An efficient Monte Carlo-based algorithm for scatter correction in keV cone-beam CT

G Poludniowski, P M Evans, V N Hansen and S Webb

Joint Department of Physics, Institute of Cancer Research and Royal Marsden NHS Foundation Trust, Downs Road, Sutton, Surrey, SM2 5PT UK

E-mail: Gavin.Poludniowski@icr.ac.uk

Abstract. A new method is proposed for scatter-correction of cone-beam CT images. A coarse reconstruction is used in initial iteration steps. Modelling of the x-ray tube spectra and detector response are included in the algorithm. Photon diffusion inside the imaging subject is calculated using the Monte Carlo method. Photon scoring at the detector is calculated using forced detection to a fixed set of node points. The scatter profiles are then obtained by linear interpolation. The algorithm is referred to as the Coarse Reconstruction and Fixed Detection (CRFD) technique. Scatter predictions are quantitatively validated against a widely-used general-purpose Monte Carlo code: BEAMnrc/EGSnrc (NRCC, Canada). Agreement is excellent. The CRFD algorithm was applied to projection data acquired with a Synergy XVI CBCT unit (Elekta Limited, Crawley, UK), using RANDO and Catphan phantoms (The Phantom Laboratory, Salem NY, USA). The algorithm was shown to be effective in removing scatter-induced artefacts from CBCT images, and took as little as 2 minutes on a desktop PC. Image uniformity was greatly improved as was CT-number accuracy in reconstructions. This latter improvement was less marked where the expected CT-number of a material was very different to the background material in which it was embedded.

1. Introduction

Cone-beam computed tomography (CBCT) has a history of several decades. Feldkamp and co-workers published their ‘practical algorithm’ in the 1980s (Feldkamp *et al* 1984). The topic has received a surge of interest, however, in recent years. This is, in part, due to the availability of high-quality amorphous silicon (a-Si) flat-panel detectors (Seibert 2006). In radiotherapy imaging in particular, it can further be attributed to the availability of CBCT units using keV x-rays, on commercial linear accelerators. Patient scans obtained with treatment-room CBCT devices, provide the potential for re-planning between treatment fractions. A well-recognised impediment to being able to do such on-line re-planning is the presence of scatter pollution within the projection images (Siewerdsen and Jaffray 2001). Scatter leads to inaccuracies in reconstructed CT-numbers, preventing their use in quantitative tasks such as radiotherapy planning. In the last few years there has been a flurry of papers concerning scatter-correction in keV CBCT. The various approaches to the problem can be assigned into two main categories: empirical, using only the image projections, or with supplementary measurements; and scatter simulation, usually using Monte Carlo (MC) methods.

A simple and robust example of the first strategy is the approach taken by the commercial vendor Elekta Limited (Crawley, UK) in their Synergy XVI CBCT system. A uniform scatter-correction is subtracted from each projection image, based on the content of the projection image¹. In most cases such correction strategies yield visually acceptable images, suitable for patient re-positioning, the purpose for which these systems were initially designed. A more sophisticated correction technique is that of Siewerdsen *et al* (2006), who suggested that the scatter at the edges of an image could be measured from pixel-values in the shadow of the collimation, and the scatter elsewhere could be estimated by interpolation. In a very different approach, Marchant *et al* (2008) developed a method for correcting for cupping-artefacts

---

¹ Clinical User Manual for XVI (2006) produced by Elekta Limited (Crawley, UK).

based on the low spatial-frequency information in a planning scan. Kachelrieß *et al* (2006) proposed an empirical cupping correction based on a set of polynomials determined from a one-off scan of a test object. Ning *et al* (2004) have suggested an algorithm based on scatter estimation using (a reduced set of) supplementary projections acquired with a beam-stop array. Zhu *et al* (2006) have demonstrated a frequency-space approach to removing scatter using supplementary projections acquired with a ‘primary modulator’ calibration sheet. Both Rinkel *et al* (2007) and Li *et al* (2008) have developed algorithms for scatter-correction using scatter kernels inferred from projection images acquired through slabs of Polymethyl Methacrylate (PMMA).

The more sophisticated varieties of the empirical approach promise accurate solutions to the scatter problem, but require additional measurements, in some cases increasing patient dose. Further, since they are empirical, the methods are based on measurements with particular scanners and equipment. As such, they cannot be readily used to *predict* the effects of making alterations in a component of the setup and equipment, such as a different detector, geometry changes or changes in the x-ray tube spectrum. Scatter simulation has distinct advantages in these respects. It entails no extra dose to the patient or supplementary measurements and, assuming the underlying physical model is realistic, it can, potentially, provide very accurate results. Further, it allows for the possibility of modelling changes in the components and parameters in the imaging system, to observe the effects on image quality. A number of authors have published papers on simulating scatter in CBCT, such as Malusek *et al* (2005). Where general-purpose MC codes have been used, calculations have been prohibitively slow, due to the large number of photon histories that need to be simulated in each of the many projections. For example, Jarry *et al* (2006) using the EGSnrc code, quote a CPU time of 430 hours. Various techniques have been applied to accelerate the calculation time (Colijn and Beekman 2004, Zbijewski and Beekman 2006, Kyriakou *et al* 2006, Malusek *et al* 2008 and Mainegra-Hing and Kawrakow 2008).

The aim of the work presented in this paper was to produce an original scatter-correction algorithm for CBCT that was fast in execution, with acceptable accuracy, taking, preferably just minutes on a desktop PC, which also retained explicit modelling of various components in the imaging chain, such as the x-ray tube spectrum and the response of the detector. The implementation needed to be applicable to experimental projection data acquired with clinical CBCT units. In reviewing the literature on scatter-correction and the underlying physics, several points seem apparent. Some of these are objective fact, while others are to some degree the opinion of these authors. These points are that:

- 1) Scatter is typically predominantly of low spatial frequency content, lower than the resolution of the detector (see e.g. Mainegra-Hing and Kawrakow 2008);
- 2) Multiple-scattering is important, at keV energies, for a patient-sized objects (see relevant attenuation coefficients: NIST 2005);
- 3) Scattering at keV energies is not strongly forward directed (Chan and Doi 1985);
- 4) The MC method is the best practical choice for calculating the diffusion of particles by wide-angle multiple-scattering;
- 5) The MC method is inefficient at scoring scatter to a detector when scattering is wide-angled because the detector covers a small solid-angle (Colijn and Beekman 2004);
- 6) If scatter-contaminated projections are used to reconstruct, without extra information, such as a planning scan, accurate CT-numbers must be arrived at iteratively, since the first estimate of scatter will be based on incorrect CT-numbers;
- 7) MC estimation of scatter is CPU time-intensive (Jarry *et al* 2006) and becomes increasingly so as the number of voxels describing a subject is increased, as does the CT reconstruction time;
- 8) Poly-energetic x-ray beams used in keV CBCT are subject to beam-hardening (Ding *et al* 2007); this, combined with the energy dependent response of the detector (Roberts *et al* 2008), complicates the relationship between measured pixel-value and the inference of radiological thickness.

It is in the context of these eight observations that the algorithm presented in this work was developed. Photon diffusion is handled by Monte Carlo techniques (2nd, 3rd and 4th

points) using a combination of quasi- and pseudo-random numbers. Scatter scoring is done using ‘fixed forced detection’, explained in the Method section, rather than probabilistically (5th point). The algorithm is iterative (6th point). At each iteration, scatter estimates are obtained and these current estimates are subtracted from the original measured data prior to CT reconstruction. Except in the final reconstruction, which occurs upon convergence, the reconstructions are performed on a coarse matrix of voxels, to accelerate reconstruction and subsequent scatter calculations (7th point). This coarse reconstruction is justified by the low spatial-frequency content of the scatter (1st point). The spectrum of the x-ray tube and the response of the detector are modelled to account for poly-energetic effects (8th point).

This correction algorithm will be referred to as the Coarse Reconstruction and Fixed Detection (CRFD) technique. In this work we introduce CRFD in detail, validate it, and demonstrate the algorithm’s efficiency using experimental cone-beam acquisitions obtained using a Synergy XVI CBCT unit (Elekta Limited, Crawley, UK).

2. Method

2.1 The CRFD algorithm

To implement CRFD, a computer program was written in the F subset of Fortran 95, and compiled on a Windows XP OS (Microsoft Corporation) using the Intel Visual Fortran compiler v11.0 (Intel Corporation, Santa Clara, USA). Aggressive optimization was applied using the ‘fast’ compilation option. All simulations were performed on a desktop PC with a single 3.4 GHz Pentium 4 CPU and 2 GB of RAM. The implementation can be summarised in seven steps:

- 1) *Signal-to-thickness conversion*. Convert the pixel-values to effective radiological thicknesses, in cm of water;
- 2) *Coarse reconstruction*. Filter and back-project the data to reconstruct a coarse matrix of voxels of ‘water-equivalent’ densities;
- 3) *Material partition*. Partition the voxels, based on densities, into materials;
- 4) *Scatter simulation*. Calculate the scatter-signal at fixed ‘nodal’ points across the detector, for a number of projections over a 2π source rotation, using a purpose-written MC code;
- 5) *Signal-correction*. Subtract the estimated scatter-signal from the measured signal, at each pixel;
- 6) *Iteration*. Repeat Steps 1) to 5) until satisfied with convergence;
- 7) *Final image*. Perform final image reconstruction at full resolution using the final estimate of scatter.

Step (1) was introduced to account for beam-hardening effects. At the initialisation stage of the program, before proceeding to reconstruction, a look-up-table (LUT) was created to translate signal in the detector to radiological thickness in cm of water. This LUT was created using models for the x-ray source and the detector response, assuming that the intervening material was only water. The x-ray tube spectrum model was that of Poludniowski and Evans (2007) and Poludniowski (2007) and was calibrated using half-value layer (HVL) measurements from routine quality assurance. The spectrum is shown in figure 1 (a), based on a HVL measurement of 7.0 mm Al, an assumed central-axis anode take-off of 17.5° and a tube potential of 120 kV. The XVI flat-panel detector is based on a-Si technology and uses indirect detection via a thallium-doped CsI scintillator. The detector energy-response model was that simulated by Roberts *et al* (2008): the data points and fit are shown in figure 1 (b). Note that the mean energy of the 120 kV beam (61 keV) is closely matched to the peak sensitivity of the detector. The presence of a patient (or phantom) modulates the energy spectrum, however: the primary beam is hardened while the scattered radiation may have a lower mean energy (Ding *et al* 2007). The non-linear response of the detector demands that it be explicitly modelled. Here, the panel response was incorporated in the MC on a photon-by-photon basis. For example, each 40 keV photon reaching the detector was assumed to deposit 25 keV of energy as signal, by reference to the average-response curve of figure 1 (b).

Steps (2) and (7) were performed using a cone-beam filtered back-projection program. The reconstruction algorithm used was that of Feldkamp *et al* (1984), using the formulation of Kak and Slaney (1988) and linear interpolation of ray-projections. The quantity back-projected, however, was the radiological thickness (rather than the natural logarithm of the ratio of pixel values). The reconstructed quantity was then water-equivalent (*weq*) density. This quantity, denoted ρ_{weq} , can be related to a material's physical density, ρ_i , by,

$$\rho_{weq} = \rho_i \left\langle \left(\frac{\mu}{\rho} \right)_i (E) \right\rangle_E / \left\langle \left(\frac{\mu}{\rho} \right)_{water} (E) \right\rangle_E, \quad (1)$$

where $(\mu/\rho)(E)$ is the mass attenuation coefficient for a photon energy of E and the $\langle \dots \rangle_E$ bracket denotes an average over the x-ray energy spectrum. Prior to the reconstruction, a 2D 5-pixel median filter was applied to the raw projection data to reduce noise. To further suppress noise in the reconstructed image, a Hamming filter was applied to the projection-data in frequency space before back-projection. In cases where, in an acquisition, the detector panel was offset from the central axis to extend the field-of-view, the pre-convolution scheme of Cho *et al* (1996) was used to weight redundant ray-projections. A limitation of reconstructing based solely on acquired projections, is that the reconstructable object is smaller than the extent of object irradiated. The reconstructable region takes the shape of a cylinder with cones capping both ends. Since scatter will contribute from all regions irradiated, a strategy must be decided on to deal with the 'extra' regions. Here, an expanded cylinder is reconstructed, to encompass the entire irradiated volume, with missing ray-projections estimated by the closest rays available. This approximation proves to be adequate and besides, the pre-existing non-exactness of the Feldkamp reconstruction, based on a single circular source trajectory, should be noted (Tuy 1983).

The material partition of Step (3) was carried out based on reconstructed *weq* densities at each iteration step. In reconstructions of a RANDO phantom (The Phantom Laboratory, Salem NY, USA), the partitioning followed the scheme presented in table 1. Material compositions were taken from ICRU Report 44. The RANDO phantom is composed of a human skeleton embedded in a soft-tissue equivalent plastic of uniform composition and density, and lung cavities of a lower density. In reconstructions of a Catphan phantom (The Phantom Laboratory, Salem NY, USA), the material was simply designated as air if $\rho_{weq} < 0.2 \text{ g cm}^{-3}$ and water, otherwise. The Catphan phantom is a 20 cm diameter cylindrical phantom consisting of several sections embedded within a close to water-equivalent plastic, enabling the evaluation of various image quality measures. If a material was assigned as air, then its density was reset to zero.

Table 1. Material partition for reconstructions of the RANDO phantom.

	$\rho_{weq} < 0.1$	$0.1 \leq \rho_{weq} < 0.5$	$0.5 \leq \rho_{weq} < 1.1$	$\rho_{weq} \geq 1.1$
	g cm^{-3}	g cm^{-3}	g cm^{-3}	g cm^{-3}
Material	Air	Lung	Soft-tissue	Cortical bone

The scatter-estimation of Step (4) used MC photon diffusion and scoring by forced detection. The scoring technique is detailed in the next sub-section. The details of the MC are described in the penultimate subsection of this section. The result of the simulation, however, is that the scatter signal is obtained for a set of equally-spaced nodes on the flat-panel detector, as illustrated in figure 2 (a). The scatter signal in a particular pixel slowly varies with changing source angle (Ning *et al* 2004). This justifies the calculation of scatter projections at a reduced number of source orientations.

The signal-correction, Step (5), begins after the set of scatter-projection simulations has been completed. The scatter signal at every pixel is estimated using trilinear interpolation of the simulated data: a bilinear spatial interpolation in the detector plane and linear angular interpolation between simulated view-projections. It was possible that, in certain cases, the scatter could be overestimated. To constrain the estimate of primary signal, after scatter subtraction, to be positive and non-zero, the estimated primary signal, in any pixel, was not allowed to fall below a pre-determined minimum. That minimum was set to a fraction 0.001

of the open-field signal. This limit, corresponding to 99.9% attenuation, is equivalent to in excess of 30 cm of water for the keV CBCT x-ray beam used in this work. In practice, the first estimation of scatter was found to be excessive, due to the lower self-attenuation caused by reduced initial CT-numbers. A fraction of one-half of the calculated scatter (arrived at by trial-and-error) was subtracted from the projection data in the first iteration only, to accelerate convergence. Convergence was determined manually in this work, although a quantitative measure could be introduced into the algorithm if desired.

Figure 1. The (a) x-ray spectrum and (b) detector energy-response used as inputs into the CRFD algorithm.

2.2 Fixed forced detection

Forced detection (FD) denotes the idea that we require a theoretical interacting photon to make a contribution to the signal in a detector, regardless of the possibility that its interaction history would lead it elsewhere. Fixed FD (FFD) demands that forced detection is applied to a fixed set of points at a detection plane. The FFD problem can then be posed thus: given that a photon of energy, E , and direction cosines, $\mathbf{U}_{int}$, interacts at a point, $\mathbf{r}_{int}$, what is the probability that it is scattered to the n th pixel centred at $\mathbf{r}_{det}[n]$? The geometry is illustrated in figure 2 (b). Phrased in this way, the problem is decoupled from the mechanics of how the photon arrived at that point, which may be by MC diffusion, or otherwise. If the pixel area is ΔA , the element of solid-angle covering that pixel is,

$$\Delta\Omega_{det}(\mathbf{n}_{det}, \Delta\mathbf{r}_{det}[n]) = \frac{|\mathbf{n}_{det} \cdot \Delta\mathbf{r}_{det}[n]|}{|\Delta\mathbf{r}_{det}[n]|^3} \Delta A, \quad (2)$$

where $\mathbf{n}_{det}$ is the a unit vector normal to the detector panel, ΔA is assumed small enough and $\Delta\mathbf{r}_{det}[n] \equiv \mathbf{r}_{det}[n] - \mathbf{r}_{int}$. (3)

In the limit that ΔA tends to zero, that is, for a point, it is necessary to re-write this as,

$$\frac{d\Omega_{det}}{dA}(\mathbf{n}_{det}, \Delta\mathbf{r}_{det}[n]) = \lim_{\Delta A \rightarrow 0} \left(\frac{1}{\Delta A} \Delta\Omega_{det}(\mathbf{n}_{det}, \Delta\mathbf{r}_{det}[n]) \right) = \frac{|\mathbf{n}_{det} \cdot \Delta\mathbf{r}_{det}[n]|}{|\Delta\mathbf{r}_{det}[n]|^3}. \quad (4)$$

Both equations 2 and 4 are invariant under coordinate system rotations. The problem requires the consideration of multiple frames-of-reference, however. Imagine three frames denoted by the superscripts l , f and d , which are the laboratory, photon-flight and detector frames, respectively. The frames share a common origin, which is chosen to be the isocentre. The detector-frame is defined such that,

$$\mathbf{n}_{det}^d = (0,0,1). \quad (5)$$

The flight-frame is defined such that the direction-cosines vector for the initial photon is along the positive z-axis: i.e.

$$\mathbf{U}_{int}^f = (0,0,1). \quad (6)$$

The scattering angle to a nodal point, $\theta_{det}^f[n]$ may then be found using,

$$\cos \theta_{det}^f[n] = \frac{\hat{\mathbf{z}} \cdot \Delta \mathbf{r}_{det}^f[n]}{|\Delta \mathbf{r}_{det}^f[n]|}, \quad (7)$$

where $\hat{\mathbf{z}}$ is the unit vector in the z-direction of the flight frame. Crucially, the probability-of-scatter will depend on $\theta_{det}^f[n]$. However, typically, the point-of-detection will be specified in the detector frame ($\mathbf{r}_{det}^d[n]$) and the interaction point in the laboratory frame ($\mathbf{r}_{int}^l$). Let us define the position of the centre of the n th pixel in the detector frame to be,

$$\mathbf{r}_{det}^d[n] = (x^d, y^d, SDD - SAD), \quad (8)$$

where x^d and y^d define the position in the detector plane and SDD and SAD are the source-to-detector distance and source-to-axis distance, respectively. Any vector in one frame may be converted to that in another by the use of a 3D rotation matrix. For example, M^{df} , may convert a vector in the detector frame, to the flight frame:

$$\mathbf{r}_{det}^f[n] = M^{df} \mathbf{r}_{det}^d[n]. \quad (9)$$

Therefore, the vector between the photon and a detection point, in the flight-frame, is

$$\Delta \mathbf{r}_{det}^f[n] \equiv \mathbf{r}_{det}^f[n] - \mathbf{r}_{int}^f[n] = M^{lf} (M^{dl} \mathbf{r}_{det}^d[n] - \mathbf{r}_{int}^l[n]), \quad (10)$$

and furthermore,

$$\mathbf{n}_{det}^f = M^{df} \mathbf{n}_{det}^d. \quad (11)$$

These formulae allow us to calculate both equations 4 and 7 working in the flight-frame. The probability-density for a photon of energy, E , given that it interacts at $\mathbf{r}_{int}^l$, to Compton scatter to the n th node at $\mathbf{r}_{det}^d$, is then,

$$q_{CS}[n] = p_{CS}(E, \mathbf{r}_{int}^l) \exp(-\mu_{water}(E)L(E')) \frac{df_{CS}}{d\Omega}(E, \theta_{det}^f[n]) \frac{d\Omega_{det}}{dA}(\mathbf{n}_{det}^f, \Delta \mathbf{r}_{det}^f[n]), \quad (12)$$

where p_{CS} is the probability that the interaction is a Compton scatter (the dependence on $\mathbf{r}_{int}^l$ enters because of material-variations over space); μ_{water} is the attenuation coefficient of water; $L(E')$ is the radiological escape length of the scattered photon of energy, E' ; and $df_{CS}(E, \theta)/d\Omega$ is the angular distribution of Compton scattering. The energy of the photon after scattering, E' , is given by the Compton relation,

$$E'[n] = \frac{E}{1 + \frac{E}{m} (1 - \cos \theta_{det}^f[n])}, \quad (13)$$

where m is the mass-energy of an electron. Similarly, the probability-density that the photon scatters to the n th node via the Rayleigh mechanism, is,

$$q_{RS}[n] = p_{RS}(E, \mathbf{r}_{int}^l) \exp(-\mu_{water}(E)L(E)) \frac{df_{RS}}{d\Omega}(E, \theta_{det}^f[n]) \frac{d\Omega_{det}}{dA}(\mathbf{n}_{det}^f, \Delta \mathbf{r}_{det}^f[n]) \quad (14)$$

where p_{RS} is the probability of a Rayleigh scatter, and, in this case, the photon energy remains unchanged throughout the collision process. The radiological path-length between the photon and node, through the phantom, is

$$L(E) = \frac{1}{\mu_{water}(E)} \int_0^{|\Delta \mathbf{r}_{det}^l|} d\lambda \left(\frac{\mu}{\rho} \right) (E, \mathbf{r}^l(\lambda)) \rho(\mathbf{r}^l(\lambda)), \quad (15)$$

where λ is the actual path-length of the scattered photon; $(\mu/\rho)(E, \mathbf{r}^l(\lambda))$ is the mass attenuation coefficient for a photon of energy E at position $\mathbf{r}^l(\lambda)$; and $\rho(\mathbf{r}^l(\lambda))$ is the density of material at that position. The vector, $\mathbf{r}^l(\lambda)$, is parameterized as,

$$\mathbf{r}^l(\lambda) = \mathbf{r}_{int}^l + \frac{\Delta \mathbf{r}_{det}^l[n]}{|\Delta \mathbf{r}_{det}^l[n]|} \lambda. \quad (16)$$

This constitutes an explicit solution to the FFD problem. What remains is to translate the probability-density into signal in the detector. If the photon energy can be assumed to be deposited at the point of incidence on the flat-panel, according to a known energy-response

function, $R(E)$, then the increment in signal at the n th node due to the k th interaction in the phantom, of the p th photon history, is,

$$\Delta S[n, p, k] = q_{CS}[n, p, k]R(E'[n, p, k]) + q_{RS}[n, p, k]R(E[p, k]). \quad (17)$$

Thus, the total signal for that node, for N photon histories and $I[p]$ interactions, is

$$S[n] = \sum_{p=1}^N \sum_{k=1}^{I[p]} \Delta S[n, p, k]. \quad (18)$$

Once the N histories have been simulated and the signal at the nodes summed, these signals can be normalized by their open-field values, S_0 , which are assumed to be given by,

$$S_0[n] = \frac{N}{\Omega_{cone}} \frac{d\Omega_{det}}{dA} \left(\mathbf{n}_{det}^d, \mathbf{r}_{det}^d[n] - \mathbf{r}_{src}^d \right) \int_0^\infty R(E)P(E)dE, \quad (19)$$

where $P(E)$ is the probability-density that an emitted photon has energy, E , Ω_{cone} is the solid-angle of the radiation cone and $\mathbf{r}_{src}$ is the x-ray source location where,

$$\mathbf{r}_{src}^d = (0, 0, -SAD). \quad (20)$$

The normalized signal at the n th node, $S_{norm}[n]$, is then

$$S_{norm}[n] = \frac{S[n]}{S_0[n]}. \quad (21)$$

2.3 Scatter simulation

The interaction volume (a cuboid of voxels) was sampled using MC photon diffusion. MC code writing for photons in the keV x-ray range is a well-established art, see for example Chan and Doi (1983). Only essential details of the implementation used in this work are quoted here. The x-ray photons leaving the x-ray tube were assumed to originate from a point source. The initial photon direction was sampled isotropically within the prescribed cone of radiation. The initial photon energy was sampled in each case from a probability distribution function, derived from the x-ray spectrum depicted in figure 1 (a). No bow-tie filter was simulated, in this initial study, as data was acquired without one.

The attenuation coefficients of the materials simulated were calculated using the data of the NIST XCOM database (NIST 2005). The interaction processes simulated were the photo-electric effect (PE), Compton scattering (CS) and Rayleigh scattering (RS). The CS angular distribution was taken to be that of the Klein-Nishina formula modified by the incoherent scattering function tabulated by Hubbell *et al* (1975). More sophisticated treatments of the binding effects for CS exist, but are neglected here. In particular, Doppler-broadening, due to the momentum distributions of electrons bound within atoms, was ignored. The RS angular distribution was taken to be that of the Thomson formula modified by the form-factor. The form-factors of materials were calculated from the Hubbell *et al* (1975) tabulations under the independent-atom approximation. Photon histories were terminated by any one of two events: if the photon was photo-electrically absorbed or if it left the voxelized phantom. Characteristic emissions in the phantom, due to atomic relaxations, were assumed negligible and disregarded. Signal was scored in the detector using the FFD scheme outlined in the previous subsection. At each interaction point, following the determination of its location and prior to assignment of scatter type, contributions due to RS and CS were calculated to nodal points on the detector.

The first two ‘random’ numbers sampled in the MC code, at the start of each history, corresponded to the cone and fan angles of the photon direction. The third number generated the photon energy and the fourth the radiological depth to first interaction. These first four numbers were calculated using quasi- rather than pseudo-random numbers. The objective of doing so was to get a more even distribution of numbers in this four-dimensional space, for a low number of photon histories (1000-10000), than would be achieved with pseudo-random numbers. Niederreiter’s number sequence in four dimensions and base-2 was used to generate these quasi-random numbers, applying the algorithm of Bratley *et al* (1994) in a Fortran 90

implementation by Burkhardt². We note a similar use of quasi-random numbers by Kawrakow and Fippel (2000) in a different medical physics application. The remaining ‘random’ numbers, following initial scattering, were calculated using calls to a pseudo-random number generator presented by Salvat *et al* (2003), based on the implementation of James (1990).

Validation of the CRFD scatter model was carried out using a general-purpose MC code containing more sophisticated physical theory: BEAMnrc/EGSnrc (Rogers *et al* 2004, Kawrakow and Rogers 2003). The test object simulated was a $40 \times 40 \times 20 \text{ cm}^3$ block of water centred at the isocentre. A 60 keV point-source was then simulated with a SAD of 100 cm and a $20 \times 20 \text{ cm}^2$ detection-area was simulated at an SDD of 153.6 cm. The cone and fan angles of radiation were each selected to be 7.41° (to exactly encompass the whole detection-area). This scenario is illustrated in figure 2 (c). The ratio of energy fluence with and without the water block in place, at the detector, is the quantity ‘ S_{norm} ’ of equation 21, with the detector-response set to $R(E)=E$. Profiles of this quantity were calculated along the x-axis of the detector using both CRFD and BEAMnrc. In the latter case, planar fluences were calculated based on the output phase-space file due to 4×10^8 photon histories and averaging over $2 \times 4 \text{ cm}^2$ rectangles, in x and y directions, respectively, using the BEAMDP analysis software. In the BEAMnrc simulation, the XCOM cross-sections were selected with AP and PCUT set to 1 keV. Bound CS, RS and atomic relaxations were turned on. Electron transport was turned off. No variance reduction was used. The CRFD calculations were performed to 100 node points along the x-axis (100, 200, 1000 or 10000 photon histories) using either pseudo-random numbers only (pCRFD) or pseudo-random numbers in combination with quasi-random numbers as discussed above (qCRFD).

Figure 2. Diagrams of (a) the concept of FFD, (b) the geometry of FFD and (c) the geometry modelled with CRFD and BEAMnrc.

2.4 Data acquisition and analysis

The CBCT unit used for the experiments was a Synergy XVI system. The nominal SAD and SDD were 100 cm and 153.6 cm, respectively. The flat-panel detector was a $41 \times 41 \text{ cm}^2$ PerkinElmer a-Si panel with 1024×1024 pixels. Approximately 650 projections were acquired over a 2π source rotation. The medium and small fields-of-view were used to acquire images of a RANDO and Catphan phantoms, respectively. Open-field images were acquired in every case, to normalise the projection-images. The requirement of not saturating the panel in the unattenuated part of the images required a relatively low exposure selection: 0.4 mAs per view was chosen. In all acquisitions, the x-ray tube was operated at 120 kV. The z-axis was taken to be the axial direction and the detector was assumed to be orthogonal to this during its rotation with SAD (source-to-axis distance) and SDD (source-to-detector distance) unvarying. There is in fact flex in the gantry during rotation (Jaffray *et al* 2002). This was approximately

² Available from: http://people.scs.fsu.edu/~%20burkardt/f_src/niederreiter/niederreiter.html.

corrected for using flex-data obtained during routine quality assurance. The optional bow-tie filter was not used for the acquisitions. For the Catphan phantom, data was acquired both with the protective touch-guard of the flat-panel removed and in-place to assess its effect on reconstructed CT-number (scatter from this is not modelled in the algorithm).

In this work, CT-number will be defined as,

$$CT = \rho_{weq} \times 1000, \quad (22)$$

where an accurately reconstructed voxel of water ($\rho_{weq} = 1.0 \text{ g cm}^{-3}$) would correspond to a CT-number of 1000. For the purposes of quantitative assessment of image uniformity, the non-uniformity, NU , was defined:

$$NU = \frac{CT_C - CT_P}{CT_C} \times 100\%, \quad (23)$$

where CT_C is the mean CT-number in a region-of-interest (ROI) placed at the centre of an image of a uniform phantom region and CT_P is the mean CT-number at the periphery. Image noise, IN , was defined:

$$IN = \frac{\sigma_C}{CT_C} \times 100\%, \quad (24)$$

where σ_C is the standard deviation (SD) in an ROI at the centre of an image.

3. Results

3.1 Validation against BEAMnrc

Simulated ratios of energy fluence (S_{norm}) at the detection plane, along the x-axis, are shown in figures 3 (a) and (b). The data points with uncertainties shown are the predictions of BEAMnrc. This data required 40 minutes to simulate using a single CPU. The dotted, dashed, thin solid and thick solid curves, in both figures, are the predictions of CRFD with 100, 200, 1000 and 10000 photon histories, respectively. Each data set, even for 10000 histories, took a fraction of a second to calculate. Reduced χ^2 values are quoted in the figures for the agreement of the CRFD variants with BEAMnrc, calculated using the statistical uncertainties on the BEAMnrc results. These χ^2 would be expected to take a value of 1.0 if the predictions of the codes were in asymptotic agreement and the CRFD predictions were noiseless. The agreement becomes increasingly good as more photon histories are sampled. A set of 10000 histories, when quasi-random numbers were used, for example, provided a χ^2 of 2.0, compared to a value in excess of 139.0 with only 100 histories. The limit on the asymptotic mean residual error between BEAMnrc and CRFD, set by the qCRFD predictions with 10000 histories, is small: <1.0%. Note, however, that with even 1000 or less photons the predictions are not wildly inaccurate. Convergence, with increasing numbers of histories, is slower when only pseudo-random numbers are used: see figure 3 (b). The pCRFD approach fares better, however, for very low numbers of histories: ~100 photons. We observe that, for the case simulated, approximately 25% of the energy fluence was due a final Rayleigh scatter towards the detector, with the remainder being due to Compton scatter. This is demonstrated in figure 4 (a). Furthermore, multiple-scattering pre-dominates in the detected signal, as demonstrated in figure 4 (b). In this example, the first five orders of scatter only account for little over 90% of the total scatter. It is clear that both Rayleigh scattering and multiple-scattering must be included in any realistic model of keV CBCT. This has been observed elsewhere (Kyriakou *et al* 2006, 2008).

Figure 3. X-axis profiles of S_{norm} at the detector. BEAMnrc predictions shown in comparison to those of CRFD with 100, 200, 1000 and 10000 photon histories: (a) qCRFD and (b) pCRFD. Reduced χ^2 is quoted for the agreement of the CRFD predictions with BEAMnrc.

Figure 4. X-axis profiles of S_{norm} at the detector. BEAMnrc predictions shown in comparison to those of qCRFD (10000 photon histories): (a) Rayleigh scatter (RS) and Compton scatter (CS) contributions shown and (b) 1st order, 1st-3rd order and 1st-5th order scatter contributions shown.

3.2 RANDO phantom

The 512x512 central slice ($z = 0.0$ cm) reconstruction of the RANDO phantom, without scatter-correction, is shown in figure 5 (a). Shading artefacts are apparent in what should be uniform ‘soft-tissue’. In the ROI delineated by the square (50x50 pixels), the mean reconstructed CT-number was 805 and the standard deviation (SD) was 50.

The CRFD algorithm was applied to the same data for four iterations. A total of 8x8 scatter nodes were calculated across the panel extent for 36 views, with 1000 photon histories being simulated in each view. The coarse reconstruction images were reconstructed on 128^3 cubes and were performed using one in five acquired views and one in four pixels, the remaining data being disregarded during this stage. Each coarse reconstruction took ~ 15 seconds and each set of scatter views ~ 12 seconds. In total, up to the final reconstruction, the four iterations of CRFD took 110 seconds. Figure 5 (b) shows the final CRFD-corrected reconstruction. The shading artefacts in the ‘soft-tissue’ region have been reduced. The mean CT-number in the delineated ROI in this image is 1055 with a SD of 134. The CT-number for ‘soft-tissue’ has therefore been restored to close to water-equivalence with a penalty of increased image noise.

The scatter-to-primary ratio (SPR) is a commonly used measure to quantify the degree of scatter in an image. We quote some figures for interest. In the first projection acquired during the scan, for example, the closest pixel to the ray-projection through the isocentre recorded 1.43% of the unattenuated signal. The corresponding SPR, estimated from the first iteration of CRFD, was 30.1. This is a greatly inflated figure, due to an initial over-

estimation of the scatter (hence the ansatz of subtracting half the scatter-estimate in the first iteration: see Section 2.1). Subsequently the algorithm rapidly converged to an SPR of 1.7.

Figure 5. Central-slice reconstructions of the RANDO phantom with: (a) no scatter-correction and (b) scatter-correction (4 iterations). The windowing is identical in the two images.

3.3 Catphan phantom

Uniformity, noise and CT-number accuracy were quantitatively assessed for the Catphan phantom. Figure 6 (a) shows a reconstructed slice ($z = 3.0$ cm) with no scatter-correction applied (i.e. 0 iterations). This will be referred to as reconstruction *R1*. The cupping artefact is apparent in both the image and the x-axis line-profile. Note that the dark arc appearing near the top of the phantom, reminiscent of a ring artefact, is in fact a physical feature of the phantom. Figure 6 (b) shows the corresponding reconstruction after scatter-correction (*R2*). The cupping has disappeared. Simulations were also carried out for an increased number of histories-per-view, to test the sensitivity to noise (*R3*). Finally simulations were carried out for less coarse initial reconstructions, increased histories-per-view combined with more simulated views and nodal points. This was to test the sensitivity of scatter-correction to higher spatial-frequencies in the scatter (*R4*). The particulars of reconstruction/corrections denoted *R2*, *R3* and *R4* are summarized in table 2, along with their typical CPU runtimes. In all cases four iterations of CRFD were carried out prior to final reconstruction. Visually, differences between the reconstructions *R2*, *R3* and *R4* were so minor that images of the latter two reconstructions are not shown. Example ROIs (50x50 pixels) are shown in the figure 6 (a) and (b), in the periphery and the centre of the object, respectively. These ROIs, on each image (*R1-R4*), were used to generate the data presented in table 3. The value of *NU* in the central slice was reduced dramatically, from $\sim 15\%$ for *R1*, to $<1\%$ for *R2*. This was at the cost of an increasing *IN* from $\sim 3.7\%$ to $\sim 5.1\%$. Similar increases in *IN*, after scatter correction, were seen with different exposure settings. For example, when the exposure per projection was lowered from 0.4 to 0.2 mAs, the uncorrected and scatter-corrected *IN* values were 5.4% and 7.7%, respectively. Increasing the number of photons simulated had a negligible effect on *IN* (*R3*). This suggests that stochastic noise from the MC is not the dominant contribution to the increase. Increasing the angular and spatial sampling (*R4*) did not have a large effect on *IN* and *NU*, although, surprisingly, in this example case, *NU* was slightly increased. Again, we quote estimates of SPR. In the first projection acquired during the scan, the closest pixel to the ray-projection through the isocentre measured 2.96% of the unattenuated signal. The corresponding SPR, estimated from the first iteration of CRFD, was 1.4. This converged, after subsequent iterations, to an SPR of 0.9.

The cylindrical material samples inside the Catphan phantom for the purpose of testing CT-number linearity are: Polyoxymethylene (PMP), Polytetrafluorethylene (PTFE), Polymethylpentene (PMP), Low Density Polyethylene (LDPE), Polystyrene (PS), Polymethyl Methacrylate (PMMA) and air. Figure 7 (a) and (b) show reconstructed slices of the Catphan, for uncorrected (*R1*) and scatter-corrected images (*R2*), respectively. To reduce *IN*, 40 adjacent slices were averaged over a region 1.7 cm in length, centred at $z = -4.8$. Again,

visually, the cupping has disappeared after scatter-correction. The expected CT-numbers, calculated using equation 1, along with the observed CT-numbers, with and without scatter-correction, are shown in figure 8 (a). Without scatter-correction all the CT-numbers are underestimated (white bars). With scatter-correction, in all cases, the accuracy of the CT-numbers is improved (grey bars). The agreement is good for PMP, LDPE, PS and PMMA. Where the expected CT-number is very different from the background material, for PTFE and air, and to a lesser degree, POM, the agreement is less good. Agreement is improved slightly by using data acquired with the touch-guard of the flat-panel removed, but some discrepancy remains (black bars). To demonstrate that this is not due to a lack of convergence in the iterative process, figure 8 (b) shows the reconstructed CT-number for PTFE, for several iterations: convergence has been reached before the fourth iteration. Finally, we note that the coarseness of reconstruction in the iterative process and the sparseness of scatter-estimation points were not responsible for the residual disagreement, as the differences in CT-numbers between the $R2$ and $R4$ reconstructions were negligible.

Figure 6. Reconstructions and x-axis line profiles (at $z = 3.0$ cm) of the Catphan phantom with: (a) no scatter-correction ($R1$) and (b) scatter-correction ($R2$). Peripheral and central ROIs are shown in (a) and (b), respectively. The windowing in the two images is identical.

Table 2. Summary of reconstruction/simulation parameters for simulations $R2$ - $R4$ and the associated reconstruction/simulation CPU times.

	$R2$	$R3$	$R4$
Coarse reconstruction matrix	128^3	128^3	512^3
Views used in reconstruction	123	123	619
Pixels used in reconstruction	256x256	256x256	1024x1024
Coarse reconstruction time	12 s	12 s	45 m
Photon histories simulated	1000	10000	10000
Number of nodes simulated	8x8	8x8	32x32
Number of views simulated	36	36	144
Simulation time per iteration	12 s	1 m 45 s	1 h 40 m
Total time (4 iterations)	1 m 34 s	8 m 12 s	10 h

Table 3. Data summary for ROIs of simulations $R1$ - $R4$.

	$R1$	$R2$	$R3$	$R4$
CT_C	755	953	948	941

CT_P	866	954	948	951
σ_C	28	49	48	48
IN	3.7	5.1	5.1	5.1
NU	-14.7	-0.1	0.0	-1.1

Figure 7. Reconstructions (at $z = -4.8$ cm) of the Catphan phantom with: (a) no scatter-correction ($R1$) and (b) scatter-correction ($R2$). The windowing in the two images is identical.

Figure 8. Graphs showing expected and observed CT-numbers: (a) POM, PTFE, air, PMP, LDPE, PS and PMMA values before and after 4 scatter-correction iterations and (b) PTFE values before correction and after each of the first 4 iterations.

4. Discussion

The CRFD algorithm is to a considerable extent built on the work and observations of previous researchers. Colijn and Beekman (2004) and Mainegra-Hing and Kawrakow (2008) are examples of the quite different implementations of the idea of forced detection in CBCT. In both papers, scoring locations for photons, at the detector, are determined stochastically. Because of the large number of pixels in a typical detection panel, for practical reasons, both sets of authors found it necessary to de-noise their data. This can be justified due to the low spatial-frequency content in scatter signal. Detection points for all photons, can, however, be fixed before simulation: this has been referred to here as ‘fixed forced detection’ (FFD). This approach becomes prohibitively slow when a large number of points are scored, for example, the centre of every pixel (Colijn and Beekman 2004). A much coarser matrix of points can, however, be representative of the scatter profile in the absence of high spatial-frequencies in the signal. Note that with the FFD approach calculates scatter densities to a point, rather than averaging over a finite area. This is the strategy adopted in this work and shown to be practicable, where interpolation is performed between nodal points to obtain the scatter profile. Advantageously, because every scattering photon contributes to every node, the node to node variation is reduced to the extent that no de-noising step is required.

The key combinations of points that make this work original are:

- The use of a coarse reconstruction in the iteration steps and the original projection data at the start of each iteration;
- The use of quasi-random (Niederreiter) numbers in combination with pseudo-random numbers to model photon diffusion;
- The use of FFD to a reduced number of node points and subsequent linear interpolations.

We also note that the x-ray tube spectra and detector response were simulated and the model was tested against real data acquired on a clinical CBCT unit. As far as these authors are aware, all these aspects have not been combined in a study elsewhere.

The scatter prediction of CRFD was quantitatively validated against a widely-used general-purpose MC code (BEAMnrc). The use of quasi-random number accelerated MC convergence, necessitating fewer simulated photon histories than if only pseudo-random numbers had been used. The CRFD algorithm was shown to be effective in removing artefacts from CBCT images polluted with scatter, taking as little as 2 CPU minutes for complete simulation with several iterations. Monte Carlo-based methods should not therefore be considered, necessarily, as clinically impractical. The improved uniformity of images was demonstrated qualitatively (RANDO phantom) and quantitatively (Catphan phantom). The application of CRFD to the image reconstruction was not without deleterious effect: the image noise increased. This was not due to stochastic noise introduced by the MC diffusion model. It seems likely that it was because the detector signal was reduced, after scatter subtraction, whereas the noise in that signal remained unchanged (Siewerdsen and Jaffray 2001). We note that noise was relatively high in the images appearing in this work, compared to routine clinical and quality assurance scans using the same CBCT unit. Firstly, this was because, clinically, the 1024x1024 projection image is typically rebinned to 512x512; this was not done here. Secondly, a relatively low exposure of 0.4 mAs per view was used, to ensure that the unattenuated parts of the beam did not saturate the flat-panel. Quantitatively, CRFD also improved the CT-number accuracy in reconstructions, although this improvement was less marked where the expected CT-number of a material was very different to the background material in which it was embedded. A possible explanation for this is the presence of a blurring effect not modelled in the simulation. This could, for example, be due to sources of scatter within the panel.

This was a preliminary study to introduce the CRFD algorithm, validate it and demonstrate its efficiency. In the future we hope to examine the algorithm further and use it to explore the importance of various aspects of the imaging chain (x-ray source, scatter model, detector response). For maximum clinical applicability, a bow-tie filter must also be modelled, since this is used clinically for many body regions. The sufficiency of a point-deposition model for the energy of photons incident on the flat-panel must also be examined. These subjects are left for subsequent work.

5. Conclusion

A new technique, designated CRFD, was proposed for the reconstruction and scatter-correction of CBCT images. The algorithm was shown to be fast, taking as little as 2 CPU minutes on a desktop PC, for several iterations of scatter simulation and reconstruction. The CRFD technique was shown to be effective in removing image-artefacts due to scatter and in increasing CT-number accuracy.

Acknowledgements

We acknowledge NIHR funding to the NHS Biomedical Research Centre. This work was partially supported by research grant C46/A2131 from Cancer Research UK. The authors would also like to thank David Roberts for useful discussions and supplying the flat-panel energy response data used in this study. Finally, we would like to thank the staff at Elekta (Crawley) for their answers to several queries.

References

Berger M J, Hubbell J H, Seltzer S M, Chang J, Coursey JS, Sukumar R, and Zucker DS 2005 *XCOM*:

- photon cross section database v1.3* (Gaithersburg, MD: National Institute of Standards and Technology). Online available: <http://physics.nist.gov/xcom>.
- Bratley P, Fox B L and Niederreiter H 1994 Algorithm-738 – programs to generate Niederreiter's low-discrepancy sequences *ACM Transactions on Mathematical Software* **20**(4) 494-95.
- Chan H-P and Doi K 1983 The validity of Monte Carlo simulation in studies of scattered radiation in diagnostic radiology *Phys. Med. Biol.* **28**(2) 109-29.
- Chan H-P and Doi K 1985 Physical characteristics of scattered radiation in diagnostic radiology: Monte Carlo simulation studies *Med. Phys.* **12**(2) 152-65.
- Cho P S, Rudd A D and Johnson RH 1996 Cone-beam CT from width-truncated projections *Comput. Med. Imaging Graph.* **20**(1) 49-57.
- Colijn A P and Beekman F J 2004 Accelerated Simulation of Cone Beam X-Ray CT Scatter Projections *IEEE Trans. Med. Imag.* **23**(5) 584-90.
- Ding X G, Duggan D M and Coffey C W 2007 Characteristics of kilovoltage x-ray beams used for cone-beam computed tomography in radiation therapy *Phys. Med. Biol.* **52** 1595-1615.
- Feldkamp L A, Davis L C and Kress J W 1984 Practical cone-beam algorithm *J. Opt. Soc. Am. A* **1** 612-19.
- International Commission on Radiation Units and Measurements 1988 *ICRP Report 44: tissue substitutes in radiation dosimetry and measurement* (Bethesda, MD: ICRU).
- Hubbell J H, Veigele J W, Briggs E A, Brown R T, Cromer D T, Howerton R J 1975 Atomic Form factors, Incoherent Scattering Functions, and Photon Scattering Cross Sections *JPCRD* **4**(3) 471-538.
- Jaffray D A, Siewerdsen J H, Wong J W and Martinez A A 2002 Flat-panel cone-beam computed tomography for image-guided radiation therapy *Int. J. Radiation Oncology Biol. Phys.* **53**(5) 1337-49.
- James F 1990 A review of pseudorandom number generators *Comput. Phys. Commun.* **60** 329-44.
- Jarry G, Graham S A, Moseley D J, Jaffray D J, Siewerdsen J H and Verhaegen F 2006 characterization of scattered radiation in kV CBCT images using Monte Carlo simulations *Med. Phys.* **33**(11) 4320-28.
- Kachelrieß M, Sourbelle K and Kalender W A 2006 Empirical cupping correction: A first-order raw data pre-correction for cone-beam computed tomography *Med. Phys.* **33**(5) 1269-74.
- Kak A C and Slaney M 1988 *Principles of Computerized Tomographic Imaging*, (IEEE Press).
- Kawrakow I and Fippel M 2000 Investigation of variance reduction techniques for Monte Carlo photon dose calculation using XVMC *Phys. Med. Biol.* **45**(8) 2163-83.
- Kawrakow I and Rogers D W O 2003 *NRCC Report PIRS-701: the EGSnrc Code System* (Ottawa, Canada: NRCC).
- Kyriakou Y, Riedel T and Kalender W A 2006 Combining deterministic and Monte Carlo calculations for fast estimation of scatter intensities in CT *Phys. Med. Biol.* **51**(18) 4567-86.
- Kyriakou Y, Meyer M and Kalender W A 2008 Technical note: comparing coherent and incoherent scatter effects for cone-beam CT *Phys. Med. Biol.* **53**(10) N175-N185.
- Li H, Mohan R and Zhu X R 2008 Scatter kernel estimation with an edge-spread function method for cone-beam computed tomography imaging *Phys. Med. Biol.* **53** 6729-48.
- Mainegra-Hing E and Kawrakow I 2008 Fast Monte Carlo calculation of scatter corrections for CBCT images *Journal of Physics: Conference Series* **102** 012017.
- Malusek A, Sandborg M and Carlsson G A 2008 CTmod-A toolkit for Monte Carlo simulation of projections including scatter in computed tomography *Computer methods and Programs in Biomedicine* **90** 167-78.
- Malusek A, Magnusson Seger M, Sandborg M and Carlsson G A 2005 Effect of scatter on reconstructed image quality in cone beam computed tomography: evaluation of a scatter-reduction optimisation function *Radiation Protection Dosimetry* **114**(1-3) 337-40.
- Marchant T E, Moore C J, Rowbottom C G, MacKay R I and Williams P C 2008 Shading correction algorithm for improvement of cone-beam CT images in radiotherapy *Phys. Med. Biol.* **53** 5719-33.
- Ning R, Tang X and Conover D 2004 X-ray scatter correction algorithm for cone-beam CT imaging *Med. Phys.* **31**(5) 1195-1202.
- Poludniowski GG and Evans PM 2007 Calculation of x-ray spectra emerging from an x-ray tube. Part I. Electron penetration characteristics in x-ray targets *Med. Phys.* **34**(6) 2164-74.
- Poludniowski GG 2007 Calculation of x-ray spectra emerging from an x-ray tube. Part II. X-ray production and filtration in x-ray targets *Med. Phys.* **34**(6) 2175-76.
- Rinkel J, Gerfault L, Esteve F and Dinten J-M 2007 A new method for x-ray scatter correction: first

- assessment on a cone-beam CT experimental setup *Phys. Med. Biol.* **52** 4633-52.
- Roberts DA, Hansen V N, Niven A C, Thompson M G, Seco J and Evans P M 2008 A low Z linac and flat panel imager: comparison with the conventional imaging approach *Phys. Med. Biol.* **53**(22) 6305-19.
- Rogers D W O, Ma C-M, Ding G X, Walters B, Sheikh-Bagheri D, and Zhang GG 2004 *NRCC Report PIRS-0509(A)revK: BEAMnrc Users Manual* (Ottawa, Canada: NRCC)
- Salvat F, Fernandez-Varea J M, Acosta E and Sempau J 2003 *PENELOPE, A Code System for Monte Carlo Simulation of Electron and Photon Transport* (OECD Nuclear Energy Agency, Issy-les-Moulineaux, France)
- Seibert J A 2006 Flat panel detectors: how much better are they? *Pediatr. Radiol.* **36** (Supplement 2) 173-81.
- Siewerdsen J H, Daly M J, Bakhtiar B, Moseley D J, Richard S, Keller H and Jaffray D A 2006 A simple, direct method for x-ray scatter estimation and correction in digital radiography and cone-beam CT *Med. Phys.* **33** (1) 187-97.
- Siewerdsen J H and Jaffray D A 2001 Cone-beam computed tomography with a flat-panel imager: Magnitude and effects of x-ray scatter *Med. Phys.* **28**(2) 220-31.
- Tuy HK 1983 An inversion formula for cone-beam reconstruction *Siam J. Appl. Math.* **43**(3) 546-52.
- Zbijewski W and Beekman F J 2006 Efficient Monte Carlo Based Scatter Artefact Reduction in Cone-Beam Micro-CT *IEEE Trans. Med. Imag.* **25**(7) 817-27.
- Zhu L, Bennett N R and Fahrig R Scatter Correction method for X-ray CT Using Primary Modulation: Theory and Preliminary Results *IEEE Trans. Med. Imag.* **25**(12) 1573-87.