A simple Monte Carlo based optimisation model to determine image contrast in an imaging system

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Abstract. Whilst a radiotherapy imaging system can be modeled accurately using full Monte Carlo simulations a quicker method is required for system optimisation. Here we present a Monte Carlo based optimisation model that takes a radiation spectrum and detector response curve as inputs and predicts contrast for a particular imaging system. The model consists of two parts. The first part looks at the interaction of mono-energetic beams with phantoms of various bone and water compositions. In particular the scatter and primary components emerging from the phantom are analysed. The second part models the response of a detector to various mono-energetic beams. Weighting of the phantom simulations with a linac spectrum results in the prediction of the scatter and primary components incident on an imaging device. Subsequent application of a detector response curve yields the scatter and primary signals in the imager. This technique has been applied to standard 6 and 4MV linac spectra as well as an experimental low atomic number (Z) target configuration to determine various imaging parameters. In particular it has been used to determine the contrast with various detector, phantom and linac spectra combinations. Use of the model shows significant benefits in using a detector with a reduced metal plate thickness for the low Z beam for thin, 5.8cm phantoms that approximate the head and neck region. Whilst contrast can be doubled using the low Z beam and standard MV imager over the current 6MV system, a further improvement in contrast is predicted when using a detector without a metal plate.

1. Introduction
Image contrast in a radiotherapy portal image is essential to allow patient setup errors to be quantified and for patient position to be corrected. Various methods[1-3] have been suggested to improve the current poor contrast megavoltage images traditionally used in radiotherapy. Here we consider modifying the metal plate thickness of an amorphous silicon flat panel imager to see if image contrast increases with the standard 4 and 6MV linac spectra and a Low Z target model outlined by Flampouri et al[3].

Whilst a radiotherapy imaging system can accurately be modeled using full Monte Carlo simulations[3] a quicker method is investigated here based upon Monte Carlo calculated kernels. This method, although not completely representative of a ‘complex’ contrast experiment highlights various factors that influence image contrast. Detector response curves are calculated allowing the model to
predict primary and scatter signal components. Application of the technique to a variety of phantoms allows contrast to be calculated for any linac and detector configuration.

2. Method
In this model (figure 1a) the interaction of mono-energetic pencil beams with a variety of slab phantoms is simulated using BEAMnrc [4]. The phantoms are 16cm wide so as to correspond to the width of a more complex phantom shown in figure 1b. The phantom was 5.8cm thick to approximate the head and neck region where the low Z technique is most useful. Thicker phantoms can be modelled but low Z beams suffer from severe beam hardening making them less advantageous for thicker, more pelvis like phantoms. The phantoms contained inserts with varying bone thicknesses ranging between 0 and 3.2cm in 0.2cm steps. Fifty mono-energetic beams between 0.001 and 10MeV in evenly spaced log10 steps are perpendicularly incident on the phantoms to obtain primary and scatter kernels at the imager layer (40x40cm area located at 35.3cm from the phantom). As a first approximation, this model does not consider the effect of photon angle on the scatter distribution or the detector response. The primary and scatter kernel spectra are analysed to form Energy in ($E_{in}$) vs. Energy out ($E_{out}$) vs. fluence graphs as in figure 2.

![Diagram](image)

Figure 1 – (a) Simple Slab Model (b) Complex Phantom.

![Graphs](image)

Figure 2 – Fluence components from a variety of mono-energetic beams perpendicularly incident on a 5.8cm phantom with no bone layer. (a) All fluence (b) Scatter fluence. Curves normalised to maximum in (a)
Using the results in figure 2 the $E_{in}$ axis can be weighted by an x-ray spectrum to yield the primary and scatter spectra incident on the detector. Subsequent application of a detector response curve results in the scatter and primary signal in the imaging device. This process is conducted for a variety of bone thicknesses to allow the contrast to be calculated using equation (1).

\[
\text{Contrast} = 2 \left( \frac{SP_0 + SS_0}{SP_0 + SS_0} \right) - \left( \frac{SP_x + SS_x}{SP_x + SS_x} \right)
\]

where $SP_0$ is the detector signal from the primary component of a phantom with 0cm bone, $SS_0$ is the scatter signal from a 0cm bone phantom, $SP_x$ is the primary signal from a x cm bone phantom and $SS_x$ is the scatter signal from a x cm bone phantom.

In this study we have considered spectra produced by an Elekta Precise Treatment System Linac producing 6 and 4MV photons and a low Z linac design based on that by Flampouri et al [3]. The x-ray spectra were calculated by simulating the linac geometries using BEAMnrc and analysing the photon spectrum in a phase space file located at a source to surface distance (SSD) of 60cm for a 20x20cm field. Spectra produced by these linacs can be seen in figure 3.

For the detector we considered a standard Elekta iViewGT electronic portal imager and modified its metal plate thickness. Detector response curves were produced by modelling perpendicularly incident mono-energetic x-ray beams on a DOSXYZnrc[5] model of the iViewGT detector as described by Parent et al[6]. These simulations produced point spread functions for a variety of energies that were then integrated to determine the total dose in the gadolinium layer. The total signal in the gadolinium oxysulphide (Gdx) scintillator layer was taken to be the response of the detector. This model was adjusted to decrease the copper plate thickness to observe its change in spectral sensitivity. The three detector models considered here are referred to as D0.1 (0.1cm Cu 0.029cm Gdx-Standard iViewGT), D0.05 (0.05cm Cu 0.029 cm Gdx) and D0.0 (0.00cm Cu 0.029cm Gdx). Figure 3 shows a marked increase in response when decreasing the copper plate thickness which will lead to an increase in quantum efficiency for the low Z beams. The step in the MV Panel (0cm Cu 0.029cm Gdx) curve is the gadolinium k-edge.

**Figure 3 – Detector response curves (left axis - solid lines) and linac spectra (right axis – dotted lines).**
For means of comparison a full Monte Carlo model of a complex contrast phantom (figure 1b) as
described by Flampouri was simulated using DOSXYZnrc_phsp[7]. This phantom consists of a series
of bone inserts of varying thickness surrounded by water 5.8cm deep. The input phase space files to
the simulation were those used to calculate the linac spectra for the simple model. The resultant phase
space file from DOSXYZnrc_phsp was then convolved with pre-calculated Monte Carlo point spread
functions for a variety of detectors to obtain an image. Subsequent analysis yielded the contrast for
the complex 5.8cm deep phantom for a series of detectors.

3. Results

Figure 4 shows the predicted contrast results for a variety of x-ray spectra and detector combinations
for a thin 5.8cm phantom.

![Contrast for thin 5.8cm phantom.](image)

It can be seen that no significant contrast improvement is seen for the 6 and 4MV cases, but a large
improvement is seen with the low Z aluminium beam. This effect can be attributed to the results in
figure 3 showing that as the copper thickness is decreased the sensitivity of the panel to lower energies
is increased. Figure 3 also shows that the 6MV spectrum lies in a region where the response of the
detector varies little between configurations and hence there is no improvement in contrast. Our
simple model also allows us to observe the effect the different panels have on the detected spectrum
from our phantoms. Table 1 highlights how by modifying the detector we significantly decrease
the average detected energy of our beam. As the linear attenuation coefficients of bone and water are more
dissimilar at lower energies, the contrast subsequently improves as shown in figure 4.

Table 1 – Average detected energy across the whole EPID surface (40x40cm) for a variety of different
panels for the low Z beam case and 5.8cm phantom (no bone layer). Average beam energy incident on
EPID is 0.840MeV.

| Detector  | Average Energy of Spectrum (MeV) | Detector Signal Scatter to Primary Ratio |
|-----------|----------------------------------|----------------------------------------|
| 0.01cm Cu 0.029cm Gadox | 0.715 | 0.119 |
| 0.05cm Cu 0.029cm Gadox | 0.646 | 0.128 |
| 0.00cm Cu 0.029cm Gadox | 0.535 | 0.148 |

Whilst decreasing the average energy of the beam improves contrast, the scatter component of the
beam is increased when reducing the copper thickness. Table 1 shows that as the copper layer is
removed the scatter to primary signal ratio increases which will have an effect on the image contrast.
and spatial resolution of the system. It must be noted that in this simple model we have not considered the effect of decreasing the metal plate thickness on spatial resolution, which in turn can impact on contrast.

Also shown on figure 4 is an experimentally verified Monte Carlo model of a complex phantom. Whilst these results differ from that of the simple model the same trends with different detectors exist. These differences occur due to the differing geometries of the two methods. In the case of the complex phantom all bone inserts are present during imaging and thus the signal detected under any bone section is a complex combination of primary radiation, and scatter from many of the bone inserts. In the case of the simple model we have assumed that only scattered radiation from one bone thickness is present. Other factors, most notably spatial resolution will affect the results from the complex phantom due to the finite size of the bone inserts.

4. Conclusion
It has been shown that reducing the metal plate thickness significantly improves contrast for the low Z beam and thin phantoms. Whilst the scatter signal and hence noise increases when decreasing the copper plate thickness the quantum efficiency (QE) will increase due to the increased sensitivity of the panel to low energy photons . A future system would have to balance the metal plate thickness against QE, contrast and noise. The simple model introduced here only considers contrast, so although the simple model was adequate for this study, deviations from the model exist with more complex phantoms. This discrepancy can be attributed to the finite size of the bone sections in the complex phantom and to the mixed scatter signals detected from such a phantom.

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