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

P. Akhlaghi, H. Miri Hakimabad and L. Rafat Motavalli[a]

Physics department, School of Sciences, Ferdowsi University of Mashhad, 91775-1436 Mashhad, Iran.

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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,

[a] rafat@ferdowsi.um.ac.ir

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 paid 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 ($CTDI_w$) 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 experimental 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 $CTDI_w$ 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 $CTDI_w$ 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.

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Table 1. Studies that calculated the dose by physical measurements or clinical procedures.

Reference	ED in mSv (mean organ dose in mGy)					CT scanner	CT parameters		Dosimeter	Subject under exposure
	Head	Chest	Abdomen	AP	Trunk		Tube voltage in kVp	Tube loading in mAs		
Fearon and Vucich, 1987	(0.02–5.3)	(0.2–14.8)	(0.08–3.22)	–	(0.32–16)	GE CT/T 9800	120	280	TlD	6-y-old phantom (Humanoid System, CA)
Axelsson *et al.*, 1996	~2 (0.05–37)	–	–	–	~10 (0.4–17)	GE HiSpeed Advantage	120	200 and 340	TlD	1-y-old phantom (manufactured by CIRS)
Ware *et al.*, 1999	–	–	4.4–7.5	–	–	GE HiSpeed Advantage	120	220–290	–	63 patients (under 18 years)
Lucaya *et al.*, 2000	–	1.1–7	–	–	–	CT Twin II-Plus	120	34, 50 and 180	–	25 patients (under 19 years)
Papadimitriou *et al.*, 2000	1.3–2.68	2.83-5.11	9.11–12.12	–	–	Philips Tomoscan LX Serial	100 and 120	142, 237, 190, and 362	TlD, IC[a]	3 cylindrical and 1 elliptical PMMA phantoms
Huda *et al.*, 2000	–	6.4 (50 kg)– 9.6 (10 kg)	–	–	–	GE Hi-Speed CT/i	120	250 and 260	–	44 pediatric patients (under 18 years) with weights 10–50 kg
Huda *et al.*, 2001	4.5–10.7	–	–	–	–	GE Hi-Speed CT/i	120	198–344	–	23 infants (average age of 9 months and average weight of 5.9 kg)
Pages *et al.*, 2003	0.4–2.3	1.1–6.6	2.3–19.9	–	–	5 SDCT and 2 MDCT scanners	120, 137 and 140	18.8–300	IC[a]	PMMA phantom for pediatric patients (1-, 5- and 10-y-old)
Moss and McLean, 2006	1.34–2.34	1.91–7.94	4.73–14.14	–	–	Helical CT scanners (SDCT and MDCT)	114.29–124.71	16.27–226.21	–	8-week-old and 5- to 7-y-old pediatric patients
Galanski *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	100–128	44–327	–	Pediatric patients (under 15 years)
Rybka *et al.*, 2007	0.2 (0.1–7.7) 0.7 (0.3–29.9)	–	–	–	–	SDCT scanner PQ-2000 (Picker)	100 120	150 225	LiF (TLD)	Rando Man phantom (Alderson, USA)
Fujii *et al.*, 2007	–	1.3–7.4 (2–21)[b]	2.8–10.5 (3–16)[b]	–	–	MDCT scanners (8, 16 and 64 detectors)	120	40–160	photodiodes	6-y-old phantom (manufactured by Kyoto Kagaku)
Donadieu *et al.*, 2007	–	1.5–29.3 (2.7–18.6)[b]	–	–	–	GE Light Speed Ultra and Siemens Somatom Plus 4	100, 120 and 140	30–300	–	65 pediatric patients (under 15 years) with cystic fibrosis
Nishizawa *et al.*, 2008	2.6 (18–80)[b]	1.67–8.17 (4.78–24.5)[b]	–	–	–	6 types of MDCT scanners (4 and 16 detectors)	120	25–160 (for chest)	TlD	6-y-old phantom (manufactured by Kyoto Kagaku)
Fahey, 2009	–	–	–	–	9.24–12.41 (5.4–15.95)	Discovery LS; GE Healthcare	120	100	IC[a]	4 phantoms simulating the torso of newborn, 1-, 5- and 10-y-old
Kim *et al.*, 2010	–	–	8.1 (0–1.6)[c] and 37.8 (0–7.2)	–	–	Cone beam CT scanner	125	1340	MOSFETs	5-y-old phantom (ATOM 705-D, CIRS)
Fujii *et al.*, 2011	2.3–2.4 (28–36)[b]	1.7–6.7 (3–11)[b]	–	3.6–7.8 (5–11)[b]	–	2 types of 64 MDCT scanners	120	13–200	RGD[d]	1-y-old child (ATOM Model 704-C, CIRS)
Bernier *et al.*, 2012	0.85 (0.4–50) 1.4 (1–73)	3.05 (0.1–31) 2.2 (1–18)	–	9.55 (4–33) 6.15 (3–28)	–	10 SDCT scanners 12 MDCT scanners	–	–	–	27362 pediatric patients undergoing CT at 14 radiology departments

[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 effect of tube voltage reduction on three cylindrical phantoms of an infant, child and adolescent based on the patient circumference to optimize dose reduction for abdominal CT with no change in image quality. Doses increased by the power function of kVp^n for increases in kVp with n being between 2.49 and 3.12.

Because of image noise and beam-hardening artifacts (which appear as nonuniformities in the CT numbers), sometimes the trend to decrease tube voltage (and correspondingly, the radiation dose) was not successful (Cody *et al.*, 2004; Nakayama *et al.*, 2005). So, 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.

	ED in mSv (mean organ dose in mGy)					CT parameters				
Reference	Head	Chest	Abdomen	AP	Trunk	CT scanner	Tube voltage in kVp	Tube loading in mAs	Phantom	Comments
Huda *et al.*, 1997	1.5–6	–	3.1–5.3	–	–	GE HiSpeed Advantage	120	150–400	Cylindrical water equivalent phantoms	EGS4
Caon *et al.*, 2000	–	2.6–2.8	2.3–2.5	–	7.8–9	GE HiSpeed Advantage	120	100	3 voxel models of 12, 14 (ADELAIDE) and 16 years old	EGS4
Khursheed *et al.*, 2002	–	6.3–7.8 8.8–17.1 11.1–15.8	–	–	–	Siemens DRH GE 9800 Philips LX	120 and 125	230, 320, and 330	5 stylized phantoms from newborn to 15 years (developed by Cristy and Eckerman) with some changes	MCNP
McLean *et al.*, 2003	–	6.33 6.09	8.16 6.88	–	–	SDCT and MDCT (4 detectors)	120	140, 150, 170, and 230	7-y-old child 8-week-old baby	Investigation in 9 radiology departments by dose calculation software (CTEXPO)
Huda and Vance, 2007	0.9–3.6 (30–40)	1.5–4 (7–18)	2–3 (7–15)	–	–	GE HiSpeed Advantage	120	40–280	Uniform cylinders of water modeling head, chest and abdomen	MC dosimetry data
Lee *et al.*, 2007	0.61–1.36 (0–20.14)[a]	3.79–5.7 (0–14.97)[a]	2.81–5.56 (0.01–14.35)[a]	6.05–9.97 (0.01–17.39)[a]	10.92–14.21 (0.08–18.71)[a]	SOMATOM Sensation 16 helical MDCT	80, 100, and 120	100	Ten pediatric phantoms, 5 stylized (ORNL) and 5 tomographic (UF voxel)	MCNPX2.5.
Lee *et al.*, 2012	0.6–1.3 (0–15.9)[b]	4.6–9.9 (0–17.9)[b]	–	6.3–11.4 (0–17.5)[b]	8.8–14.5 (0–18.3)[b]	SOMATOM Sensation 16 helical MDCT	80, 100 and 120	100	Pediatric hybrid phantoms (developed at University of Florida and National Cancer Institute)	MCNPX2.6.

[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.

34 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 tube 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.

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