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# Ionizing Radiation in Medical Imaging and Efforts in Dose Optimization

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## 1. Introduction

Medical-related radiation is the largest source of controllable radiation exposure to humans and it accounts for more than 95% of radiation exposure from man-made sources. Its direct benefits in modern day medical practices are beyond doubt but risks-benefits ratios need to be constantly monitored as the use of ionizing radiation is increasing rapidly. From 1980 to 2006, the per-capita effective dose from diagnostic and interventional medical procedures in the United States increased almost six fold, from 0.5 to 3.0mSv, while contributions from other sources remained static (NCRP report no 160, 2009).

This chapter will review radiation exposure from medical imaging initially starting from a historical viewpoint as well as discussing innovative technologies on the horizon. The challenges for the medical community in addressing the increasing trend of radiation usage will be discussed as well as the latest research in dose justification and optimization.

## 2. Sources and trends

Medical radiation is by far the largest artificial source of population exposure to ionising radiation, accounting for 90% of all doses from artificial sources (United Nations, 2010). This radiation burden is increasing. In the US, this increase has been primarily due to the increasing use of computed tomography (CT). Despite the fact that CT only accounts for 11% of the examinations it contributes 68% of collective dose. In comparison, conventional radiography constitutes 90% of the examinations but only 19% of collective dose (Mettler et al., 2009). From 1980s to 2006, there is an increase of approximately 6-fold in cumulative effective dose per individual in the US from 0.5 to 3mSV (NCRP report no 160., 2009).

As an overall exposure to all kinds of radiation to humans, medical-related radiation exposure now accounts for 48% (an increase from 15%) with background radiation remaining relative static at 50%. The remaining 2% of total radiation exposure came from consumer-related products and activities. These include cigarette smoking, building materials, commercial air travel, mining and agriculture.

| Exposure Category | Collective Effective Dose (person-sievert) | | Effective Dose per Individual in the U.S. Population (mSv) | |
|---|---|---|---|---|
| | Early 1980s | 2006 | Early 1980s | 2006 |
| Ubiquitous natural background | 690,000 | 933,000 | 3 | 3.11 |
| Medical Procedures | 123,000 | 899,000 | 0.53 | 3 |
| Consumer products | 12,000 - 29,000 | 39,000 | 0.05 - 0.13 | 0.13 |
| Industrial, security, medical, educational, and research exposures | 200 | 1,000 | 0.001 | 0.003 |
| Occupational exposures | 2,000 | 1,400 | 0.009 | 0.005 |
| Overall | 835,000 | 1,870,000 | 3.6 | 6.2 |

Data from NCRP report no 160 (2006 data) and no 93 (1980s data).

Table 1. Changes in collective effective dose and effective dose per individual in the US population between early 1980s and 2006.
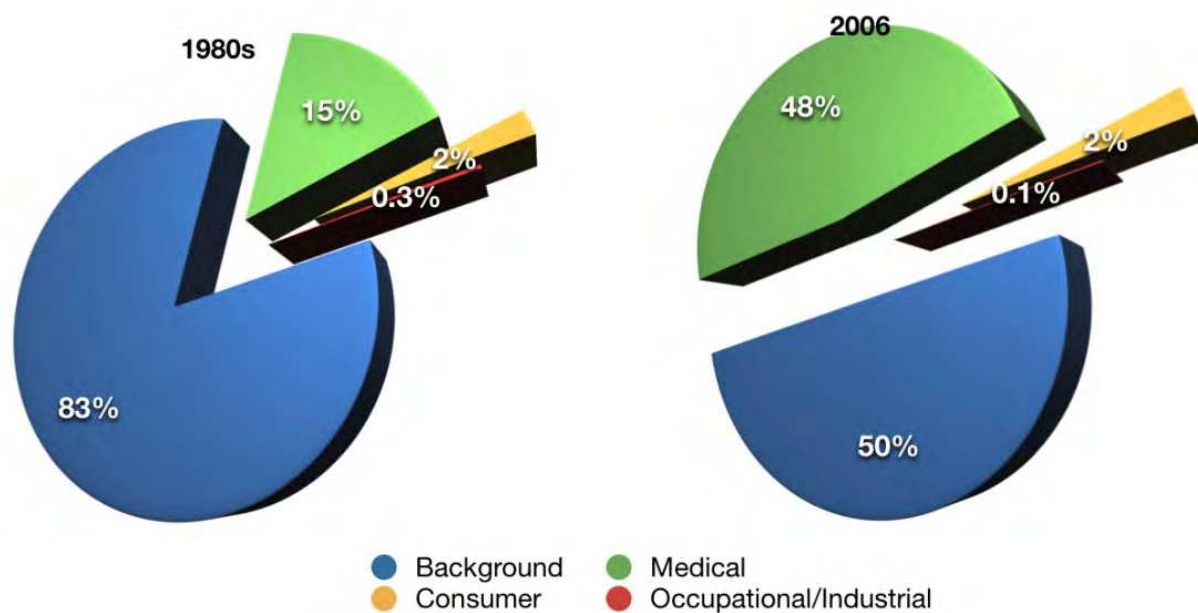


Fig. 1. Exposure of the US population in early 1980s and in 2006 according to NCRP report no 93 and 160.

From data in 2006, in radiology and nuclear medicine procedures, the biggest contribution as previously mentioned is from computed tomography (49%), followed by nuclear medicine studies (26%), then interventional procedures (14%). The remaining 11% came from radiographic and fluoroscopic studies despite accounting for more than 74% of total number of procedures. Not surprisingly, the per capita dose for CT is highest at 1.47 mSv. Data is summarised in the following table.

| Type of Procedure | No. of Procedures (in millions) | Percentage of Total No. of Procedures | Collective Effective Dose (person-sievert) | Percentage of Collective Dose from Procedures | Per-Capita Dose (mSv) |
|---|---|---|---|---|---|
| Diagnostic radiographic and fluoroscopic studies | 293 | 74% | 100,000 | 11% | 0.33 |
| Interventional procedures | 17 | 4% | 128,000 | 14% | 0.43 |
| CT scanning | 67 | 17% | 440,000 | 49% | 1.47 |
| Nuclear medicine studies | 18 | 5% | 231,000 | 26% | 0.77 |
| Total | 395 | 100% | 899,000 | 100% | 3.01 |

Data taken from NRCP report no 160

Table 2. Estimated number and collective effective doses from radiologic and nuclear medicine procedures in the US for 2006.

The overall trend is similar in the UK. A recent Health Protection Agency publication (Hart et al., 2010) showed a 28% increase in 2008 compared to 1997/8 in the United Kingdom. This increase is again mainly due to a doubling of the computed tomography (CT) examinations performed over this same period. The increase in radiation dose in the UK is thought to be modest and this is partly due to the number of examinations performed. A comparison with other European countries show that the 19 out of 20 most commonly performed examinations, UK shows a lower than European average in frequency (Aroua et al., 2010). Only barium enemas were shown to be more frequent than average compared to the European counterparts. In terms of dose, 17 out of 20 examinations in the UK are less on average dose compared to their European counterparts. This illustrates that even in a country like United Kingdom, the trend for increasing use of medically related ionising radiation continues apace despite the fact that there has been more traditional emphasis on

| Country | Annual collective dose per caput (mSv) |
|---|---|
| Germany | 1.52 |
| Belgium | 1.39 |
| Switzerland | 1.37 |
| Iceland | 1.28 |
| France | 1.11 |
| Norway | 0.94 |
| Sweden | 0.59 |
| Netherlands | 0.48 |
| Lithuania | 0.47 |
| Denmark | 0.46 |
| Finland | 0.35 |
| United Kingdom | 0.30 |

Table 3. Collective dose per caput in 2008 for the "top 20" examinations in Europe (from Aroua et al, 2010).

dose awareness. Table below shows annual collective dose per capita of European countries. The UK collective dose per caput from all medical and dental X-ray examinations stands at 0.3 mSv. This number can be compared with corresponding values of 2.2 mSv assessed for medical and dental X-rays in the USA in 2006 (NCRP, 2009) and the similar figure of 1.9 mSv (UNSCEAR., 2010) as the average for people living in Healthcare Level 1 (HCL1) countries. The UK collective dose per caput is clearly very low for a HCL1 country.

For CT, the UK annual per caput dose stands at 0.27 mSv. This number can be compared with corresponding levels of 1.5 mSv from CT in the USA in 2006 (NCRP., 2009) and 0.74 mSv from CT in Canada in 2006 (Chen and Moir., 2010). The UK per caput dose from CT is relatively low compared to North American countries.

At the time of writing, there is likely to be an even further increase in the use of CT compared to the level quoted in 2006. This is mainly due to rapid rise in the use of CT in emergency setting (White and Kuo, 2007; Street et al, 2009) and also the increasing use of PET/CT (Elliot A, 2009; Chawla et al, 2010; Devine et al, 2010). However, the dose increase may not be as high as recent concentrated efforts in the medical community have focused on strategies in dose reduction. Recent effective novel innovations have helped to reduce some of the dose burden (see later).

## 3. Risks

Ionizing radiation has long been known to increase the risk of cancer and is officially classified as a carcinogen by the International Agency for Research on Cancer (IARC). It is now on the official list of carcinogen by the World Health Organization (IARC list of carcinogen, 2011). The exact relationship between dose exposure and cancer induction, however, is complex and several issues merit further discussion.

First, perceived medical-related radiation has traditionally been quantified in comparison with radiation we all receive from background levels. It is worth noting that medical-related radiation exposure is not the same as background radiation. The exposure from background radiation is generally of mixed-energy particles (high and low LET-radiation) while the exposure from diagnostic medical procedures is generally of low-energy x-rays. Low-LET radiation deposits less energy in the cell along the radiation path and is considered less destructive per radiation track than high-LET radiation. Examples of low-LET radiation include X-rays and γ-rays (gamma rays), which are used in medical imaging. Low-LET radiation produces ionizations sparsely throughout a cell; in contrast, high-LET radiation transfers more energy per unit length as it traverses the cell and is more destructive per unit length.

Background radiation are naturally occurring from sources in soil, rocks, bricks and building material, and from radon gas which seeps out to homes. Radon is a colourless, odourless gas that emanates from the earth and as it decays also emits LET radiation. Average annual exposures worldwide to natural radiation sources (both high and low LET) would generally be expected to be in the range of 1–10 mSv, with 2.4 mSv being the present estimate of the central value (UNSCEAR, 2000). Of this amount, about one-half (1.2 mSv per year) comes from radon and its decay products. Some areas will have more radon gas background than others (e.g. Cornwall in the UK generally has twice the amount of background radiation than the rest of the UK).

After radon, the next highest percentage of natural ionizing radiation exposure comes from cosmic rays. This is followed by terrestrial sources, and "internal" emissions. Cosmic rays

are particles that travel through the universe and are filtered by the earth atmosphere (therefore this varies from sea level to higher altitudes where less atmospheric filtration occurs). The Sun is a source of some of these particles. Other particles come from exploding stars called supernovas. The amount of terrestrial radiation from rocks and soils varies in different parts of the world. Much of this variation is due to differences in background radon levels. "Internal" emissions come from radioactive isotopes in food and water with uranium and thorium series of radioisotopes present in food and drinking water (UNSCEAR, 2000).

To quantify medical exposures (which are of predominantly low-LET radiation) to background radiation (which are of mixed-LET radiation) is uncertain due to this respect but it is now generally accepted that of the 2.4mSv total average background radiation of mixed-LET, the total average annual population exposure worldwide due to low-LET radiation would generally be expected to be in the range of 0.2–1.0 mSv, with 0.9 mSv being the present estimate of the central value. The pie chart below illustrates proportions of high and low-LET radiation of the background radiation worldwide (adapted from BIER VII, 2006).
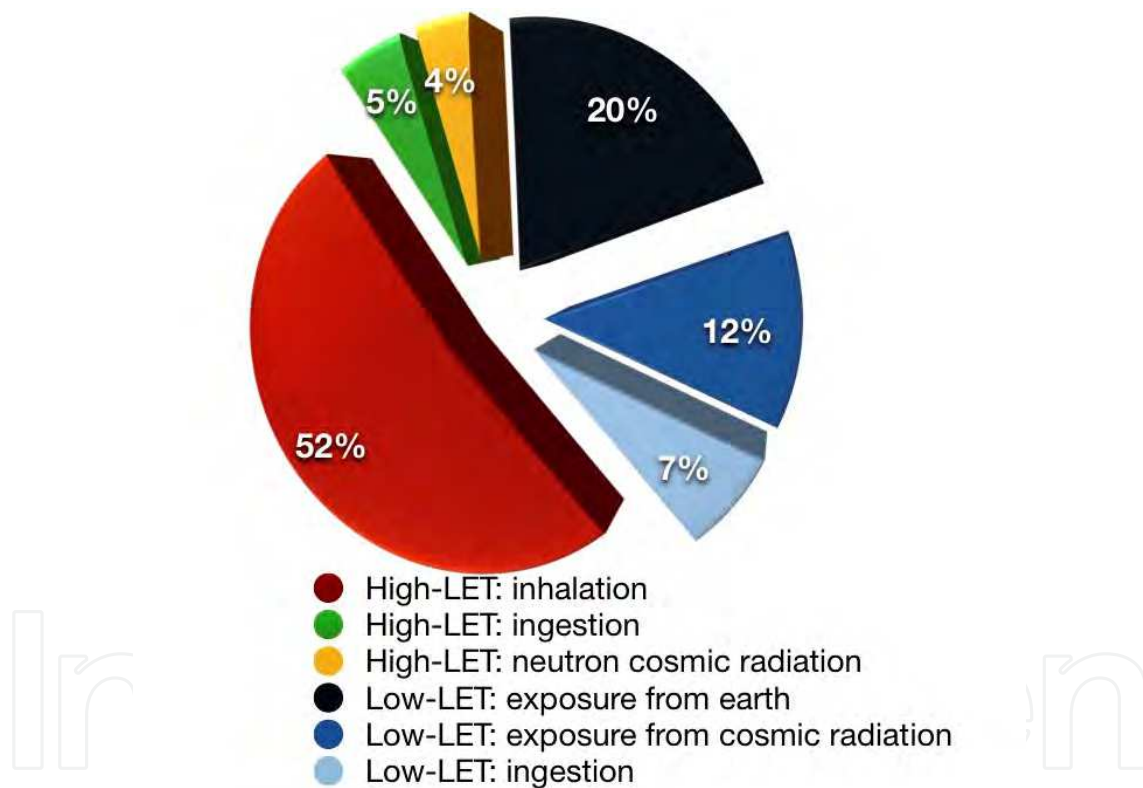


Fig. 2. Pie chart proportions of high and low-LET radiation of the background radiation worldwide (adapted from BIER VII, 2006).

Ubiquitous background radiation represents exposure to whole population (all ages, gender, health status) whereas medical radiation distributions are often skewed towards higher age groups and also sicker individuals. There may also be skewed in gender distribution (there is rightly a reluctance to image patients who are pregnant, for example). Therefore, one must acknowledge that the background radiation level is an approximation and will vary from individual to individual living at different locality having varying exposures.

Current radiation protection standards and practices are based on the premise that any radiation dose, no matter how small, can result in detrimental health effects, such as cancer and genetic damage. Further, it is assumed that these effects are produced in direct proportion to the dose received, i.e., doubling the radiation dose results in a doubling of the effect. These two assumptions lead to a dose-response relationship, often referred to as the linear no-threshold model, for estimating health effects at doses of interest. Although, this is of much benefit in practical terms in risk estimation in the context of radiation protection quantification, there is, however, substantial scientific evidence that this model is an oversimplification of the dose-response relationship. In particular, it results in an overestimation of health risks in the low dose range. Biological mechanisms including cellular repair of radiation injury, which are not accounted for by the linear, no-threshold model, reduce the likelihood of cancers and genetic effects. Therefore, it is now generally accepted that quantification of estimated health risk in the dose range similar to that of background radiation should be strictly qualitative and encompass a range of hypothetical health outcomes, including the possibility of no adverse health effects at such low levels (Burk Jr RJ 1996, updated in 2004).

Biological Effects of Ionizing Radiation (BEIR) VII committee defines low doses of ionizing radiation as less than 100mSv and agrees that at doses of 100 mSv or less, statistical limitations make it difficult to evaluate cancer risk. The current best estimation model for risks associated with low dose radiation exposure is that approximately one individual in 100 persons would be expected to develop cancer (solid cancer or leukemia) from a dose of 100 mSv while approximately 42 of the 100 individuals would be expected to develop solid cancer or leukemia from other causes (BEIR VII, 2006).

It is now generally accepted that at effective doses above 50 mSv the risk of cancer induction increases linearly with dose, this dose-response relation has not been demonstrated at doses below 50 mSv. Below 50 mSv no convincing epidemiological evidence exists currently for cancer risk induction.

There is, however, more convincing evidence to support risk of cancer induction using high dose radiation. A recent article from the USA estimated that CT scans performed in 2007 could result in 29,000 future cancers based on current risk estimations (Berrington de Gonzales et al., 2009). The risk to an individual patient is also high. For example a cardiac scan in a 20-year-old female can produce a lifetime cancer risk from that scan of approaching 1% (Einstein et al., 2007).

Specific effects of radiation are thought to be either deterministic and/or stochastic although the exact relationship is difficult to quantify with certainty. There is reasonable epidemiological evidence (though not definitive) from 30,000 A-bomb survivors that organ doses from 5 to 125 mSv result in a very small but statistically significant increase in cancer risk (Preston et al., 2007). Other low dose epidemiological studies from the occupational exposure of radiation workers are also generally in keeping with this trend with increasing cancer risk with increasing radiation dose (Muirhead et al., 2009 and Cardis et al., 2007). Of note, the third analysis of the National Registry for Radiation Workers in the UK (NRRW-3 from Muirhead et al., 2009) provides the most up-to-date and precise information on the risks of occupational radiation exposure based on cancer registrations as well as mortality from an enlarged cohort of 174,541 workers dating back to 1955. Data were available for 26,731 deaths during 3.9 million person-years of total follow-up period. It clearly shows a statistically significant increasing trend with dose in both mortality and incidence for all

malignant neoplasms (leukaemia and several solid organs tumours). Table below summarises the findings from NRRW-3. Despite this, one must bear in mind that radiogenic excess cancer risk associated with medical-related radiation is in orders of magnitude smaller than the spontaneous cancer risk (Burk Jr RJ 1996, updated in 2004 and BEIR VII (2006).

| | Leukaemia excluding CLL | All neoplasms excluding leukaemia | All malignant neoplasm excluding leukaemia, lung and pleura cancer |
|---|---|---|---|
| 3rd NRRW Analysis | | | |
| *Mortality* | 1.712 (0.06-4.29) | 0.275 (0.02-0.56) | 0.323 (0.02-0.67) |
| *Incidence* | 1.782 (0.17-4.36) | 0.266 (0.04-0.51) | 0.305 (0.05-0.58) |
| | | | |
| 2nd NRRW Analysis | | | |
| *Mortality* | 2.55 (-0.03-7.16) | 0.09 (-0.28-0.52) | 0.17 (-0.26-0.70) |
| | | | |
| 15-country Nuclear Worker Study | | | |
| *Mortality* | 1.93 (0-7.14) | 0.97 (0.27-1.80) | 0.59 (-0.16-1.51) |
| | | | |
| Japanese A-bomb survivors | | | |
| BEIR VII: *Mortality* | 1.4 (0.1-3.4) | 0.26 (0.15-0.41) | - |
| BEIR VII: *Incidence* | - | 0.43 | - |

Table 4. Comparison of estimates of ERR per Sv (and 90% CI) for cancer in NRRW, the 15-country nuclear worker and the Japanese A-bomb survivors.

In addition, there appears to be other determinants that also affect the risk of developing cancers. Among these, genetic considerations, age at exposure, sex and fractionation and protraction of exposure appear to play important roles.

**Genetic considerations**

There are 2 epidemiological studies that suggest that there is a subgroup of population who are more likely to develop cancer when exposed to radiation although in neither case has the genes responsible for the increased radiosensitivity been identified. Ronckers et al (2008) performed a retrospective study looking at 3,010 young women with scoliosis who regularly underwent radiographic follow-up to monitor disease progression between 1912 and 1965. They had found that there was a borderline but significant dose response in a subset of women with a family history of breast cancer in first- or second-degree relatives. Flint-Richter et al (2007) performed a case-control study looking at children who were epilated with x-rays for the treatment of tinea capitis and found that 1% of children developed meningioma with marked clustering in certain families suggestive of genetic susceptibility to the development of tumours after exposure to radiation. A meta-analysis by Jansen-van der Weide (2010) also supports increase risk of radiation in genetically susceptible group. They performed a meta-analysis from seven studies evaluating the effects of low-dose radiation exposure, such as mammography, on cancer risk in women with a familial or genetic predisposition. They had found that there is an increased risk with exposure to radiation that results in 1.3 times increased breast cancer risk. The risk is also higher in women who were exposed before the age of 20 or who were frequently exposed to radiation.

**Age at exposure**

Following on from the last point, there is convincing evidence to support a relationship between life-time attributable risk of cancer incidence and age at exposure. Graph below show analysis of lifetime attributable risk of radiation-induced cancer incidence derived from BEIR VI committee based on data of A-bomb survivors. This, in general, supports that children are more radiosensitive than adults. However, it is also true that social and environmental factors play a role as for some solid cancers, these risks do not decline with age. When looking at effective risks ratio (ERR), there is little difference in risk between 10 and 40 years of age, while for some cancers such as lung and bladder, there appears to be a significant increase in risk with increasing age of exposure.
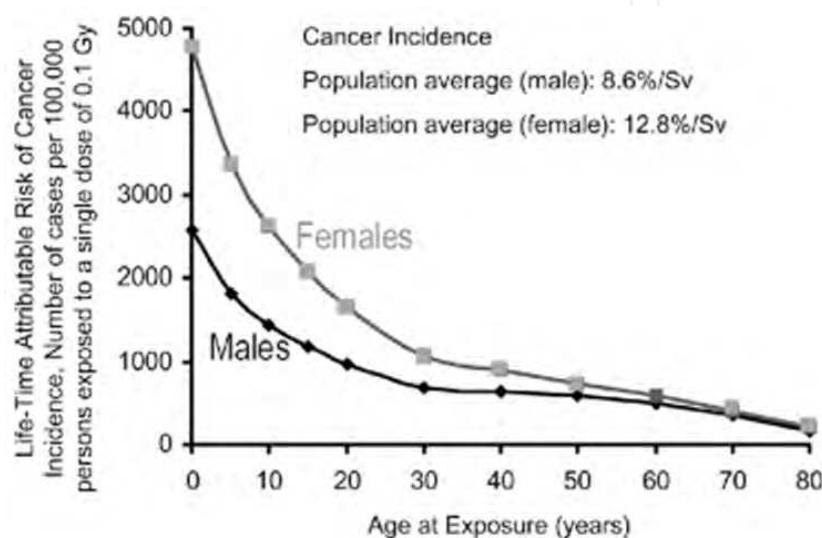


Fig. 3. Lifetime attributable risk of radiation-induced cancer incidence, as a function of age at exposure for males and females (data based on BEIR VII, 2006).
(with permission from Hricak et al, 2011)

**Sex**

BEIR VI report supports the notion that there is substantially higher lifetime attributable risk of cancer incidence in females compared with males. Figure 5 shows breast cancers risks are higher, but what is more notable is the risk for lung and bladder cancer in women is much higher than in men. This is, in spite of the fact that in 1945 Japan, men were heavy smokers while smoking was deemed as very uncommon in women.

**Fractionation and protraction of exposures**

It was previously thought that radiation risks per unit dose at low levels and at low dose rates were smaller than that of higher dose and dose rates. This is due to the perceived influence of DNA repair. A suggested dose and dose rate effectiveness factor (this is a multiplication factor use for low dose rates compared to high dose rates) by the ICRP was two, while BEIR VII suggested a value of 1.5. There are a couple of studies that have already been mentioned but are again worth mentioning in this regard. First the International Agency for Research on Cancer 15-country study looked at around 600,000 nuclear workers who were exposed to an average cumulative dose of 19mSv. The estimated ERR for this cohort for developing solid cancers was almost four times larger than that for the A-bomb
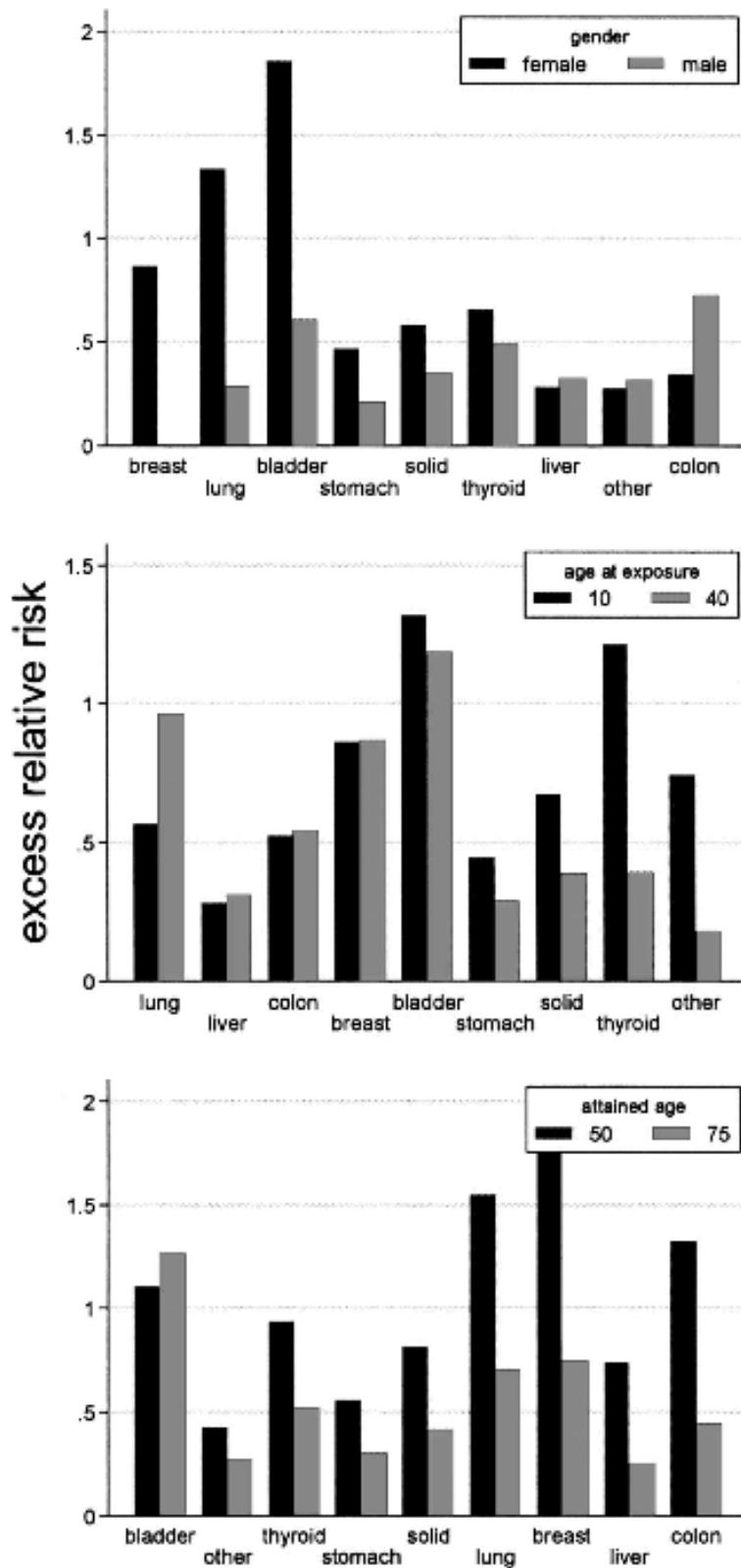
Fig. 4. Comparison of site-specific gender (top), age-at-exposure (middle) and attained-age (bottom) effects on standardised $ERR_{1Gy}$ estimates for selected sites and all solid cancers. (with permission from Preston et al, 2007)

survivors. However, the study noted that there are likely to be important confounders. First, the results are likely to be skewed by the Canadian workers who were relatively few in number with high number of death rates. Second, the predominance of lung cancer suggests possible confounding effect of smoking. More recently, the update from the National Registry for Radiation Workers in the UK (NRRW-3 from Muirhead et al., 2009) followed cohorts with cumulative dose of 25mSv. Cancer risks do increase with dose and the estimated ERR per Sievert was very similar to that for the A-bomb survivor suggestive of a rather small reduction in cancer risk induction with dose protraction which is somewhat surprising. These 2 large studies in themselves suggest that it is not at all clear what the relationship is between dose fractionation and protraction with the risk of cancer induction.

Currently, there are several epidemiological studies following up patients who were exposed to CT at a young age in UK, Australia, Canada, France, Israel and Sweden. Results of such studies will shed more light into the precise influence of radiation and add to the existing body of evidence. This is likely to take time, however, due to the inherent nature of these studies.

## 4. Strategies in dose optimisations in Computed Tomography (CT)

The fact that CT accounts for most of the ionising radiation used in medical imaging is the reason for its focus on this chapter. Other imaging modalities will also be discussed but in lesser detail.

**Dose elimination**

Whilst, undoubtedly medical imaging using ionising radiation has several advantages, one must bear in mind that the best way to reduce radiation is to not perform the investigation at all. There is evidence that increasingly CT has been over utilised in various clinical settings meaning that unnecessary scans are being performed, or incorrect examinations are being performed without appropriate justification. The usual practice is to refer to clinical guidelines or appropriateness criteria to direct or justify an examination according to a clinical scenario. The American College of Radiology (ACR), the Royal College of Radiologists in the UK (RCR), and the European Commission all have published decision guidelines for the appropriate use of CT in different clinical scenarios. A retrospective study was performed in a level I trauma centre looking at appropriateness of scans (Hadley et al, 2006). It was found that 44% of the studies ordered would not have been indicated had the guidelines been rigorously followed. One recent innovative approach addressing this has been to incorporate these guidelines into computerised imaging order entry system. Pre-authorisation of CT examinations according to the ACR and RCR guidelines were utilised in an institution which showed significant deferral rate and substantial decrease in the use of CT and MRI. After reauthorisation was implemented, CT annual performance rates decreased from 25.9 examinations per 1,000 in 2000 to 17.3 per 1,000 in 2003 (Blachar et al, 2006). Despite being evidence-based and recommended for routine clinical usage, these guidelines still show poor uptake in general usage (Bautista et al, 2009). When a survey was performed looking at how physicians decide what the best imaging test to use for their patients, the use of ACR Appropriate criteria showed very low uptake in one institution (2.4%) compared with other available resources (e.g. Radiologist consult, specialty journal, UpToDate, Google, Pubmed, etc).

**Dose reduction**

CT dose reduction can broadly be divided into ways to reduce the total radiation emitted by the X-ray tube and ways to reduce scanning time. Ways to reduce scanning time include ECG gating in cardiac studies, or increasing the pitch of the scanner, for example.

Traditionally, the ways to reduce total dose have included X-ray beam filtration, X-ray beam collimation, X-ray tube current modulation and adaptation for patient body habitus (automatic exposure control), peak kilovoltage optimisation, improved detection system efficiency, low dose protocols for specific indications (e.g. CT KUB for renal stones).

Since the 1980s, a number of technical innovations have been responsible for dose reduction in CTs including the use of solid state scintillating detectors, electronic circuitry, multi-detector arrays, more powerful x-ray tubes and beam-shaping filters. More recently, a number of newer dose reduction techniques have gained widespread acceptance and these are as follows:

**Automatic exposure control**

Automatic exposure control (AEC) is one of the most important techniques in clinical practice to reduce radiation dose without compromising image quality. AEC is a broad term that encompasses not only tube current modulation (to adapt to changes in patient attenuation), but also determining and delivering the "right" dose for any patient (infant to obese) in order to achieve the diagnostic quality images. It is technologically possible for CT systems to adjust the x-ray tube current in real-time in response to variations in x-ray intensity at the detector (McCollough et al, 2005), much as fluoroscopic x-ray systems adjust exposure automatically. The modulation may be fully pre-programmed, occur in near-real time by using a feedback mechanism, or incorporate pre-programming and a feedback loop.

**Tube current modulation**

This is done by maintaining a constant image noise level through longitudinal and/or angular modulation of x-ray tube current according to patient size, shape and the resultant attenuation. This means that the tube current varies across different scan length. This is in contrast to fixed tube current methods where it is constant throughout the scan length meaning in effect that in certain areas, there are wasted radiation as one will not yield increased diagnostic capability.

**Longitudinal (z) mA modulation**

In the longitudinal (z-axis) modulation (AutomA) technique, the basic strategy is to provide predictable image quality to achieve a reliable diagnosis with the lowest necessary radiation dose depending on patient size and attenuation. This is done along the patient's longitudinal axis (i.e. shoulders to pelvis). For a specific patient anatomy and diagnostic task, a specific parameter (defined as noise index - NI) is prescribed by the user to specify the targeted image quality that represents the average noise in the centre of an image of a uniform water phantom. A 5% decrease in NI demands approximately a 10% radiation increase, whereas a 5% increase in NI decreases radiation dose by approximately 10% (Karla et al, 2004a). Therefore, an appropriate NI selection is imperative to control the balance between radiation dose and image quality. This is often recommended by a combination of manufacturer recommendation and clinical experience. This also varies depending on the type of scan performed. With a given NI, the AutomA automatically adjusts x-ray tube current in the scan to maintain the same noise level in all images regardless of patient size and attenuation. Previous studies have shown that in abdominal CT studies a NI of 10.5 to

15 leads to a reduction in radiation dose by 16.6-53.3% in comparison to that using a constant x-ray tube current (Karla et al, 2004b).

**Angular (x,y) mA modulation**

Angular (x,y) mA modulation addresses the variation in x-ray attenuation around the patient by varying the mA as the x-ray tube rotates about the patient (e.g. in the A.P. versus lateral direction). The operator chooses the initial mA value, and the mA is modulated upward or downward from the initial value within a period of one gantry rotation. As the x-ray tube rotates between the AP and lateral positions, the mA can be varied according to the attenuation information from the CT radiograph (i.e. Scout image), or in near real-time according to the measured attenuation from the 180° previous projection.

Using both angular and longitudinal mA modulation, significant dose reduction can be achieved. Although an approximately 50% reduction in dose has been found with automatic exposure control, the system is not foolproof. It seems that at the extremes (i.e. smaller and larger patient sizes), there needs to be adjustment of the noise level such that a higher noise level was recommended for a large sized patient to avoid a higher radiation exposure (Karla, 2004a), and a lower noise level for smaller patients was also suggested by a previous investigation (Karla, 2004b). Some of these studies have shown that there is an influence of patient weight on image quality and dose when a constant noise level is chosen for all patient sizes. Kalra et al (2004b) showed that smaller patients (in their study, defined as having weight less than 68 kg and corresponding to smaller transverse and anteroposterior diameters) had subjective image quality scores lower than larger patients (weights greater than 68 kg) despite using a fixed noise index parameter.

Adjusting noise level based on weight alone is also fraught with difficulties. Several studies have shown that weight is not the ideal factor for required dose calculations. This is because two patients with the same weight can have different regional dimensions and tissue attenuation properties on CT scanning, which can affect the image quality significantly (Karla et al, 2003). Schindera et al (2008) found that a phantom with increased anthropomorphic size received significantly increased skin and deep organ dose than a smaller sized phantom for fixed noise level. Some people have proposed a correction factor for patient size to find the optimal noise index using a combination of patient's weight, BMI or information body diameter from CT scout images (Li et al, 2011).

**Tube Angle Start Position and Pitch**

Various investigators have demonstrated that there are significant dose variations with a sinusoidal pattern on the peripheral of a CTDI 32 cm phantom or on the surface of an anthropomorphic phantom when helical CT scanning is performed, resulting in the creation of "hot" spots or "cold" spots (Svandi et al, 2009). Exploiting this in conjunction with adjustment of pitch can result in dose saving (Zhang et al, 2009). For example, at a pitch of 1.5 scans, the dose is usually lowest when the tube start angle is such that the x-ray tube is posterior to the patient when it passes the longitudinal location of the organ. For pitch 0.75 scans, the dose is lowest when the tube start angle is such that the x-ray tube is anterior to the patient when it passes the longitudinal location of the organ. For organs that have a relatively small longitudinal extent, dose can vary considerably with different start angles. While current MDCT systems do not provide the user with the ability to control the tube start angle, these results indicate that in these specific situations pitch 1.5 or pitch 0.75, small organs and especially small patients, there could be significant dose savings to organs if that functionality adjustment was available.
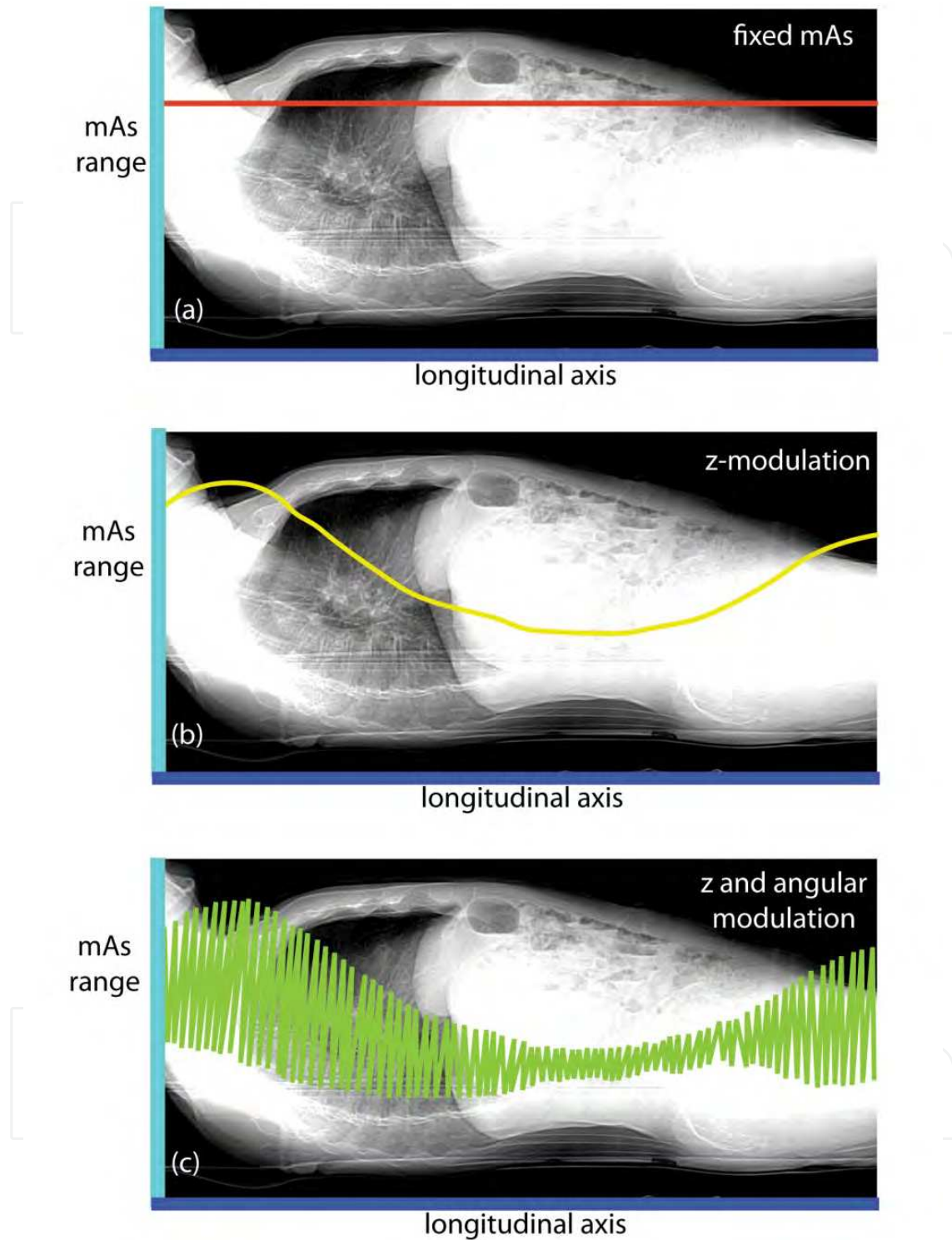
Fig. 5. Diagrammatic illustration of differences between fixed mA (a), z-modulation (b) and combined z and angular modulation (c). Note that the use of fixed mAs means that mA is too high for most parts of the scan. Alternatively, solely relying on z-axis modulation can also mean that tube currents are higher than fixed mA at the edges of the scan where the tube current is increased to account for shoulder girdles and bony pelvis (which inherently have higher attenuation than body scan).
(images are authors' own reproduction, modified from Hricak et al, 2011 with permission)

**Tube voltage**

There have been various studies looking at the adjustment of tube potential. The most reliable determinant when adjusting the tube voltage at present is the patient's BMI but factors such as fat distribution (e.g. around the thorax) should also be considered. The traditional setting for voltage has been at 120kV. Diagnostic images have been shown to be possible at 100 or even at 80kV (Hausleiter et al, 2010). This can be done according to the patient's size with published reports demonstrating up to 70% reduction in chest scans and 40% reduction in abdominal scans (Yu et al, 2010) by lowering the kV in selected groups to 100kV from 120kV. Lowering the voltage decreases penetration (therefore less useful in patients with high BMI) but also results in increased subject contrast as it approximates towards the k-edge of iodine (33 keV). This means that there is superior enhancement of iodine at lower potentials. As a result, there is improved conspicuity of hypervascular or hypovascular pathologies in contrast-enhanced studies as there is improved contrast-to-noise ratio. There is a trade-off in that there is increase image noise mainly due to higher absorption of low-energy photons by the patient. Therefore, this only works best in patients of smaller sizes and the improved contrast-to-noise is negated in larger patients. Dose is proportional to the current but to the square of tube voltage and so it remains more advantageous to lower the kV than the mAs.

**Cardiac CT**

Various innovations related to cardiac CT have markedly reduced the dose of this traditionally high dose examination and are worth discussing in more detail (Roobottom et al, 2010). The 2 main methods of data acquisitions in cardiac CT angiographic studies are inherently linked to the cardiac cycle (ECG-gating). These are retrospective and prospective gating. The former tends to be utilised when the heart rate is greater than 65 beats per minute (bpm) or irregular heart rate and the latter with regular heart rate below 65 bpm. Radiation reduction strategies for these are as follows.

**ECG-linked tube current modulation**

Traditionally, tube current is applied throughout the cardiac cycle in retrospective acquisition. But since the coronary arteries are best image at end diastole where they are most still (usually 75% of the RR interval), the tube current only needs to be maximum at this point. At all other points, the tube currents can be reduced (see Fig. 6 below). Phase data is still available for the rest of the cardiac cycle (although image will be affected by quantum mottle) and functional data can still be obtained. The traditional ECG-gated retrospective spiral acquisition, though results in good image quality, the dose still poses concerns. Thus sequential scanning as an alternative has been developed (Schoenhagen P, 2008).

**Prospective axial gating**

Axial acquisition (rather than spirally) on a pre-defined block (e.g. 4cm) can be used – so called 'step and shoot' technique (Stolzmann et al, 2008) – is a more radiation-efficient technique in that the tube current is only applied in that block and is zero outside that pre-determined window (see Fig 7, below). This results in a dramatic reduction in dose to a level where 5mSv and below can be regularly achieved. The current reconstruction technique limitation, however, determines that the heart rate needs to be below 65bpm. Therefore aggressive beta blockade should be utilised where possible to achieve this (unless there are

contraindications). This technique is also prone to step artefacts. The heart rate also needs to be regular but the use of more 'padding' with centering of the acquisition to include end-systolic (35%) to end-diastolic (85%) phases can combat this to a certain extent. This also allows multi-phase analysis. Expanding phase data, however, increases dose penalty but still remains advantageous as this is still less than the use of retrospective spiral scanning, even with aggressive dose modulation.
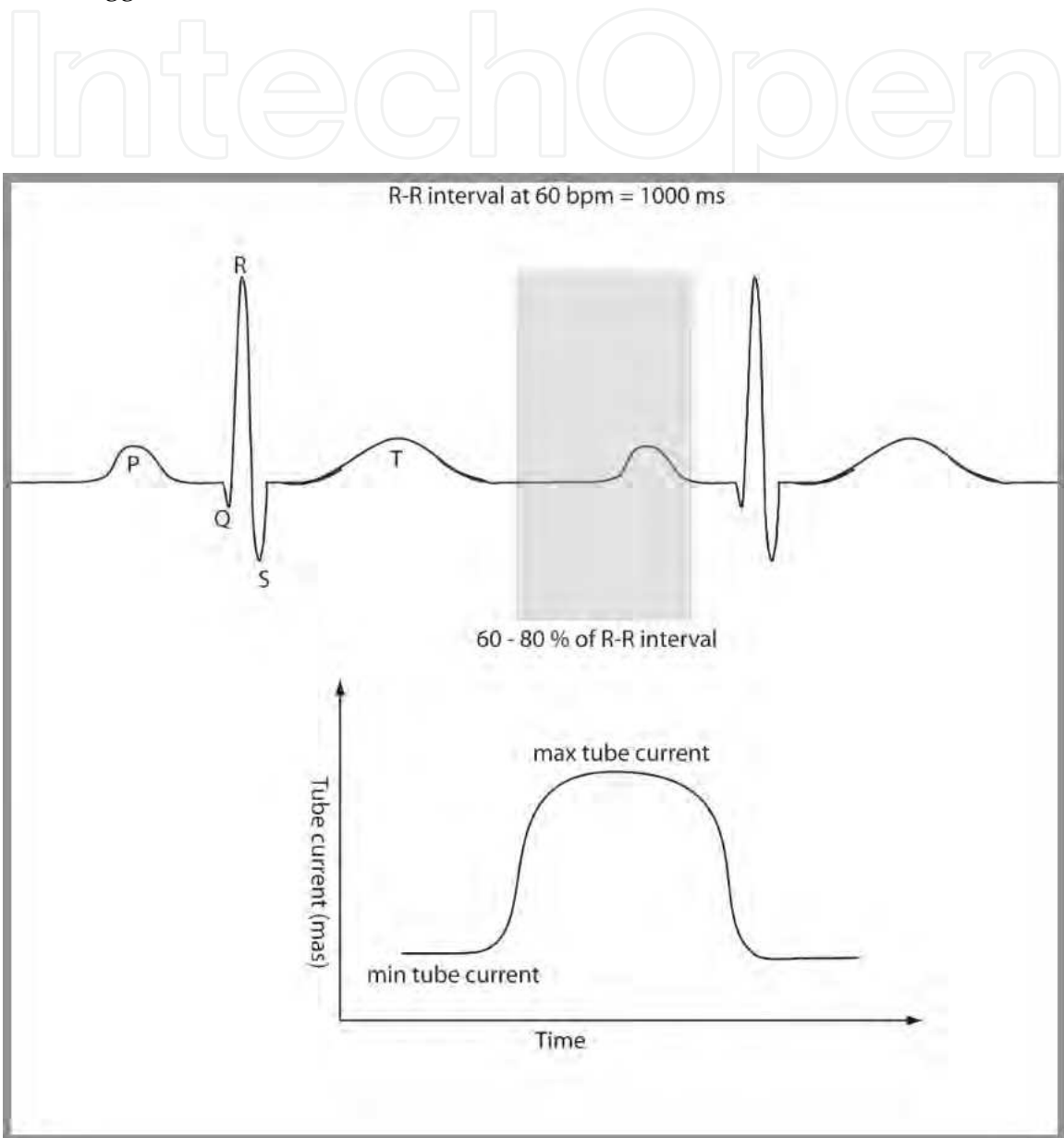


Fig. 6. ECG-linked dose modulation with maximum tube current during diastolic phase and minimum for the rest of the cardiac cycle.
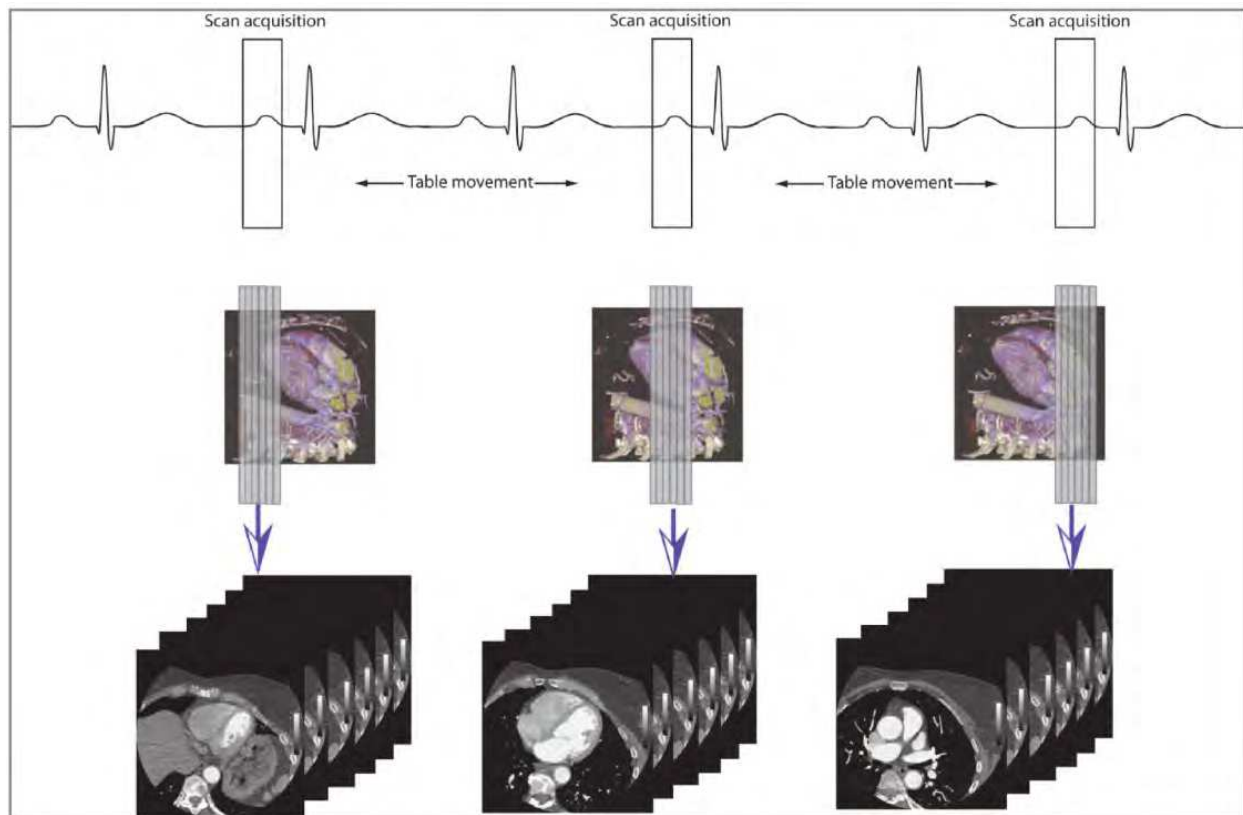(images are authors' own reproduction)

Fig. 7. The "step and shoot" technique with sequential axial scans covering the entire heart. (images are authors' own reproduction)

**Scan length**

Scan length should be kept at a minimum to lower the dose. In general CT, this is done by the operator from the scout view obtained on initial scanning. In certain studies, increase coverage is required but this is difficult to ascertain on the scanned projection radiograph (i.e. scanogram, scoutview or topogram). For example, increase coverage is needed in patients with vein grafts (e.g. internal mammary artery grafts require coverage from lung apex). At our institution, we performed the initial low dose unenhanced scan to look for coronary calcifications which also allows accurate delineation of the coronary arteries. The added information means that we do not have to perform overcoverage (and therefore eliminate unnecessary radiation) and we now routinely perform this at our centre (Roobottom et al, 2010).

**High pitch spiral acquisition**

High pitch ECG-gated acquisition using dual source CT scanner is sufficiently fast enough so that in a patient that has a slow heart rate, images of the whole heart can be acquired within a single heart beat. In particular, cardiac imaging have shown great efforts in dose reduction where a dose reduction of up to 90% has been achieved using novel acquisition techniques (Flohr et al, 2009 and McCollough et al, 2009).
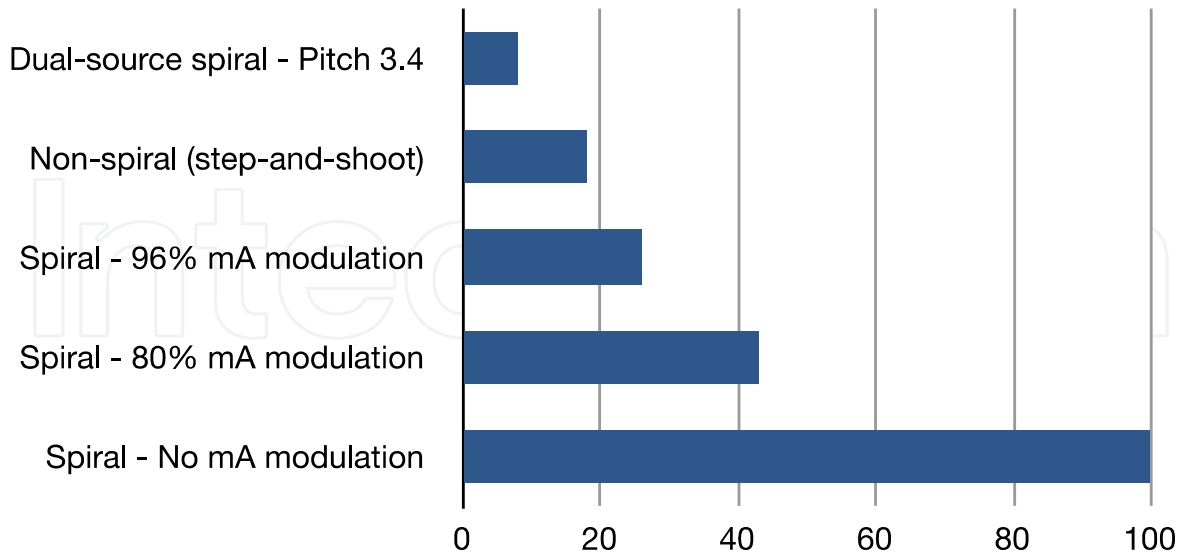
Fig. 8. Graphical illustration showing dose reduction in different techniques of cardiac CT angiography.
(modified from Hricak et al, 2011, with permission)

**Displacement and shielding methods**

There are several types of selective organ shielding available since 1997 (Hooper et al, 1997) with various studies focusing on the use of shields for the eyes, thyroid breasts and gonads. The evidence shows that they reduce dose (especially surface organ dose), however, whether they affect image quality is still a hugely contentious issue.

Breast shielding has been the most studied shielding method and is worth a more extensive mention for this reason. These are made of latex sheets consisting of bismuth and have the property of attenuating x-ray beam in the energy range used in CT. This can be up to about 50% depending on the tube voltage or filtration. The use of bismuth breast shield does not affect the mean energy of the x-ray beam, but reduces the overall intensity. There is fairly consistent agreement throughout the literature on its effectiveness at reducing radiation dose. Fricke et al (2003) found that bismuth breast shields for paediatric CT reduce dose by 29% in anthropomorphic phantoms. Coursey et al (2008) reported a dose reduction of 52% when the bismuth breast shield was employed after the acquisition of scout view using the automatic tube current modulation in a chest MDCT protocol. Foley et al (2011) showed that the use of breast displacement and lead shielding in a group of women undergoing cardiac CT for investigation of chest pain lead to significant reduction in mean breast surface dose without affecting coronary image quality. This varies with breast size but the mean reduction was around 24% of breast surface dose.

Other organ specific shields have also been studied. Mukundan et al (2007) reported a 42% lens dose reduction when using a 2-ply bismuth shield for axial brain and helical craniofacial paediatric CT protocols. Perisinakis et al (2005) also investigated the orbital bismuth shielding technique in children and found the average eye dose reduction of 38% and 33% for CT scans of the orbit and whole head in their Monte Carlo simulations as well as 34% and 20% for the entire and partial eye globe scans in paediatric patients, respectively.

However, despite some evidence supporting the use of shielding, routine use of these shields has been called into question. Geleijns et al (2006) and Vollmar et al (2008) reported that although dose reduction can be achieved with the use of shields, their use is also associated with significant artefacts and increase in image noise. Their findings were supported by Kalra et al (2009) indicating that as well as increasing noise, there is artifactually increased in attenuation values in the region immediately behind the shields. More recently, Lee et al (2011) showed that the use of thyroid shielding with cotton wool spacer reduces the dose to thyroid by up to 27% without affecting image noise although it was noted that there was noticeably increase in the attenuation values of the superficial neck structures such as the neck muscles. It was thought that this was likely related to a metal artefact caused by bismuth implanted within the shield itself.

Some of the streak artefacts in earlier studies can partly be explained by close apposition of the shields to the skin surface. This was the case for both Geleijns et al (2006) and Vollmar et al (2008). When a spacer or plastic shields are used the streak artefacts are noticeably eliminated (Hohl et al 2006, Karla et al, 2009 and Lee et al, 2011). However, the increase in attenuation appears to be a real issue (Karla et al, 2009, Lee et al, 2011). This could have implications when looking at coronary artery calcifications, renal cyst and adrenal mass characterisation, for example, where increase in attenuation is used specifically to characterise disease entities.

Another contentious issue concerns image quality. Although some of the studies have examined the effect on image quality, the robustness of these assessments has also been called into question. In particular, the image quality was not quantified objectively but rather only in qualitative terms with statements such as "we did not see any differences in quality between the shielded and unshielded lung" (Yilmaz et al, 2007). Some investigators have argued that when noise and image quality were analysed objectively and in a robust manner, studies appear to suggest that there is increase in image noise and deterioration of image quality (Geleijns et al, 2006, Vollmarr et al, 2008, Karla et al, 2009). Although, it must be noted that a more recent study by Lee et al (2011) looking at bismuth thyroid shield did not show significant differences in mean noise values. In general, there needs to be more evidence that robustly assess image quality prior to the routine use of shielding to be universally accepted.

Another issue concerns the effect on image reconstruction and wasted radiation. Some investigators argue that bismuth shielding has very different effects on patient dose for the frontal (AP) projection compared to the dorsal (PA) projection (Geleijns et al, 2010). The attenuation of the incoming and exiting beam ultimately has an effect of image quality and dose efficiency. It is argued that if the x-ray tube was in the dorsal position, the exiting x-ray beam is attenuated prior to reaching the detector and therefore this wastes unnecessary radiation, which may have been useful for image formation process. They argue that if the aim of bismuth shield is to reduce low energy photons that mainly deposit their energies on the surface of the patients, then this can be similarly achieved with lowering the tube current without the added artefacts.

The use of bismuth shielding in combination with automated tube current modulation has also been investigated. This is also fraught with danger as there is the potential for either a dose increase to the patient or unequal noise levels within/between images. The reason for this is that the AEC system may attempt to increase the dose to account for the extra patient attenuation. Some of this can be eliminated by placing the shields after the scanned

projection radiograph (i.e. scanogram, scoutview or topogram), however, some manufacturer system performs continual monitoring of patient attenuation and adapts tube current in real time (e.g. when angular modulation is used - see earlier section). If this is the case, then the use of shielding can be detrimental and would actually increase the dose to the patient.
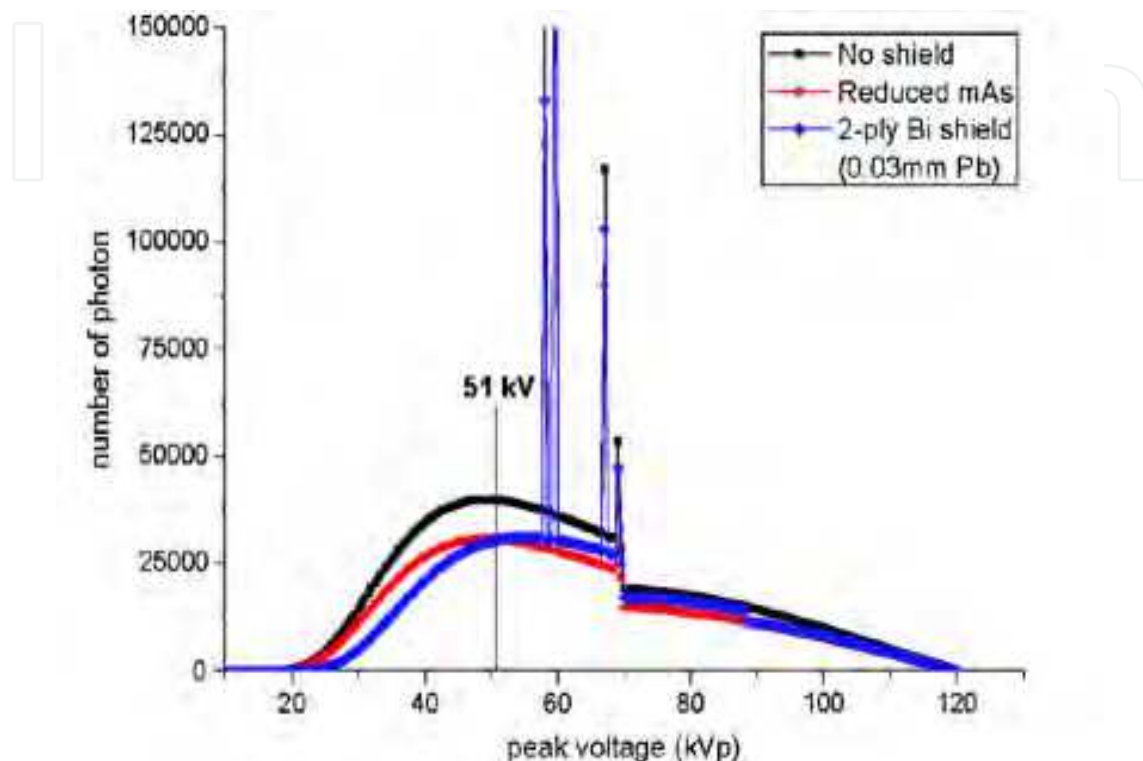


Fig. 9. Comparison of 120 kVp CT beam x-ray spectra with no shield, reduced mAs or 2-ply Bi shield. Note that the reduction of the low-energy photons is substantial in the 2-ply bismuth shield compared to the reduced tube current.
(with permission from Kim et al, 2010)

**Patient factors**

Increasingly, we are moving towards tailoring our examination according to individual patient to optimise image quality and minimising dose. Previous discussion regarding tube current modulation have already shed light on the benefits of this. In general, factors such as age, chest circumference, body mass index, and specific individual factors such as presence of stents and coronary calcification may influence the type of study being performed as well as dose limitation.

**Patient scan length**

Patient scan length is also worthy of consideration. Larger length scans produce larger dose. In the PROTECTION I trial, the average dose for a cardiac CT angiography was 12mSv with the average scan volume of 12cm (Hausleiter et al, 2006)). For cardiac scanning (at least for retrospective scanning), an increase in 1cm results in an increase of approximately 1mSv of added radiation. The volume of acquisition must therefore be tailored to suit each individual patient. Both radiologists and radiographers will need to have active role in selecting this.

**Special patient groups**

**Paediatric patients**

Usual dose reduction strategies such as tube current modulation, ECG-gating, etc as is used in adults can be similarly applied to children with further dose reduction. Adjustment of kV to 80kv is not an unreasonable approach as well as changing mAs values by employing weight-based specific protocols (Ben Saad et al, 2009, Lee et al, 2006, Young et al 2011).

**Women's imaging and pregnancy**

Efforts in dose reduction in pregnant patients owing to dose concerns to the foetus have previously been to not perform the study unless it is absolutely paramount. If this was deemed necessary, then various approaches have been adopted but mostly revolve around the use of shielding (see above). Some specific adjustment of protocols have been tried for certain clinical scenario such as for investigation of pulmonary embolism. Litmanovich et al (2009) compared reduced-dose pulmonary CT angiography (200 mA and 100 kV) with matched control group standard protocol (400 mA, 120 kv). The CT dose index, dose-length product, effective dose, image quality, and signal-to-noise ratio were assessed. There was a significant dose reduction of more than 65% using low dose protocols while maintaining diagnostic imaging quality. Though the dose to the chest has been substantially reduced, the dose to the foetus due to scatter radiation still poses concerns. Others have adopted a more simple approach to tackle this. Danova et al (2010) used lead aprons as shielding to the uterus when scanning thoracic CT achieving up to 34% reduction using the wrap around apron to cover for scatter radiation demonstrating that protective aprons are an effective dose reduction technique without additional costs and little effect on patient examination time.

**Iterative reconstruction**

Out of all the dose reduction strategies discussed, iterative reconstruction shows the most promise. CT workstations have used filtered back projection as the preferred method for producing images from the raw data acquired by the receptors. Filtered Back-Projection (FBP) algorithm has been used as the foundation of commercially available CT reconstruction techniques since the 1970's. Comparatively, it is robust and relatively undemanding on computer processing. It is still widely used today and considered acceptable for clinical diagnosis, but it does not provide optimal results for depiction of the patient as it makes many incorrect assumptions about the data. This is apparent in the inherent level of artefacts and noise in FBP images. To compensate for such noise, larger patient doses are required to overcome this.

Iterative reconstruction algorithms have been put forward as promising recent advances in CT technology but were in fact initially proposed by Shepp and Vardi back in 1982. Only recently, this has been shown to be superior to filtered back projection algorithms for noise reduction (Liu et al, 2007). Even though iterative reconstruction algorithms exhibit great advantages in situations of low signal to noise ratio, their use in real-time CT was previously limited by the time and computing power required to perform the iterations. Recently, however, due to improving computing technology, it is now possible to utilise various facets of iterative reconstruction to reduce the noise and thus achieving significant dose reduction.

- Adaptive Statistical Iterative Reconstruction (ASIR)

Adaptive statistical iterative reconstruction (ASIR) is a post-processing method marketed by GE (General Electric Medical systems, WI, USA) where images are obtained by applying adaptive statistical iterative reconstruction to filtered back projection images. The images are obtained in a low dose mode and the noise in the image is then reduced by applying ASIR. The amount of noise reduction applied can be varied from 1% to 100%. This technique allows modest (up to 40%) dose reduction with similar recorded levels of noise in the image. Several manufacturers have subsequently released similar technology and now all manufacturers are offering iterative reconstruction methods and a means for improved image quality and dose reduction.

There is increasing amount of research to suggest that utilising iterative reconstruction improves image quality, and thereby allows for lowering of kV and mAs and thereby reducing the dose (Prakash et al, 2010). Marin et al (2010) utilized ASIR at low kV and high mAs setting comparing with standard of care FBP technique scanning hepatic organs and found that there is improvement in both image quality and reduction in dose. Similarly, Singh et al (2010) found that abdominal scanning using ASIR compared with FBP yields significant benefit in terms of improved image quality. Due to improved contrast-to-noise ratio, lower radiation parameters can be utilised thereby achieving significant dose reduction. Similar results have been shown using chest CT in comparing ASIR with standard FBP. Leipsic et al (2011) showed that compared with FBP images, ASIR images had significantly higher subjective image quality, less image noise, and less radiation dose with around 30% reduction on average. Studies that have high dose burden such as CT colonography also shows much promise with the use of new ASIR technique. Flicek et al (2010) utilised ASIR in a pilot study using 18 patients undergoing CT colonography using an altered protocol of standard scan 50mAs (supine scan) and 25mAs (prone scan with 40% ASIR). The results show that the radiation dose can be reduced 50% below currently accepted low-dose techniques without significantly affecting image quality when ASIR is used. More recent investigations appear to confirm that using ASIR yields benefits in achieving dose reduction as well as improved image quality. Pontana et al (2011a) studied the utility of iterative reconstruction algorithm in comparison to FBP on 80 patients and found that iterative reconstructions provided similar image quality compared with the conventionally used FBP reconstruction at 35% less dose and also suggested that even higher dose reductions than 35% may be feasible by using higher levels of iterations. Significant noise reduction can also be achieved using the same dose/raw data (Pontana et al 2011b). More and more evidence are appearing in the literature for specific uses of ASIR in specific clinical setting (e.g. CT enterography in Crohn's patients - Kambadakone et al 2011; coronary CT angiography - Leipsic et al 2011) further emphasising its increasing usefulness in quest for improved image quality and dose reduction.

- Model-Based Iterative Reconstruction (MBIR)

Compared with ASIR, newer method so called 'Model-Based Iterative Reconstruction (MBIR)' has now been developed and instead of relying on a single model (as is used in ASIR), performs multiple iterations from multiple models. These models account for a complete three-dimensional assumptions that comprises of focal spot, beam shape, voxel size, and size of detector. In addition, MBIR also accounts for noise from photon flux as well as system noise from the scanner itself.

There is currently no literature research on the practical applications on MBIR as this has only recently been commercialised in December 2010. Preliminary work by the authors on phantoms (unpublished data) have revealed that MBIR shows the most reduction in terms of noise and is superior in terms of objective and subjective image quality and diagnostic confidence compared with ASIR and FBP technique. Dose can be further reduced by up to 80%. Studies are underway to see how image quality compares with traditional methods of FBP and ASIR. From the preliminary work, there is a distinct possibility of achieving body scanning at under 1mSv thus paving a way for significant dose reduction of up to 80% of current levels.
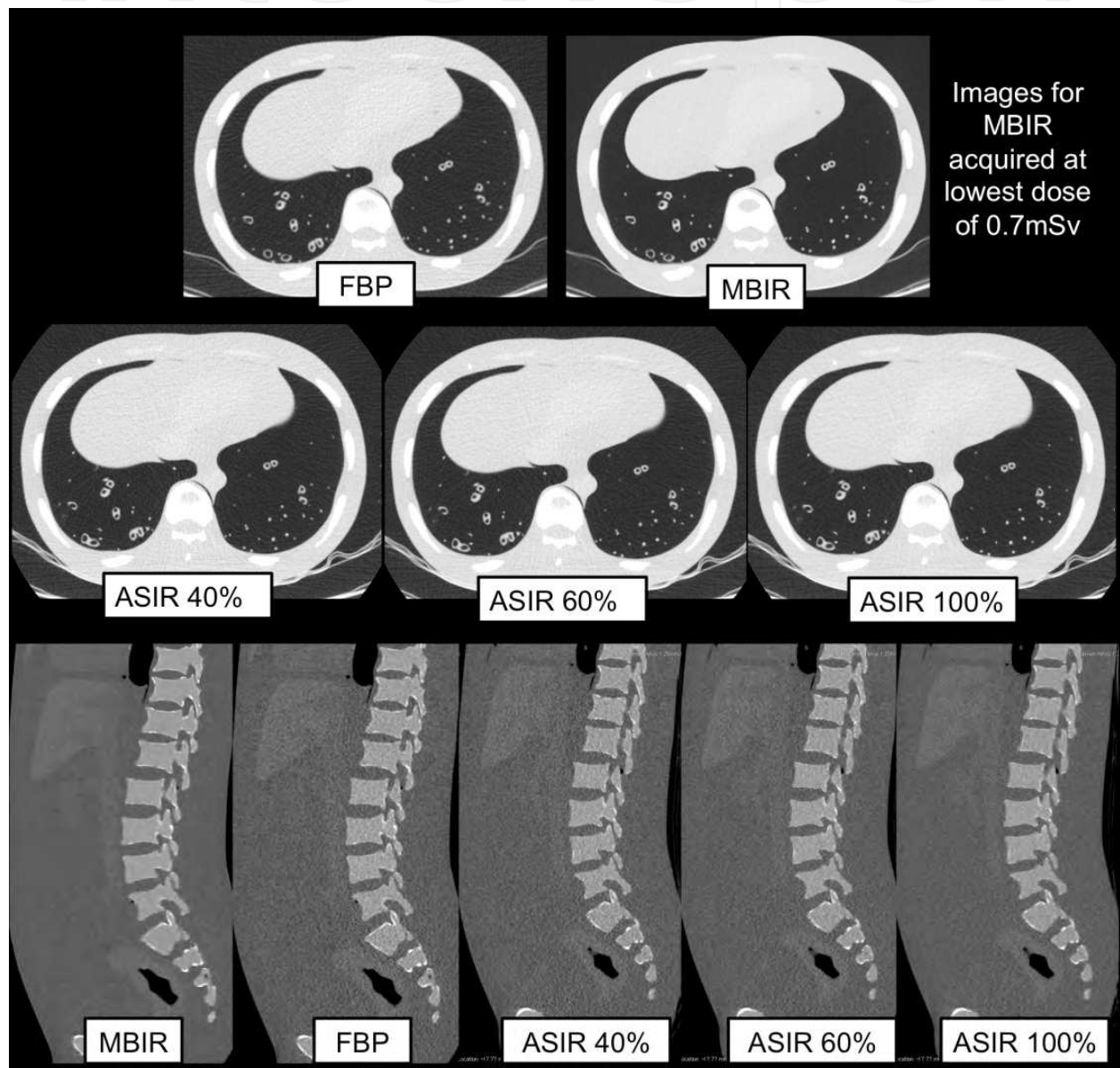


Fig. 10. Side-by-side comparison of scanned torso phantom showing different image quality between traditional filtered back projection, ASIR and MBIR. Note that despite MBIR being acquired at very low dose, this still shows remarkable superior image quality.
(images are authors' own work)

**Garnet technology**

Recently, Garnet-based detectors have been developed which are 100 times faster and have 25% less afterglow compared to the traditional GOS (Gd2O2S)-based detectors. Reduced inherent noise as well as greater contrast resolution should allow for lower voltage/current techniques even for patients with high BMI. Moreover, due to greater contrast resolution, it is hoped that this might aid more accurate detection of in-stent restenosis in cardiac CT angiography – something that is difficult to assess accurately previously due to streak artefact (Cademartiri et al, 2007; Haraldsdottir et al 2010).

**320-Row detector**

With new 320-row detectors now available, axial volumetric scanning of a 16cm segment range in a single 0.35s rotation with an acquisition configuration of 320x0.5mm is now possible. This has the advantage of preventing overranging, and offers high spatial and temporal resolution. This has the potential to reduce the dose considerably. It has been determined that axial volumetric scanning constitutes a dose saving of up to 55% (Kroft et al, 2010; Al-Mousily et al 2011).

## 5. Dose reduction in fluoroscopy

At the turn of the century, there is little awareness of the risk of associated with fluoroscopic procedures, and in particular fluoroscopically-guided intervention. This has in some parts been due to the lack of adequate measuring equipment, complex dose relationships, and false sense of perceived lack of risk. The mood has gradually shifted since 2 key publications in 2004 calling for better dose management for fluoroscopically-guided interventions (Hirshfeld Jr et al 2004; Miller et al, 2004). Since then the FDA in the US have included some of the recommended requirements into new safety-related regulations for manufacturers, and these include features such as last image hold, display of cumulative exposure time, cumulative air kerma, and real-time display of the air kerma dose rate.

Dose delivery in fluoroscopic procedure is complex interactions of numerous factors. There are some excellent review articles (Hirshfeld Jr et al 2004; Steckler et al, 2009) but in-depth discussion is beyond the scope of this chapter. The issue of adequate dose monitoring and effective dose estimations have previously been called to question. The latest dose management technology, which is a feature that is made available on all angiographic equipment in the US, is 'cumulative dose at reference point'. This actually refers to the cumulative 'free-in-air' air kerma and quantify total amount of radiation delivered to a specific point located a fixed distance from the x-ray source. Although, actual dose to patient is a complex, procedure-specific, and varies between operators and is dynamic process (patient's movement during procedure will affect dose received, for example), the cumulative dose at reference point has been shown to correlate reasonably well with the absorbed dose at a specific skin site (Miller et al, 2003).

## 6. Dose reduction efforts in nuclear medicine

In the US, between 1972 and 2005, diagnostic nuclear medicine procedures increased 5- to 6-fold whereas the U.S. population increased by approximately 50%. Between 1996-2005, there was 5% annual growth in the number of nuclear medicine procedures while the growth of the U.S. population has been less than 1% annually. Between 1982 and 2005, the estimated

per capita effective dose from in vivo diagnostic nuclear medicine increased by 550% and the collective effective dose increased by 720% (Mettler et al, 2008). In fact, the estimated 2005 per capita effective dose from diagnostic nuclear medicine (0.75 mSv) is greater than the total per capita dose from both diagnostic radiology and nuclear medicine examinations combined in 1982 (0.14 and 0.40 mSv, respectively). As might also be suspected, the estimated 2005 collective effective dose from diagnostic nuclear medicine (220,000 person Sv) is greater than the total per capita dose from both diagnostic radiology and nuclear medicine examinations in 1982 (32,600 and 92,000 person Sv, respectively).

| Year | Procedures (Millions) | U.S. Population (Millions) | Exams per 1000 Population | Per-Capita Effective Dose (mSv) | Collective Effective Dose (Person-Sv) |
|---|---|---|---|---|---|
| 1972 | 3.3 | 209.9 | 15.7 | | |
| 1973 | 3.5 | 211.9 | 15.6 | | |
| 1974 | | 213.9 | | | |
| 1975 | 4.8 | 216.0 | 22.2 | | |
| 1976 | | 218.0 | | | |
| 1977 | | 220.2 | | | |
| 1978 | 6.4 | 222.5 | 28.8 | | |
| 1979 | | 225.0 | | | |
| 1980 | (5.8 to 6.4) | 226.5 | | | |
| 1981 | 7.0 | 229.5 | 30.5 | | |
| 1982 | 7.55 (7.4 to 7.7) | 231.6 | 32.6 | 0.14 | 32,100 |
| 1983 | | 233.0 | | | |
| 1984 | 6.3* | 235.8 | 26.7 | | |
| 1985 | 6.2* | 237.9 | 26.2 | | |
| 1986 | 6.7* | 240.1 | 27.9 | | |
| 1987 | 6.8* | 242.3 | 28.1 | | |
| 1988 | 7.1* | 244.5 | 29.0 | | |
| 1989 | 7.1* | 246.8 | 28.9 | | |
| 1990 | | 249.7 | | | |
| 1991 | | 252.1 | | | |
| 1992 | | 255.0 | | | |
| 1993 | | 257.7 | | | |
| 1994 | | 260.3 | | | |
| 1995 | 10.2† | 262.8 | 38.8 | | |
| 1996 | 10.5† | 260.3 | 40.3 | | |
| 1997 | 10.9† | 262.8 | 41.5 | | |
| 1998 | 11.8† | 265.2 | 44.5 | | |
| 1999 | 12.6† | 272.7 | 46.2 | | |
| 2000 | 13.5† | 282.1 | 47.9 | | |
| 2001 | 14.5† | 284.8 | 50.9 | | |
| 2002 | 14.9† | 287.9 | 51.8 | | |
| 2003 | 15.7† | 290.8 | 54.0 | | |
| 2004 | 16.5† | 294.7 | 56.0 | | |
| 2005 | (19) 17.2† | 296.0 | 58.1 | 0.75 | 220,500 |

*FDA 1985 Radiation Experience Data 1980, Survey of U.S. Hospitals, DHEW Pub FDA 86-8253, National Technical Information Service, Springfield, VA.
†Data from IMV Inc Benchmark Report 2005 are patient visits, not procedures.

Table 5. Number of Nuclear Medicine Examinations Performed in the U.S. between 1972-2005.
(with permission from Mettler et al, 2008)

There has been a marked shift in the type of procedures with the studies of the brain and thyroid decreasing from a combined percentage of more than 56% of all procedures in 1973 to less than 4% in 2005. The most dramatic increase occurred in cardiac procedures increasing from 1% in 1973 to 57% in 2005. Cardiac studies are relatively high dose procedures and account for more than 85% of the effective dose to the patient population. Currently, more than 75% of all studies fall into 2 categories- cardiac and bone and these 2

types of examinations account for almost 95% of the collective effective dose. In 1982, the estimated number of nuclear medicine procedures was about 7.5 million. The per-capita effective dose from nuclear medicine was 0.14 mSv and the collective dose was 32,000 person Sv. By 2005, the estimated number of procedures had increased to about 19.6 million. The per-caput effective dose increased to about 0.75 mSv and the collective dose to about 220,000 person Sv.

| Procedure | 1973* | | 1982* | | 2005* | |
|---|---|---|---|---|---|---|
| | Number | (%) | Number | (%) | Number | (%) |
| Bone | 125 | (3.6) | 1811 | (24.5) | 3450 | (20) |
| Cardiac | 33 | (1.0) | 950 | (12.8) | 9800 | (57) |
| Lung | 417 | (11.9) | 1191 | (16.1) | 740 | (4) |
| Thyroid | 460 | (13.1) | 677 | (9.1) | — | (<2) |
| Renal | 122 | (3.5) | 236 | (3.2) | 470 | (3) |
| GI | 535 | (15.2) | 1603 | (21.7) | 1210 | (7) |
| Brain | 1510 | (43.0) | 812 | (11.0) | — | (<2) |
| Infection | | | | | 380 | (2) |
| Tumor | 14 | (0.4) | 121 | (1.6) | 340 | (2) |
| Other | 294 | (8.4) | | | — | (<2) |
| Total | 3510 | (100) | 7400 | (100) | 17200 | (100) |

*National Council on Radiation Protection and Measurements.[2]
†IMV is for patient visits. Ratio of visits to procedures is about 1.14.

Table 6. Change (in Thousands) and Percentage of Total Examinations from 1972 in diagnostic nuclear medicine examinations.
(with permission from Mettler et al, 2008)

As with CT, dose reduction efforts have focused on improving technology with innovations such as increased sensitivity of single photon emission computed tomographic (SPECT) instrumentation. This is currently the principal modality for nuclear studies of the heart. Improvement in detectors such as the new cadmium-zinc-telluride detectors will facilitate higher speed/resolution gamma camera and reduce noise (Erlandsson et al, 2009). This equates to higher detector sensitivity, and the hope is that this will generally lead to less administered dose requirement over time (Peterson et al, 2011).

## 7. General notes on issues relating to dose monitoring

Modality-specific metrics are generally used for dose monitoring. These include administered radioactivity for radiopharmaceuticals, entrance skin kerma for x-rays, CT dose index for CT and dose-area product and cumulative air kerma for fluoroscopy. This can be confusing. In an ideal world, radiation safety should be assessed meaningfully across all imaging modalities using a single standard metric for assessment of radiation risk. This should be done not least because it would be more practical for clinicians, but also for ease of risk stratification for patients who are likely to undergo various imaging investigations of different modalities throughout their lifetime. Today, various methods exists, and more are under development to achieve that very goal but at the same time the validity of such endeavour are being questioned as standardised dose specification with a single quantity/unit may not necessarily encompass for both stochastic and deterministic effects. As it stands, no single unifying unit exists for dose monitoring for all medical-related procedure at present.

## 8. Conclusion

In summary, recent efforts in dose reduction in the medical community has arisen due to increasing awareness of radiation-related risks. As we perform more imaging, the risk to the population becomes more concerning. Trends to this effect show that this is increasing at an almost exponential rate. Fortunately, the medical community has made great strides both in utilising new technologies but also aided by closer examination of current practices. Some of the improvements can be implemented right away without much efforts and costs but rely on adherence to already available guidelines and decision making systems. Others rely on the use of new technologies that should make it easier for clinicians to both monitor accurately the dose delivered to the patient but also aid in dose reduction/optimization. In the future, medical imaging is likely to become intertwined with individualised medicine to such an extent that each patient will receive an appropriate test, at an appropriate dose level and according to his or her specific characteristics. Tailoring an investigation to the individual patient should become the norm rather than a one-size-fits-all solution. Several advances are on the horizon and further progress in computational power, new tube design, novel detectors and advances in medical physics and engineering will likely overcome current limitations which can only be beneficial.

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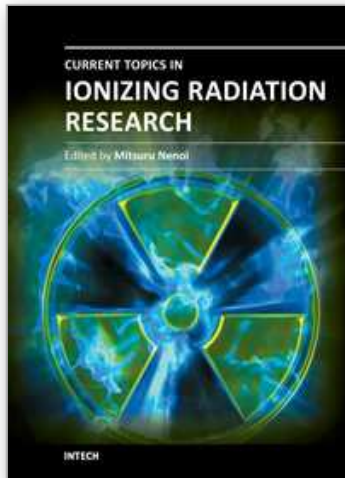
**Current Topics in Ionizing Radiation Research**

Edited by Dr. Mitsuru Nenoi

Since the discovery of X rays by Roentgen in 1895, the ionizing radiation has been extensively utilized in a variety of medical and industrial applications. However people have shortly recognized its harmful aspects through inadvertent uses. Subsequently people experienced nuclear power plant accidents in Chernobyl and Fukushima, which taught us that the risk of ionizing radiation is closely and seriously involved in the modern society. In this circumstance, it becomes increasingly important that more scientists, engineers and students get familiar with ionizing radiation research regardless of the research field they are working. Based on this idea, the book "Current Topics in Ionizing Radiation Research" was designed to overview the recent achievements in ionizing radiation research including biological effects, medical uses and principles of radiation measurement.

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