An overview of exposure parameters, dose measurements and strategies for dose reduction in pediatric CT examinations

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Received 23 December 2012 – Accepted 29 April 2013

Abstract – CT scanning technology is a valuable tool to diagnose many diseases; however, the level of the radiation dose is a source of concern, especially for children. CT scan systems and dose measurement methods have evolved over the years; but reported pediatric effective doses (EDs) have sometimes exceeded the annual dose limit recommended by the ICRP (1 mSv per year for persons under 18 years) (ICRP, 2007a). Efforts have been made to reduce organ doses and EDs by adjusting the scan parameters. This paper describes the determinants of the ED, and the dose reduction techniques in pediatric imaging from the early age of CT examinations until now. The first epidemiological results regarding the associated risk of cancer are also briefly presented.

Keywords: pediatric CT scan / effective dose / scan parameters / dose reduction

1 Introduction

Since its introduction in 1973, CT has established itself as a valuable diagnostic imaging modality. More than 1000 CT scanners were in use in 50 countries in 1979 (Friedland and Thurber, 1996), a 10% annual growth in the global CT market was reported in the year 2002 (ICRP, 2007a), and currently 6,000 scanners are in use in the United States (Medicine Health, 2012). According to surveys conducted at US medical facilities, the annual number of CT examinations increased from approximately 3.6 million in 1980 to 33 million in 1998 (Nickoloff and Alderson, 2001), and now this value is over 70 million (Brenner, 2010). Currently, the proportion of pediatric patients undergoing CT scans ranges from 0 to 38% depending on the country and examination type (Muhogora et al., 2010).

CT examinations contribute 40 to 67% of the collective dose (UNSCEAR report, Annex D, 2000). This imaging procedure delivers about 67% of the overall radiation dose to the pediatric population (Mettler et al., 2000). Crude estimations showed that the ED ranges between 6 mSv and 100 mSv for pediatric patients (Brenner, 2002).

CT is a major source of medical radiation and its availability and frequency of scanning is responsible for increasing the dose from CT practice. Due to the high ED of CT, an effort to minimize it is critically important. This is especially important in children, because the younger the patient is at the time of exposure to radiation, the greater the risk (BEIR VII Phase 2, 2005). Due to the higher radiosensitivity of children’s cells, the lifetime cancer risk associated with an individual CT examination is higher in children than in adults (ICRP, 2007a) and there is an increased risk for thyroid, skin, brain and breast cancer in children (UNSCEAR report, Annex I, 2000). In addition, due to children’s longer lifetime to manifest radiation-induced cancer, and the fact that cancer risk is cumulative over a lifetime, radiation risk from CT in children is one of the major current concerns in CT dosimetry (Frush et al., 2003).

Over the years, CT technology has evolved, with various impacts on the radiation dose. After the introduction of conventional CT, helical CT became commercially available in the USA and it was on the market in 1991 (Zeman et al., 1998). Because of its new advantages, the use of CT imaging increased in the pediatric population. Although helical technology provides additional opportunities for CT in children, the radiation dose associated with helical CT is much greater than the dose associated with most other imaging procedures (Donnelly et al., 2001).

Concerns about the radiation dose to children increased with multi-detector row CT (MDCT) introduced in the late 1990s (Donnelly et al., 2000). This is because multi-slices are acquired in each gantry rotation. Relative to CT scanners from the early 1990s, MDCT scanners result in doses that are ∼1.7 higher per unit mAs in body phantoms (Huda and Vance, 2007).

In addition, until 2001, children and adults were scanned with identical protocols, which did not differentiate between the large differences in patient sizes (Paterson et al., 2001; Brody et al., 2007). Since there is great variability in body size in the pediatric population, adjusting CT scan parameters such as tube current and voltage is necessary. As reported,
the ED to children from CT examinations increases as body size decreases if the exposure factors are kept constant (Caon et al., 2000). Pediatric protocols should therefore have lower tube current and voltage than those for adults. If spatial resolution is not an issue, the lower tube current settings should be selected as much as possible (Frush and Donnelly, 1998).

The results of the Society for Pediatric Radiologists survey indicated that radiologists now pay more attention to size-based adjustments (Hollingsworth, 2003). If CT parameters used for children are not adjusted based on examination type, or age or size of the child, then some patients will be exposed to an unnecessarily high radiation dose during CT imaging (Donnelly et al., 2001; Khursheed et al., 2002; Pages et al., 2003). Colang et al. (2007) declared that if settings were adjusted based on neonate weight, the dose to the brain (head CT) and stomach (abdominal CT) would be 2 and 4 times smaller than that of unadjusted settings, respectively. Recently, in a cohort study in Britain, it was estimated that in children CT, delivering cumulative doses of about 50 mGy and 60 mGy might almost triple the risk of leukemia and brain tumors, respectively (Pearce et al., 2012). According to the importance of pediatric CT dosimetry, the aim of this paper is to review the main parameters influencing doses received by children, the associated risk and some dose reduction methods.

2 Quantifying the pediatric dose from CT

Radiation doses in CT (organ dose and ED) are estimated in two different ways, by experimental procedures and computer simulations. Two additional dose quantities, the weighted CT dose index (CTDIw) in mGy for a single slice and dose length product (DLP) in mGy.cm per complete examination, which are measured in the CTDI phantoms (homogeneous cylinders of PMMA, with diameters of 16 or 32 cm), give information about relative changes in dose (Shrimpton and Wall, 2000).

2.1 Experimental procedures

This section includes dose determination in clinical procedures or physical measurements with physical phantoms which are scanned by the CT machine to determine dose distributions within the human body. Some of these physical phantoms use simple shapes (rectangular and cylindrical) to display human anatomy (Liu et al., 1996) but they are often made of a human skeleton with tissue-equivalent material simulating the soft tissues and are constructed as vertical slices with small holes for dosimeter placement. Pediatric organ dose calculation started with determining the surface and internal radiation doses in abdominal CT (Brasch et al., 1978; Brasch and Cann 1982). Later, the surface dose was compared with CTDI data to estimate the entrance exposures to a set of ATOM phantoms (a family of physical phantoms manufactured by CIRS which include head, torso, upper femur and genitalia) in chest and abdomen-pelvis (AP) CT (Cody et al., 2004). Commonly, the CT dose is represented by the ED and organ absorbed doses. In Table 1, some studies that investigated the pediatric received dose by clinical procedures or physical measurements are tabulated.

In addition to the dose, the diagnostic reference level (DRL) is specified to promote optimization of patient protection. Shrimpton and Wall (2000) calculated the third quartile of the CTDIw and DLP as the DRL, in brain, chest, abdomen and pelvis scans for a baby to a 15-year-old child, which were 20–70 mGy and 50–800 mGy.cm, separately. Later, in a national survey, they determined UK DRL for pediatric head and chest scans. They reported that the CTDIw and DLP ranged from 15 to 56 mGy and 76 to 508 mGy.cm, respectively (Shrimpton et al., 2006).

2.2 Monte Carlo simulation

Computer simulation for dose estimation is the most reliable way to obtain accurate values of organ doses under CT imaging (Lee et al., 2011). Some Monte Carlo (MC) programs using MCNPX (Khursheed et al., 2002; Lee et al., 2007; Gu et al., 2009; Lee et al., 2012) and PENELOPE (Li et al., 2011) were developed which simulate the dose inside the computational models of the human body.

Using MC in pediatric CT started with determination of the organ dose per air kerma for head and chest scans with single detector CT (SDCT) at tube voltages of 80 and 125 kV for GSF phantoms (BABY and CHILD). The maximum organ doses per air kerma in chest scans were in the breast (0.96 for BABY and 0.88 for CHILD) (Zankl et al., 1995). Some studies only investigated the amount of the dose absorbed in one organ. In head and neck CT examinations, the mean dose to the thyroid was calculated using stylized phantoms representing 1 year to 15 years old. The thyroid dose varied between 0.6–8.7 mGy and 15.2–52.0 mGy in head and neck CT imaging, respectively (Mazonakis et al., 2007).

In a retrospective cohort study of over 240,000 children in UK and by using an organ dose database from MC simulation, Kim et al. (2012) reported the absorbed dose in the brain, thyroid, breast and RBM of a newborn to a 22-year-old in head, chest and abdomen scans before and after 2001. The maximum dose of the brain, thyroid, breast and RBM before (after) 2001 were 56 (44.2), 27.7 (13), 36.9 (13.3) and 17.1 (8.6) mGy, respectively. The EDs calculated in some studies by MC methods are given in Table 2.

3 CT dose reduction methods

Improvements in CT technology (e.g. detector efficiency, geometry efficiency, current modulation and reconstruction algorithms) have decreased patient doses significantly. Starting in the 1990s, significant efforts have been made to lower the dose to the pediatric population (ICRP, 2000). By changing the CT parameters based on the patient’s weight or age, the dose is reduced significantly. However, the radiation dose should only be reduced under the condition that the diagnostic image quality is not sacrificed to ensure appropriate diagnosis.
Table 1. Studies that calculated the dose by physical measurements or clinical procedures.

<table>
<thead>
<tr>
<th>Reference</th>
<th>ED in mSv (mean organ dose in mGy)</th>
<th>CT scanner</th>
<th>CT parameters</th>
<th>Dosimeter</th>
<th>Subject under exposure</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fearon and Viesche, 1987</td>
<td>(0.02–5.5)</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>Phantoms (6–10 years)</td>
</tr>
<tr>
<td>Akeleto et al., 1996</td>
<td>≤ 2 (0.05–37)</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>1–7 y-old phantom</td>
</tr>
<tr>
<td>Wue, 1999</td>
<td>–</td>
<td>4.4–7.5</td>
<td>–</td>
<td>–</td>
<td>GE HiSpeed Advantage</td>
</tr>
<tr>
<td>Lucaya et al., 2000</td>
<td>–</td>
<td>1.1–7</td>
<td>–</td>
<td>–</td>
<td>GE Twin II-Plus</td>
</tr>
<tr>
<td>Papadimitriou et al., 2004</td>
<td>1.3–2.68</td>
<td>2.83–5.11</td>
<td>9.11–12.12</td>
<td>–</td>
<td>Philips Tomoscan LX Serial</td>
</tr>
</tbody>
</table>
| Huda et al., 2001 | – | 6.4 (50 kg)–9.6 (10 kg) | – | – | GE Hi-Speed CT (80)
| Pages et al., 2000 | 0.4–2.3 | 1.1–6.6 | 2.3–10.9 | – | 5 SDCT and 2 MDCT scanners |
| Mox and McLear, 2006 | 1.34–2.34 | 1.91–7.94 | 4.73–14.14 | – | Helical CT scanners |
| Galaski et al., 2006 | 1.8–2.4 | 1.6–3.7 | 2.9–7.9 | – | MDCT and SDCT scanners of Siemens, Philips, Toshiba and GE |
| Rylka et al., 2007 | 0.2 (0.1–7.7) | – | – | – | SDCT scanner PQ-2000 (Picker) |
| Fujii et al., 2007 | – | 1.3–7.4 | (2–21) | 2.8–10.5 | MDCT scanners (8, 16 and 64 detectors) |
| Donadien et al., 2001 | – | 1.5–29.3 | (2.3–18.6) | – | GE Light Speed Ultra and Siemens Somatom Plus 4 |
| Nishizawa et al., 2003 | 2.6 (16–40) | 1.87–8.17 | (4.36–24.5) | – | 6 types of MDCT scanners |
| Fahey, 2004 | – | – | – | 9.24–12.41 | Discovery LS, GE Healthcare |
| Kim et al., 2010 | – | 8.1 (0.4–16) | (37.8–0.72) | – | Cone beam CT scanner |
| Fujii et al., 2011 | 2.3–2.4 | 1.3–6.7 | – | 16–7.8 | 2 types of 64 MDCT scanners |
| Berme et al., 2006 | 0.85 (0.4–5.0) | 3.05 (0.1–31) | – | 9.55 (4–33) | 10 SDCT scanners |

a Pencil ion chamber.
b Mean absorbed dose for organs within the scanning area.
c Values are related to two different protocols.
d Radiophotoluminescence glass detector.

3.1 Tube voltage adjustment

The tube voltage determines the energy distribution of the X-ray beam, so many authors have investigated the effect of tube voltage variation on changing the CT dose. Reid et al. (2010) determined the trend to decrease tube voltage (and correspondingly, the radiation dose) was not successful (Cody et al., 2004; Nakayama et al., 2005). Tube voltage reduction is allowed only on the condition that it does not affect the ability to detect low-visibility structures. By reducing the X-ray tube potential from 120 kVp to 80 kVp at 160 mA, the value of the CT dose index of a 10-year-old phantom decreased about 67% (Fahey, 2009). Using 80 kVp instead of 120 kVp lowered the dose to the patient by approximately 30% (Siegel, 2003; Yekeler, 2004). A reduction in dose of about 78% in a circular phantom was obtained by Siegel et al. (2004) by decreasing the tube voltage from 140 kVp to 80 kVp (at 165 mAs).

3.2 Tube current reduction

Adjustments in the tube current are more frequently used to improve management of the radiation dose for children. The survey of Hollingsworth et al. (2003) showed a trend to increase tube current with increasing age. In 1999, tube current reduction from the default setting of 200–250 mAs to 125–150 mAs resulted in a 40% reduction in the radiation dose to children (Chan et al., 1999). Another study (Lucaya et al., 2000) shows that a dose reduction of 72% and 80% could be obtained when the standard 180 mAs was decreased to 50 and
Table 2. Studies that calculated the dose by MC.

<table>
<thead>
<tr>
<th>Reference</th>
<th>ED in mSv (mean organ dose in mGy)</th>
<th>CT parameters</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hull et al., 1997</td>
<td>1.5–6</td>
<td>GE HiSpeed Advantage</td>
<td>120</td>
</tr>
<tr>
<td>Cron et al., 2000</td>
<td>2.6–2.8</td>
<td>GE HiSpeed Advantage</td>
<td>120</td>
</tr>
<tr>
<td>Khursheed et al., 2002</td>
<td>6.3–7.8</td>
<td>Siemens DRH</td>
<td>120 and 125</td>
</tr>
<tr>
<td>McLean et al., 2003</td>
<td>6.33</td>
<td>GE HiSpeed Advantage</td>
<td>120</td>
</tr>
<tr>
<td>Hull and Vance, 2007</td>
<td>0.9–3.6</td>
<td>GE HiSpeed Advantage</td>
<td>120</td>
</tr>
<tr>
<td>Lee et al., 2002</td>
<td>0.61–1.36</td>
<td>SOMATOM Sensation 4 helical MDCT</td>
<td>80, 100, and 120</td>
</tr>
<tr>
<td>Lee et al., 2012</td>
<td>0.6–1.5</td>
<td>SOMATOM Sensation 4 helical MDCT</td>
<td>80, 100, and 120</td>
</tr>
</tbody>
</table>

a At tube voltage of 120 kVp and based on ICRP Publication 60 (ICRP, 1990) weighting factors.

b At tube voltage of 120 kVp and based on ICRP Publication 103 (ICRP, 2007b) weighting factors.

43 mAs, respectively. In 2003, dose reduction factors were determined for head and abdominal MDCT in children. Using the reduction factors, pediatric doses were reduced to about 23% and decreased the number of fatal cancers per year by 384 (from 500 to 116) (Boone et al., 2003). The effect of lower tube current on structure detection was investigated in pediatric CT. Frush et al. (2002) found that lowering the current to 67% of the current of the original abdominal MDCT scan did not affect the ability to detect high-visibility structures. Even tube current reductions of 33–50% were acceptable for detection of low-visibility structures. Li et al. (2008) indicated that with 75% current reduction, there is no general statistically significant difference in diagnostic accuracy, and the radiation dose decreased by up to 75%. Currently, automatic tube current modulation (ATCM) is a new technique for radiation dose management (Paterson and Frush, 2007; Coursey et al., 2008).

3.3 The optimum level of tube current and voltage

There are some rules to optimize doses in pediatric CT scans with no loss of diagnostic ability (Vock, 2005). In a study of 30 abdominal helical CT scans of children aged 3 months to 7 years, the optimum level of tube current at a tube voltage of 100 kVp was investigated. It was declared that, most anatomical structures in children were demonstrated at low tube current, and just for imaging a few anatomic structures with small details, performing a CT scan at higher mA would be useful (Wormanns et al., 2001). Using three CTDI phantoms simulating the abdomen of an infant, child and adolescent, Reid et al. (2010) optimized abdominal CT procedures; adjusting mAs and kVp depending on the abdominal circumference.

The results of a recent survey conducted in 2008 showed changes in pediatric body MDCT scanning parameters. Now, 98% of radiologists use either a weight-based or an age-based protocol for pediatric CT. The average tube current has decreased to between 31 and 61 mA for all age ranges. All radiologists now use a peak kilovoltage of 120 kVp or less for routine pediatric chest and abdominal CT (Arch and Frush, 2008).

According to the results of another survey in 2012, using pediatric-specific adjustments a newborn received a lower absorbed dose in the thyroid, breast and brain than an adult male in a chest and brain scan, respectively (Kim et al., 2012).

3.4 Adjusting the pitch

With the advent of helical CT, scanning techniques became more sophisticated. In addition to tube current and voltage, pitch is a selectable parameter (Paterson et al., 2001) which can be increased while tube current decreases (Donnelly et al., 2001; Karabulut and Ariyürek, 2006). For pediatric SDCT, pitches of 1.5 or greater have been recommended for general body scanning (Hollingsworth et al., 2003). By increasing the pitch from 1.0 to 1.5, Paterson et al. (2001) decreased the radiation dose by 33%.

3.5 Shielding of superficial organs

Radiation dose reduction using organ shields was started in the early 2000s. Fricke et al. (2003) studied the amount of dose reduction by using a bismuth breast shield for MDCT of the chest and abdomen in female pediatric patients. The results indicated the shield enabled a 6.7% decrease in the radiation dose to the lungs and a 29% decrease to the breast with no appreciable loss in diagnostic quality. Coursey et al. (2008) assessed the effect of bismuth breast shields on the radiation dose during pediatric chest 16-MDCT. Using this shield with a
tube current of 65 mA, the breast dose was reduced by 26%. In 2007, orbit dose was measured during pediatric cranial MDCT with and without bismuth shielding. The average dose reduction to eyes thanks to bismuth shielding was 42% at 120 kVp (Mukundan et al., 2007). In 2011, eye and thyroid doses were assessed using a bismuth shield in Slovakia. The best reduction in the eye dose due to the use of bismuth shields was within the range of 56–65% and for the thyroid it was 25%. Using an eye shield, some artifacts were observed but the decrease in image quality was not unsatisfactory (Gbelcova et al., 2011).

4 The cancer risk associated with CT radiation

The principal long-term disadvantage of CT is the radiation exposure. It should be noted that the risk of cancer increases linearly with increasing dose until extensive cell killing takes place at very high exposures. The cancer risk depends on both sex and age, with higher risks for females and for those exposed at younger ages. A strong decrease in risk was observed with increasing age (BEIR VII Phase 2, 2005).

Some authors assessed lifetime cancer risks attributable to the exposure in pediatric CT from the value of the received dose (Brenner et al., 2001; Galanski et al., 2006; Paterson and Frush, 2007; Iakovou et al., 2008). Recently, in an epidemiological study, the excess risk of leukemia and brain tumors (with Poisson relative risk models) after CT scans of patients without previous cancer diagnoses who were younger than 22 years were assessed. They declared that for head, chest and abdomen CT compared with doses of less than 5 mGy, the relative risk of leukemia for patients who received a cumulative dose of at least 30 mGy was 3.18 and the relative risk of brain cancer for patients who received a cumulative dose of 50–74 mGy was 2.82 (Pearce et al., 2012).

5 Discussion and conclusions

The IAEA survey shows that use of CT in the 2-year interval from 2007 to 2009 has increased and the lowest frequency of pediatric CT examinations was in European facilities (4.3%). The highest frequency of CT in children was reported in Asia (9.4% in 2007 and 12.2% in 2009) and in Africa (9.6% in 2007 and 7.8% in 2009). The results show that although the total number of CT examinations in children has increased globally, the recommendations on imaging are not always followed in some developing countries (Vassileva et al., 2012).

Although databases for organ doses and EDs in pediatric CT examinations were developed primarily in the 1990s, there is still a critical need to update these values. Moreover, all the studies focused on dose estimation for pediatric reference models, and according to the dependence of the radiation dose on the shape and size of the body, the amount of the dose in non-reference anatomies would be vital. Therefore, the use of NURBS-based hybrid phantoms can help in modeling non-reference subjects and improving the patient-specific dose estimates. Such a vast range of databases can provide more accurate estimation of cancer risk and patient-specific reporting of organ doses due to CT imaging (Xu and Eckerman, 2010).

Although CT scanners have been improved and dose reduction techniques have been introduced, in some countries, exposure of children remains a concern. There is, therefore, a strong need to implement guidelines in pediatric CT examinations and use of alternative examinations. It is in parallel also critical to follow populations of exposed children in well-designed epidemiological studies.

References


