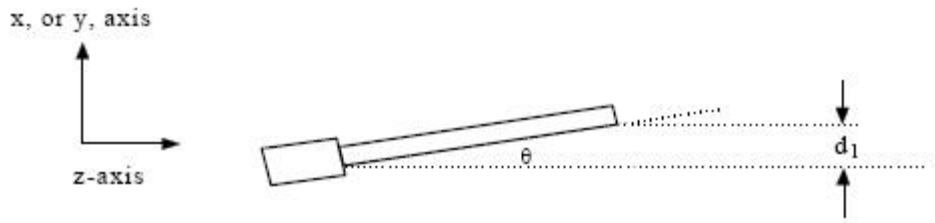


Figure 9. Chamber swivel and tilt. Tolerance: with θ = 2 degrees, d_1 = 3.5 mm. [23]

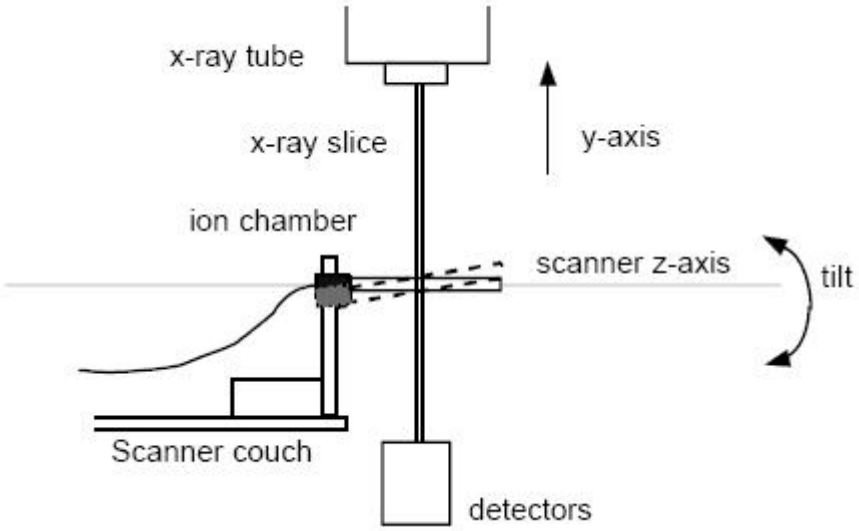


Figure 10. Chamber Tilt, Lateral View. [23]

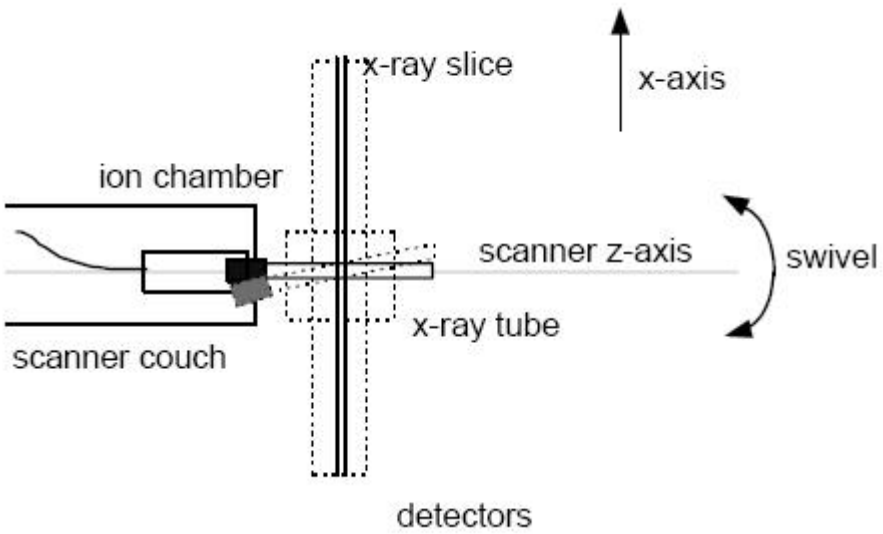


Figure 11. Chamber Swivel, Plan View. [23]

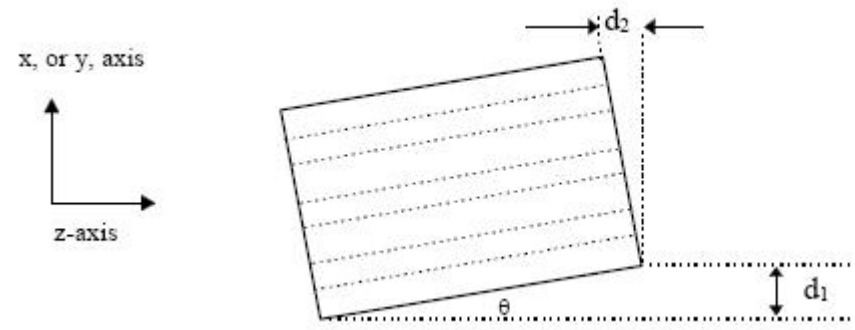


Figure 12. Phantom swivel and tilt. Tolerance: with θ = 2 degrees, d_1= 5 mm for both phantoms and d_2 = 5mm (head phantom) and 11 mm (body phantom). [23]

Rotation alignment

In addition to swivel and tilt there is the 'rotation' alignment in the scan plane, Figure 11 and Figure 12. The phantoms have chamber hole positions which should lie at 0, 90, 180 and 270 degrees (with the 90 to 270 degree line being horizontal),Figure 13. (These positions are referred to as N, E, S and W. In some other literature and mostly in QA datasheets from original CT vendors they call it as B, C, D, E and the central hole is A).

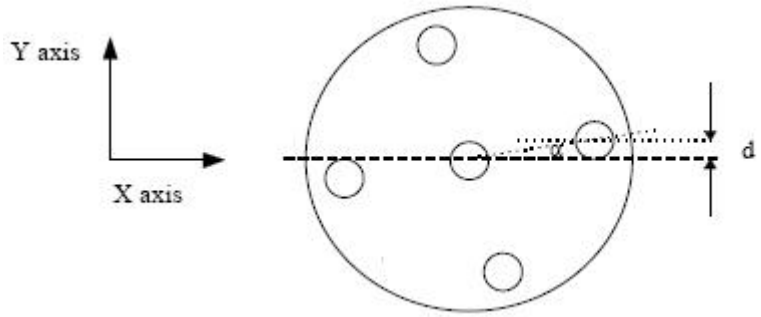


Figure 13. Phantom rotation. Axial view, Tolerance: with θ = 2 degrees, d = 2.5 mm (head phantom), or 5.5 mm (body phantom). [23]

Lateral and vertical displacement

The translational tolerances are given in terms of displacement from the X and Y axes in the scan plane, more simply described as 'up and down', and 'left and right' by an observer looking into the gantry along the scanner axis, see Figure 14.

The centre phantom measurements are not unduly affected by the phantom being positioned away from the iso-centre. The periphery phantom measurements are more affected, but by taking the mean value around the phantom any effects are cancelled out. A tolerance of 5 mm is allowed.

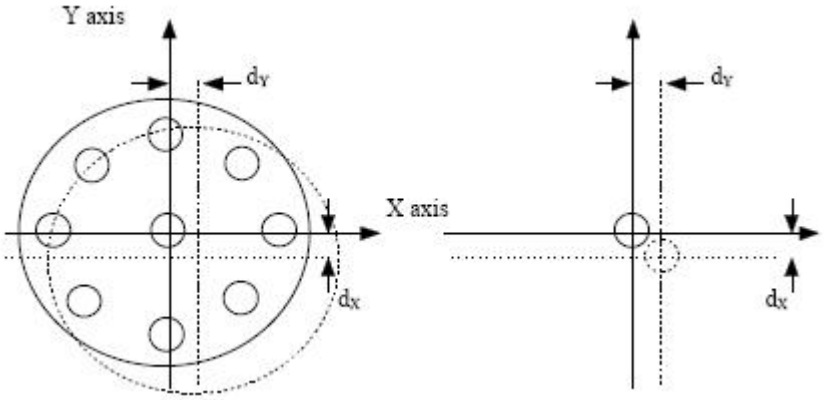


Figure 14. Phantom or chamber lateral or vertical displacement. Axial view, Tolerance : dx and dy = 5 mm. [23]

Electrometer and ionization chamber

CTDI measurements were performed using a pencil shaped ionization chamber (Model DCT 10 RS Lemo; Barracuda, RTI Electronics AB, Sweden) connected to a radiation measuring device (Barracuda, RTI Electronics AB, Sweden) on the axis of rotation of each scanner. The electrometer in its turn is connected to a laptop using over 5m long COM or USB cable. Information on measured doses and waveform images was displayed by oRTIgo QA and Dosimetry software came in bundle with Barracuda electrometer. The system was calibrated according to International Electrotechnical Commission (IEC) standards and traceable to SSI (Sweden Radiation Protection Authority) standards. According to the device's original documentation, standard uncertainty of the ion-chamber is within ± 3 %, and standard uncertainty of whole system including the electrometer and the ion-chamber is within ± 4 %.

CT scanners information

The CT scanners investigated in this study are located in four hospitals in three different cities in Estonia - see Table 4 for details. West-Tallinn Central Hospital has a GE HiSpeed QX/I 4-row helical scanner (GE, Milwaukee, USA), East-Tallinn Central Hospital has a Philips LX 4-row helical scanner (Philips Medical Systems, The Netherlands), in Pärnu Hospital there is GE HiSpeed DX/I (GE, Milwaukee, USA) helical scanner and the last scanner is a recently installed Siemens SyngoCT 6-row helical scanner in Kohtla-Järve Hospital.

Table 4. Investigated CT scanners

Hospital Name	CT Scanner Manufacturer and Model	Scanner technique
1. East-Tallinn Central Hospital	Philips LX (2003, Philips Medical Systems, The Netherlands),	4-row helical scanner
2. Kohtla-Järve Hospital	Siemens SyngoCT (2005, Siemens, Germany)	6-row helical scanner
3. Pärnu Hospital	GE HiSpeed DX/I (2001, GE, Milwaukee, USA)	2-row helical scanner
4. West-Tallinn Central Hospital	GE HiSpeed QX/I (2003, GE, Milwaukee, USA),	4-row helical scanner

General measurement information

At the beginning of the measurements we took exposure readings (accordingly to [23]); if the readings were consistent (< 3% variation) it was possible to save time by making only one exposure for subsequent studies. However, since the majority of the time spent in these measurements is usually the setting up time of the chamber and phantoms, we would prefer 3 exposures for each measurement unless time was very pressing (due to overheating limitations of the X-ray tube). Three readings are definitely required for the off-axis CTDI in-phantom measurements, in order to obtain some information related to scan start position. We recorded the meter readings without any correction factors or calibration factors applied. If there was a possibility we also measured and recorded temperature and pressure (for ion-chamber values correction), but this is still acceptable to make measurements without these since even within significant temperature and pressure variations the correction factor is usually around 1% of measured dose value.

Examinations were categorized as follows: (1) brain; (2) chest; (3) abdomen and pelvis. Examination protocol parameters such as kilovoltage (kV), tube current–exposure time product (mAs), slice thickness T, slice increment I, window width and window level were taken for standard sized patients. Head examinations on all scanners were performed using axial techniques; standard chest, abdomen and pelvis exams performed using helical protocols. All available technique and equipment parameters were recorded see Table 5 and Table 6. Here, each scanner has its unique nickname from A to

D in these tables. However, those names are mixed up for anonymization reason and do not directly correspond to the scanners' names brought in Table 4.

Table 5. Number of slices *N* and irradiation length *L* of examined body region

Examination	Parameter	A	B	C	D
Head	*N*	14 + 16 = 30	20	25	18 + 15 = 33
	L (cm)*	3.25 + 7.50 = 10.75	10.00	12.00	5.1 + 8.4 = 13.00
Chest	*N*	28 + 33 = 61	40	40	61
	L (cm)*	20.25 + 4.00 = 24,25	20.15	27.30	30.00
Abdomen, pelvis	*N*	33	40	33	41
	L (cm)*	32.00	20.15	22.40	25.50

*) Precision of L value here is ±0.05 cm or ±0.5 mm

$CTDI_c$ and $CTDI_p$ were measured with the ionization chamber in a head phantom using brain examination protocols on corresponding scanners and in a body phantom for chest, abdomen or pelvis examination

Comparison of CT Dose protocols.

It was decided to investigate so called CT Dose protocols – automatic dose reports, generated for every performed CT study. The CT Dose protocol feature was only available on some modern CT scanners, see Table 8. CT Dose protocol data from 1000 (one thousand) studies made on each scanner was collected and segmented by study type. However, in this work we will use this information only for reference. Find the collected data in the next chapter, see Table 13.

Table 6. Comparison of CT examination protocols within the following parameters: kilovoltage (kVp), tube current-exposure time (mAs), slice thickness (*T*), increment (*I*) and pitch (*p*).

Exam.	Scanner	kVp	mAs	*T* (mm)	*I* (mm)	*p*
Head	A	140	200	2 x 2.5	5.0	Axial
		120	160	2 x 5.0	10.0	Axial
	B	120	150	2 x 5.0	10.0	Axial
	C	120	100	5.0	5.0	Axial
	D	130	250	3 x 2.0	6.0	Axial
		130	270	6 x 1.0	6.0	Axial
Chest	A	120	350	7.5	7.5	1.5
		120	350	7.5	7.5	1.5
	B	120	360	6.5	5.0	1.6
	C	120	200	7.0	7.0	1.8
	D	130	84	5.0	5.0	1.2
Abdomen, pelvis	A	120	350	7.5	7.5	1.5
	B	120	360	6.5	5.0	1.6
	C	120	200	7.0	7.0	1.5
	D	130	90	6 x 2.0	12	1.2

Table 7. CT scanners investigated to collect CT Dose Protocol data

Hospital Name	**CT Scanner Manufacturer and Model**	**Scanner technique**
1. Kohtla-Järve Hospital	Siemens SyngoCT (2005, Siemens, Germany)	6-row helical scanner
2. South-Estonian Regional Hospital	GE (2005, GE, Milwaukee, USA)	16-row helical scanner
3. South-Estonian Regional Hospital	Siemens (Siemens, Germany)	2-row helical scanner
4. West-Tallinn Central Hospital	GE HiSpeed QX/I (2003, GE, Milwaukee, USA),	4-row helical scanner

7. RESULTS

Examinations were performed with parameters kV, mAs, *T* and *I* taken from routine protocols for head (or brain), chest and abdomen (or abdomen + pelvis) examinations at each scanner, for standard sized patients. The kilovoltage parameter was 120 kVp for almost all examinations at all scanners except for 140 kVp value within the 2nd series of brain examination protocol at the scanner A. The most variable parameter between scanners was mAs, with the C using the lowest value (100 mAs) at brain examinations and the B using the highest value (360 mAs) at abdomen examinations. In practice scanners A and C may apply lower mAs due to the technique of automatic tube current modulation; ACS-DoseRight technique (Automatic Exposure Control) can be used on the scanner B. Examination protocol details are shown in Table 5 and Table 6.

Table 5 contains values of total number of slices N and irradiation length L for each type of examination. In routine protocols for all three scanners there can be found comparable L for brain, abdomen and chest examinations. But the fact must be taken in account that in the case of brain or head examinations scan length is very similar for standard patients and this is usually within 10-12cm. L for chest and abdomen examinations can have a broader range of values (25–40 cm) depending on the size of the thorax.

$CTDI_w$ values

$CTDI_w$ and DLP were then calculated for each examination, the mean results are shown in Table 8 and Table 10, see also Figure 15 for $CTDI_w$ data comparison. $CTDI_w$ was calculated for each scanner from an average of three measurements in the head phantom and another three measurements in the body phantom. EC and SSI RDLs for $CTDI_w$ and DLP are found in Table 1. Brain examination was performed only without contrast medium. So, in practice, if the patient is scanned using both methods – with and without contrast -, the radiation dose doubles. The same condition concerns abdominal studies; in practice double or triple series can be made during the same study. Correspondingly, the DLP and effective dose may be several times larger. However, in first iteration only single series for routine examination protocols will be performed.

$CTDI_w$ of each examination protocol investigated was below the EC RDL and SSI, except for head (brain) examination at the scanner A and D. Performance of all scanners was satisfactory as far as $CTDI_w$ is concerned.

Table 8. Mean weighted computed tomography dose index (CTDIw) results compared with European Guidelines (EG) and SSI regulations (SSI)

Examination	Quantity	A	B	C	D	EG	SSI
Head	CTDIw, mGy	65,7	42,9	47,7	128,6	60,0	75,0
Chest	CTDIw, mGy	12,5	14,0	12,6	9,6	30,0	20,0
Abdomen	CTDIw, mGy	7,7	14,0	15,1	10,3	35,0	25,0

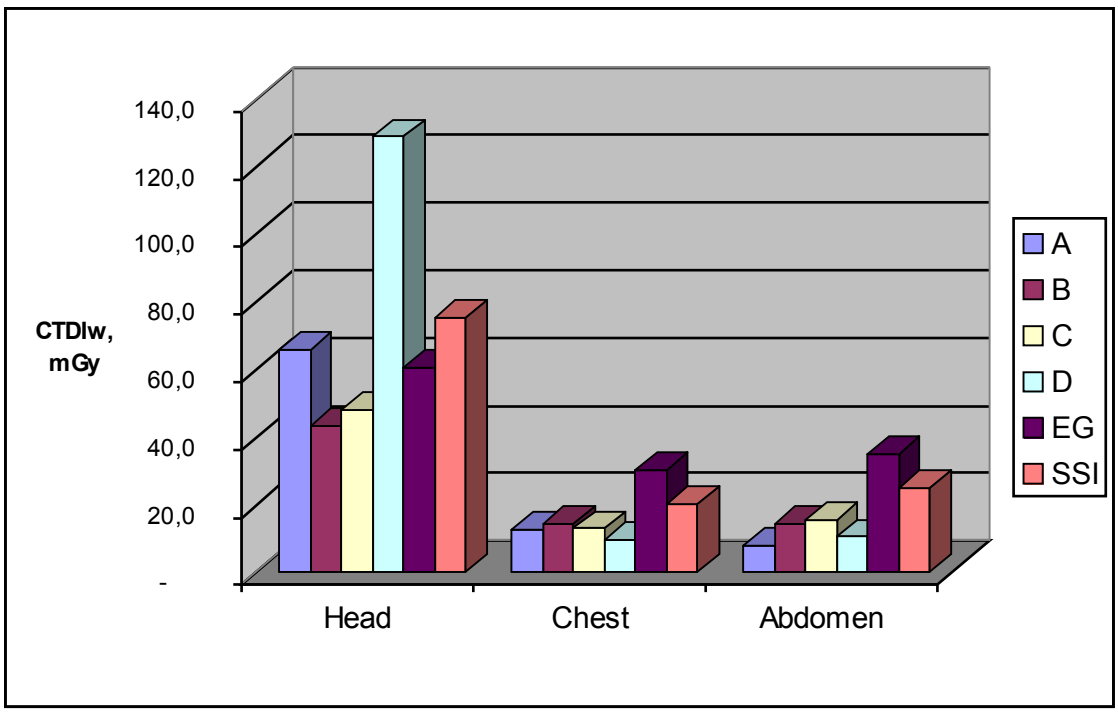


Figure 15. Comparison of mean $CTDI_w$ values to EG Reference Levels (EG) and to reference CTDI values stated in SSI regulations (SSI).

The survey data include a substantial sample (4 of 10) of scanners in Estonia, with a reasonable spread in terms of geography and technology that is sufficiently representative for the purposes of setting preliminary national reference doses. What is important, the survey establishes the methodology for conducting further periodic reviews of CT practice.

Historically, such guidance levels have been set pragmatically on the basis of third quartile values of the dose distributions from wide-scale surveys [3]. As we have very

low statistics in terms of observed installation number, we may not apply on statistical quantification with segmentation of results into quartiles. In this case we may figure out maximal doses for each kind of study as RDLs, see Table 9.

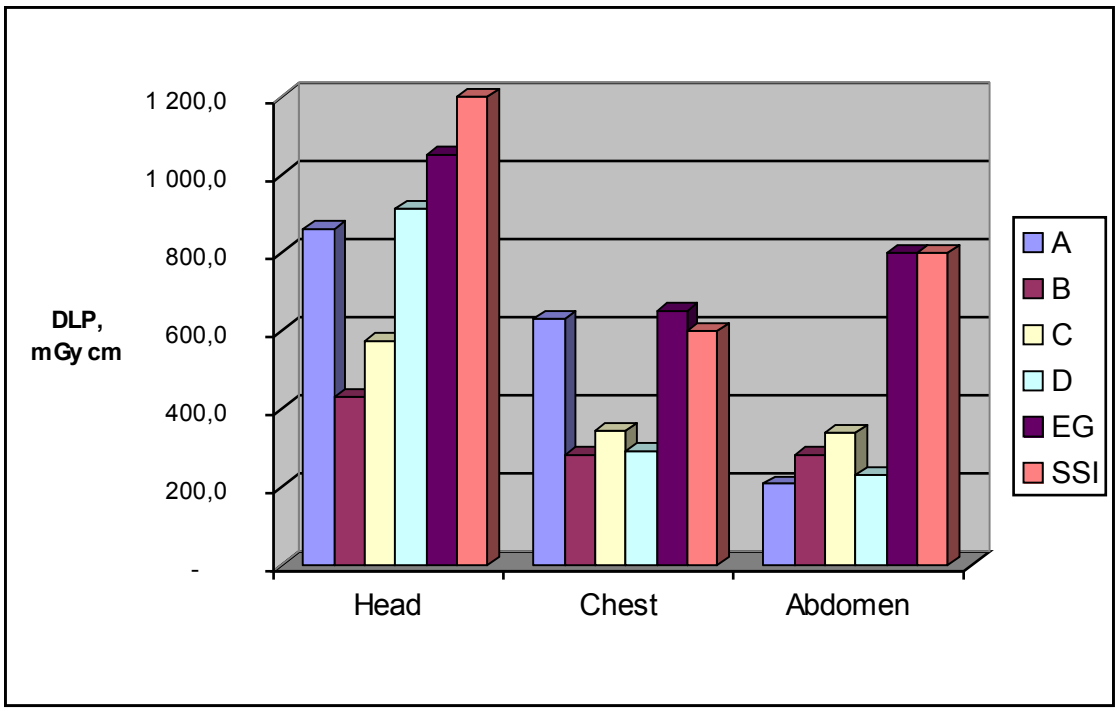


Figure 16. Comparison of mean DLP values to EG Reference Levels (RL).

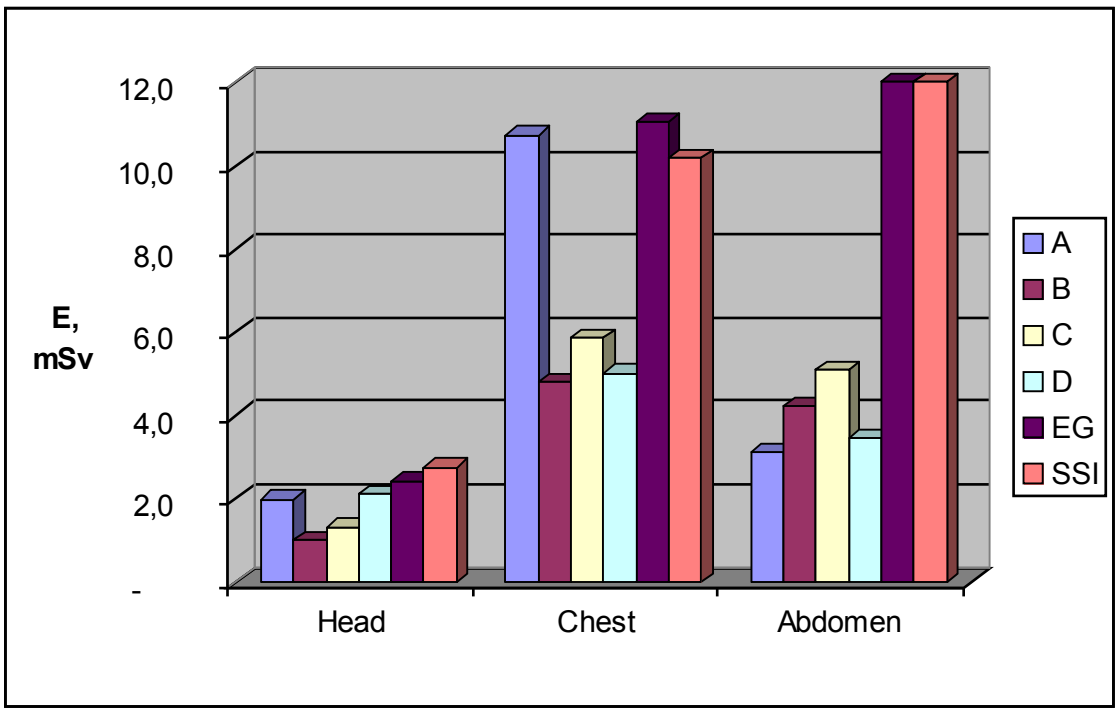


Figure 17. Comparison of mean E values to derived EG and SSI effective dose Reference Levels.

Data from 1991 year UK CT survey were subsequently used when establishing reference doses as part of European Guidelines on quality criteria for CT [5]. In Table 9, there are presented NRPB values taken from the latest CT UK survey performed in 2003 year [9].

Table 9. Suggested RDLs for weighted computed tomography dose index ($CTDI_w$) results compared with European Guidelines (EG), SSI and NRPB

Examination	Quantity	Est RDL	NRPB	SSI	EG
Head	CTDIw, mGy	65	65	75	60
Chest	CTDIw, mGy	14	16	20	30
Abdomen	CTDIw, mGy	15	20	25	35

In case of RDL for $CTDI_w$ suggested by the authors, it decided not to take in account $CTDI_w$ value measured at scanner D as it twice higher than EG RDL, moreover, it's known that D is a newly installed scanner with 'raw' CT protocols. Thus the next maximum value for $CTDI_w$ is 65 mGy, which is the same as NRPB RDL for $CTDI_w$.

Table 10. Mean dose–length product (DLP) results compared with European Guidelines (EG)

Examination	Quantity	A	B	C	D	EG	SSI
Head	DLP, mGy·cm	862,4	429,0	572,4	910,0	1 050,0	1 200,0
Chest	DLP, mGy·cm	631,6	282,1	344,0	295,0	650,0	600,0
Abdomen	DLP, mGy·cm	208,5	282,1	338,2	230,0	800,0	800,0

*) No diagnostic reference level has been established for DLP by SSI [6]. However, we assume it should be within EG RDL.

DLP values

DLP was found to be within proposed EG for brain, abdomen and chest examinations and compared to EG and SSI RDL , for details see Table 10 and Figure 16. Running brain examination on the scanner A the $CTDI_w$ value was slightly above the EC and SSI RDL due to the higher value of $CTDI_w$, within the almost the same short L in comparison with the other scanners. And in case of scanner D near brain exam we get almost twice higher dose than EG consider for that kind of examinations. That may be of the fact that this is a brand new installation with non-well optimised protocols came with manufacturer defaults.

All chest DLP were sufficiently under EG of 650 mGy cm, except for scanner A, where it was equal to 631 Gy·cm, which is within the reference level. Since RDLs act as

parameters to help identify relatively poor or inadequate use of technique, the exposure settings and the extent of the scan should be further investigated to lower the dose without affecting image quality.

Table 11. Suggested RDL values for DLP compared with EG, NRP and SSI

Examination	Quantity	Est RDL	NRPB	SSI	EG
Head	DLP, mGy cm	910	930	1 200	1 050
Chest	DLP, mGy cm	630	580	600	650
Abdomen	DLP, mGy cm	340	560	800	800

Here, all suggested RDLs for DLP are rounded maximum numbers for DLP measured on Estonian CT scanners. It can be seen that suggested values are in good correlation with other RDL proposed by NRPB, SSI and EG, whereas we suggested slightly smaller values for DLP RDL.

Effective Dose

Here we calculated mean effective doses for Head, Chest and Abdomen examinations multiplying DLP values with the correcting factor of correspondent tissues, see Table 12 and Figure 17. However, it must be considered that these effective dose numbers are derived applying effective dose conversion factors (proposed by EG, [5]) to mean DLP values introduced above.

Table 12. Evaluated mean effective dose (D) results compared with European Guidelines (EG) and SSI evaluated effective dose values

Examination	Quantity	A	B	C	D	EG	SSI
Head	*E*, mSv	1,98	0,99	1,32	2,09	2,42	2,76
Chest	*E*, mSv	10,74	4,80	5,85	5,02	11,05	10,20
Abdomen	*E*, mSv	3,13	4,23	5,07	3,45	12,00	12,00

Comparison of results to other authors

Clarke et al [22] and Tsapaki V. [19] presented $CTDI_w$ and DLP results for the same examinations. Their results have a broad range of values, and one of the reasons is the larger number of scanners included in the study.

Values show by V. Tsapaki et al's were well within proposed EG for both $CTDI_w$ and DLP, apart from chest DLP on one of examined scanners exceeded the EC or SSI RDL. For chest examination, Clarke et al's scanned volume length (range 13.4–28.7 cm) was

generally lower than in our study range 20–45 cm), which seems to have great implication for DLP. Verdun et al [20] presented DLP results for standard abdominal examinations in the range 421–904 mGy cm. Range of DLP value in our abdominal protocols was lower (208-338 mGy cm) and not as broad as at Tsapaki et al's (278–582 mGy cm), probably because our scanning length was 20–32 cm, which is shorter than the scanning length of 38 cm presented in Verdun's [20] study and very close to results reported by Tsapaki et al. [19].

CT Dose Protocol Data

At each scanner we collected information for 1000 consecutive CT studies performed in period from the beginning of year 2006, see Table 13. Here, each scanner is assigned an unique code from A to D. However, those names are mixed up for anonymization reason and do not directly correspond to the scanners' names brought in Table 7. CT Dose Protocol information was collected to show that measured DLP results, presented in Table 10, are of the same order as real studies have.

On the basis of DLP data taken from Dose Protocols there was calculated mean effective doses for Routine Head, Chest and Abdomen examinations multiplying DLP values with the correcting factor of correspondent tissue (see Table 2). Evaluated effective doses for Dose Protocol data are presented in Table 14.

Table 13. CT Dose Protocol data with comparison to EG and SSI DRL values. Shown are mean DLP values for third quartile of each exam type, mGy cm

Body part	Examination	A	B	C	D	EG	SSI
Head	*Brain routine*	1 240	1 243	839	922	1 050	1 200
	Sinuses	303				1 050	1 200
	Angio	2 577	1 701			1 050	1 200
	Cervical spine	942			416	1 050	1 200
Chest	*Chest routine*	741	410	310	320	650	600
	Chest + abdomen		1 418			650	600
	Chest + abdomen + pelvis	2 069	2 253			650	600
Abdomen, pelvis	*Abdomen routine*	1 047	298	556	253	800	800
	Abdomen+ pelvis	1 680		1224	646	800	800
	Lumbar spine	474	580	336	453	800	800

Table 14. Evaluated effective dose based on CT Dose Protocol data and DLP conversion factors. E, mSv

Body part	Examination	A	B	C	D	EG	SSI
Head	*Brain routine*	2,9	2,9	1,9	2,1	2,4	2,8
	Sinuses	0,7	-	-	-	2,4	2,8
	Angio	5,9	3,9	-	-	2,4	2,8
	Cervical spine	2,2	-	-	1,0	2,4	2,8
Chest	*Chest routine*	12,6	7,0	5,3	5,4	11,1	10,2
	Chest + abdomen	-	24,1	-	-	11,1	10,2
	Chest + abdomen + pelvis	35,2	38,3	-	-	11,1	10,2
Abdomen, pelvis	*Abdomen routine*	15,7	4,5	8,3	3,8	12,0	12,0
	Abdomen + pelvis	25,2	-	18,4	9,7	12,0	12,0
	Lumbar spine	7,1	8,7	5,0	6,8	12,0	12,0

More deep investigation of dose protocols is beyond of the scope of this work and will be covered in upcoming publications of the author.

8. DISCUSSION

CT examinations on head or brain appear to have the highest DLPs among examined CTs, and also have a higher than EG $CTDI_w$ criteria. The large irradiation volume of investigations seems to be an important factor since $CTDI_w$ is within RDLs. Reducing the extent of the scan as much as possible, without missing any vital anatomical regions, could be a first step to lower DLP and effective dose to patient *E* [24].

Furthermore, reducing mAs of the examination protocol is also important, especially for patients who are thinner than the standard sized patient. W. Kalender et al. [7] presented a study regarding the minimum tube current required for good image quality with the least radiation dose on CT chest examination. Results of the study indicate that the lowest mAs can be used without affecting diagnosis, despite of the fact that images may be noisier. A number of other scanning parameters can be easily adjusted for lower doses and better images, very good source for that was found in W. Kalender [7] and P. Hiles [25] references.

As far as the other examinations (brain, abdomen and chest) are concerned, the protocols utilized in observed hospitals have $CTDI_w$ and DLP values that are well within EG dose criteria. This is encouraging, since the most important aspect of radiation protection is to have the amount of dose absorbed by the patient as low as reasonably achievable, provided that this does not affect image quality and precise diagnosis [3].

The $CTDI_w$ and DLP values found for each CT scanner will be used as a local reference level for each examined hospital. It is important that CT RDLs should be monitored at certain time intervals to constantly assure optimization of the procedure. Furthermore, all examination protocols performed in each hospital will be investigated in terms of technique and radiation dose and compared with EG, SSI and UK RDLs, since they appear to be a very useful tool in assessing standard CT performance [2].

To compare radiological examinations in terms of radiation risk, taking into account the relative radiosensitivity of different body regions involved, it is necessary to estimate effective dose E, which is the sum of the products of organ doses and corresponding weighting factors [1]. Shrimpton et al [3] calculated *E* from CTDI measurements using

Monte Carlo conversion coefficients. Organ doses can also be measured using TLDs inside and at the surface of phantoms [1][2][20]. Estimating of collective doses and doses to young patients will be our next project in the field of CT dose evaluation and monitoring in our country.

9. CONCLUSION

For the first such CT dose tests were successfully performed in four hospitals of Estonia. There we have evaluated $CTDI_w$ and DLP values with further comparison to European Guidelines and SSI (Sweden Radiation Protection Authority) quality criteria, which showed good correlation with EU normative; compared routine head, chest and abdomen CT protocols.

Current work provides essential data to facilitate further initiatives in the optimisation of patient protection in CT. In particular, the report includes suggested national reference dose values (derived as rounded maximum doses for $CTDI_w$ per sequence and DLP per examination) as simple direction to help identify Estonian authorities where levels of dose are unusually high.

Today there are no RDLs for *E* regarding CT examinations. Since *E* provides a direct estimation of radiation risk and is useful for comparison with other radiological examinations, it should be always evaluated as we did it in our work. However, we proposed RDL for *E* as the corresponding values were evaluated using rough calculations and gained results differ quit a lot from other authors (ours results for *E* 1.5-5 times less) [9]. So, RDL for *E* will stay the matter to be further investigated.

However, diagnostic reference dose values should not be applied locally on an individual patient basis, but rather to the mean dose observed for representative groups of patients [26]. For the establishment of national reference doses, all of 10 CT scanners in Estonia should be monitored with thorough investigation of routine CT exam protocols. Periodic review of these data will allow timely analyses of trends and the updating of national reference doses for CT. There is hope that this will be the goal of the authors' next study.

Having carried out this survey, now we have the knowledge base and possibilities for reducing dose to patient and improving the quality of CT images at minimal dose at national level in Estonia..

10. KOKKUVÕTE

Töö eesmärgiks oli dooside mõõtmine kompuutertomograafia (Computer Tomography - CT) kabinettides üle Eesti ning tulemuste võrdlemine Euroopa referentsdoosidega (European Guidlines, [5]). Uuringute käigus mõõdeti järgmisi dosimeetrilisi väärtusi: $CTDI_w$ (kaalutud kompuutertomograafia-doosiindeks), DLP (doospikkus) ja *E* (efektiivdoos). Võrreldi tüüpiliste CT protokollide korral leitud $CTDI_w$ ja DLP väärtusi Euroopa ja Rootsi referents doosidega. Võrdlusanalüüs näitas, et tüüpilised kopsu- ja kõhuuuringute protokollid vastavad Euroopa Komisjoni poolt soovitatud tasemele kiirgusdooside ja kasutatavate uurimismeetodite osas. Kahes haiglas saime kolju- ja ajuuuringute protokollide korral vastavalt 30% ja üle 100% kõrgema $CTDI_w$ väärtuse. Kõigil neljal skanneril on $CTDI_w$ ja DLP väärtused kopsu- ja kõhuuuringute puhul Euroopa referentstasemest vähemalt 2 korda madalam. On arutatud edaspidiseid võimalusi patsiendidoosi vähendamiseks ning tüüpiliste uuringuprotokollide optimeerimiseks kompuutertomograafias.

11. ABSTRACT

The purpose of this paper is to perform dose measurements in Computed Tomography (CT) cabinets in around Estonia and compare the results with EU reference dose levels for CT ([5]). The measured dosimetric quantities are weighted computed tomography dose index ($CTDI_w$), dose–length product (DLP) and effective dose (E). Along with dose measurements we performed a comparison of standard routine protocols for most common CT examinations. Measured values show that examined CTs meet EC RDLs for chest and abdomen examinations, in terms of radiation dose and examination technique. Only the $CTDI_w$ value for routine brain and skull examinations in two hospitals exceed recommended value by approximately 30% and over 100% respectively. $CTDI_w$ and DLP values for chest and abdomen examinations on all four scanners are lower than EU recommended levels at least by a factor of 2. Further ways of reducing CT radiation doses and estimation of effective and collective doses to patients are discussed.

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13. TABLE OF FIGURES