Beam energy considerations for gold nano-particle enhanced radiation treatment

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Abstract
A novel approach using nano-technology enhanced radiation modalities is investigated. The proposed methodology uses antibodies labeled with organically inert metals with a high atomic number. Irradiation using photons with energies in the kilo-electron volt (keV) range shows an increase in dose due to a combination of an increase in photo-electric interactions and a pronounced generation of Auger and/or Coster–Krönig (A–CK) electrons. The dependence of the dose deposition on various factors is investigated using Monte Carlo simulation models. The factors investigated include agent concentration, spectral dependence looking at mono-energetic sources as well as classical bremsstrahlung sources. The optimization of the energy spectrum is performed in terms of physical dose enhancement as well as the dose deposited by Auger and/or Coster–Krönig electrons and their biological effectiveness. A quasi-linear dependence on concentration and an exponential decrease within the target medium is observed. The maximal dose enhancement is dependent on the position of the target in the beam. Apart from irradiation with low-photon energies (10–20 keV) there is no added benefit from the increase in generation of Auger electrons. Interestingly, a regular 110 kVp bremsstrahlung spectrum shows a comparable enhancement in comparison with the optimized mono-energetic sources. In conclusion we find that the use of enhanced nano-particles shows promise to be implemented quite easily in regular clinics on a physical level due to the advantageous properties in classical beams.

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Recently, methodologies using mono-clonal antibodies that target specific tumor cells have been used to bring active compounds in the vicinity of these cells. One approach uses
radioactive compounds of $\alpha$- or $\beta$-emitters (Nayak et al. 2007, Pless et al. 2004, Miederer et al. 2008). Alternatively, chemotherapeutic compounds have been attached to this delivery mechanism. The use of such approaches is interesting but limited due to the fact that the therapeutic compound is already active at time of delivery and during secretion by the body. More in particular with radioactive compounds an important whole body dose (red marrow dose) as well as renal toxicity is the limiting factor for the efficacy of the treatment (Otte et al. 2002, Schumacher et al. 2002, Behr et al. 2002, Cybulla et al. 2001).

It is the goal of this paper to investigate a delivery method that could potentially have most of the benefits associated with the previously listed therapeutic modalities and has almost none of the disadvantages. Which means the following.

(i) Differentiation between malignant and healthy cells.
(ii) Enhanced effectiveness.
(iii) Image guidance possibilities.
(iv) Activation methodology (i.e. only active where it needs to be active).
(v) Large therapeutic window.

Dose enhancement due to the presence of gold nano-particles has been proposed already both by means of an injectable contrast agent and by the use of mono-clonal antibodies or other targeted delivery methods. However, the enhancement relied on an increased interaction due to the increased probability of the photo-electric interaction being dependent on the atomic number at the proposed energies and the specific contribution of Auger electrons was not investigated (Hainfeld et al. 2004, Verhaegen et al. 2005, Cho 2005, Cho et al. 2009, Carter et al. 2007). All sources used in these studies were spectral sources and/or brachytherapy sources. Moreover, proposals to use more sophisticated photon sources have been put forward, in the hope to maximize the efficiency of the conversion of the beam energy to deposited energy as well as generate a high amount of Auger electrons (Silver et al. 2008).

In the dose deposition model proposed here a significant part of the energy is deposited by Auger electrons. There is reason to believe that Auger electrons deposit their energy more efficiently than those emanating from Compton or photo-electric effect processes. The exact mechanism behind this apparent dose enhancement effect is still unclear. A possible cause is the fact that Auger electrons have a very low energy and deposit all of the energy within a range comparable to a few cell diameters. Furthermore, there is a possible change in the stopping power energy dependence at very low energies (<10 keV), where the Bethe formalism breaks down. Alternatively, it could be that on average more than a single Auger electron is being produced, increasing the probability of clustered double strand breaks.

A number of authors have investigated the biological effects indirectly and support the notion that Auger electrons indeed have high LET characteristics (Chen 2008, Urashima et al. 2006, Balagurumoorthy et al. 2008).

To our knowledge, a systematic study of the impact of different spectral sources on the enhancement and the possible biological enhancement has not been published.

2. Methods and materials

To perform the planning simulation we used MCNPX (Monte Carlo N-Particle eXtended) (Waters et al. 2007) version 2.7a running on a 738-node cluster at the University of Leuven. The department of experimental radiotherapy at Leuven is part of the beta-test group for MCNPX. The following physical parameters were used during these simulations.
Photon energy cutoff: 1 keV

Electron energy cutoff: 1 keV

EM interaction library: ENDF/B-VI Release 8 Photoatomic Data 02/07/03.

For the spectral dose deposition from Auger electrons, version 2.7b was used. As the version is only available for use as a single processor binary, it was not possible to run this on the cluster. Therefore, if we did not need information on the Auger electrons separately we chose to use the earlier version.

2.1. Geometry

The simulation geometry, shown in figure 1, consisted of a tank filled with water containing three 1 mm thick cylindrical slabs holding tissue as defined by ORNL report TM8381. One slab was positioned at the surface of the tank representing a skin surface. A second slab was positioned at a 5 cm distance from the tank surface downstream from the source. In this slab varying concentrations of gold were added in a homogeneous distribution. Additionally, geometrically identical slabs containing tissue were positioned downstream adjacent to the structure containing the gold particles. The source in this geometry was a plane source (not divergent). The divergence can be introduced depending on the position of the source in a clinical situation by applying an inverse square rule.

2.2. Source

The radiation source is modeled after mono-ennergetic radiation sources obtained by Bragg–Gray diffraction of regular x-ray sources. Although these sources are called mono-energetic,
they do exhibit some spectral spread, which was modeled as a normal distribution with \( \sigma = 1.5 \text{ keV} \).

2.3. Simulations

2.3.1. Energies. To investigate the energy dependence of the therapeutic window, we performed the same simulation with quasi-mono-energetic beams of energies ranging from 10 to 200 keV, with special consideration to the \( K_{\alpha} \)-energy of gold (i.e. 80.67 keV). Additionally, a broad spectrum beam was investigated. The spectrum was taken to be identical to that coming from an Acuity simulator’s x-ray tube (Varian Inc.) running at 110 kVp.

2.3.2. Concentrations. The medium used in the activated environment was considered to be tissue as defined in the (Oak Ridge National Labs (ORNL) Report TM-8381). Gold (\( Z = 79 \)) in natural isotope abundance was added with all other components diminished to yield a normalized weight. The structures were embedded in water. The concentrations of gold in one of the structures varied between 0 and 10% in steps of 1%. A concentration of 10% is highly unlikely; however, as reported by Verhaegen et al it is the concentration of off-the-shelf contrast material and should serve as an upper limit of the enhancements achievable with this technique. Furthermore, it is to be expected that once a distribution methodology for the gold particles is implemented we are bound to see very heterogeneous concentrations of gold in the irradiated medium, reaching high concentrations locally.

All concentration-related simulations were performed with the broad spectrum, for reasons made clear in section 3.

2.3.3. Auger electrons. As mentioned above, the contribution of the Auger electrons could be estimated by tagging the electrons released due to Auger cascades in the cell of interest. The number of generated Auger electrons could then be linked to the energy of the source.

This to find the optimal energy for Auger electron generation. Below energies of 1 keV MCNPX do not track the electrons and the energies are considered to be deposited locally. The simulations were performed for all energies as listed above.

2.4. Analysis

2.4.1. Energy dependence. The energy dependence is reviewed for the maximal concentration of 10%; this is because any differences between the energies would be magnified, as well as reduce the statistical errors in our Monte Carlo calculations. We define three types of dose enhancement.

**Absolute enhancement** \( (E_a) \): the ratio of the dose deposited in the gold-containing structure to the dose deposition in a run with exactly the same geometry with no gold present.

**OAR enhancement** \( (E_{OAR}) \): the ratio of the dose deposited in the gold-containing structure to the dose deposited in another structure in the geometry not containing gold, representing an organ at risk (OAR).

**Skin ratio** \( (E_s) \): the ratio of the dose deposited in the gold-containing structure to a layer 1 mm under the skin, which is a special case of \( E_{OAR} \).

We determined the energy deposited per unit mass by counting the energy deposited by electrons using the MCNPX tally F6:E. This underestimates the effectively deposited dose
slightly as it does not take into account the energy expended to generate the photon initiated ionization. However, this ionization is likely not contributing to a biological effect as it is mainly through interaction with the gold atom, which does not form part of the cell structure. The difference between F6:E and F6:P (energy transferred by photons, which includes electrons) is about 1%. However, the electron energy deposition is better defined spatially.

2.4.2. Concentration. The results from the concentration study were analyzed as a function of absolute enhancement. The variation $E_a$ was fitted using a linear relationship and a second-order polynomial. The fit was performed using a minimization of a $\chi^2$-function taking into account the simulation errors.

2.5. Radiobiological effect

To estimate the relative effect of the change in the spectrum and the increase in the contribution by Auger-electrons, we used a fast Monte Carlo model of biological damage as proposed by Semenenko and Stewart (2006), which has been shown to obtain the same results as track Monte Carlo codes as proposed by Nikjoo et al (1997). The approach used here determines the amount of different damage to a DNA molecule from direct ionization as well as through the generation of radicals. It then produces a yield ($y$) in percentage of the different types of damage ranging from single strand breaks to complex clustered double strand breaks for interacting electrons of a given energy ($E$). The methodology used to incorporate this information in our calculations is as follows.

(i) For every energy available in the energy deposition histogram we used the code provided by Semenenko and Stewart to generate the yields of the different types of DNA damage making sure that the lowest energy is included. The yield is given as percentage per cell and per Gy (% cell$^{-1}$ Gy$^{-1}$).

(ii) The data were fitted as a function of energy of the electron using an equation of the form

$$y(E) = a + bE^c$$

with $a$, $b$ and $c$ being the variable parameters. Figure 2 shows $y(E)$ for single strand breaks and double strand breaks together with the fitted parameters, which are provided in table 1.
Enhancement ratios as a function of energy. For all points error bars are drawn but are too small for visualization, as all simulations were performed to yield errors smaller than 1%.

(a) Skin ratio, (b) OAR enhancement.

Table 1. Parameter fit for the number of single, resp. double, strand breaks Gy$^{-1}$ cell$^{-1}$ using a function of the form $a + bx^c$. Energies are expressed in MeV, and confidence levels from the fit procedure are given.

| Parameter | SSB       | DSB       |
|-----------|-----------|-----------|
| $a$       | 1136.3 ± 0.040 | 49.656 ± 0.019 |
| $b$       | −0.1735 ± 0.0037 | 0.0569 ± 0.0020 |
| $c$       | −0.9450 ± 0.0034 | −0.9067 ± 0.0054 |

(iii) The total relative yield of damage of type D ($Y_D$) was then given by

$$ Y_D = \sum_{E=E_{\text{min}}}^{E_{\text{max}}} y_D(E) F(E) $$

with $F(E)$ being the normalized histogram of the deposited energy as a function of the energy of the depositing electrons (in MCNPX tally F6:E divided by the total energy).

(iv) This is repeated for all damage types and volumes with and without nano-particles.

(v) The ratios of the different yields for these volumes provide the relative yield.

For reasons of simplicity we chose to concentrate on the yields of single and double strand breaks. The latter are defined as strand breaks on different DNA-helices not more than ten base pairs apart (Nikjoo et al 1997).

3. Results

3.1. Enhancement

Figures 3(a) and (b) show the maximal enhancements (i.e. 10% solution) of the dose in the volume containing GNP (target) compared to the skin and organ at risk structures. The skin ratio shows two local maxima at 60 keV and at 90 keV. The OAR, which is positioned downstream of the target structure, decreases monotonically reflecting a shielding effect due to the increased photon absorption in the target.
Figure 4. The number of electrons generated from the different possible channels in the medium containing the gold concentration as a function of energy for quasi-mono-energetic beams. The point at 110 keV represents the result from a bremsstrahlung spectrum. The Auger electrons show a maximum around 40 keV while a sudden increase is also noted at the Kα-edge. Note that there still is a substantial contribution of Auger electrons in the bremsstrahlung (110 kVp) beam.

Figure 5. Energy spectra of the energy depositing electrons in the activated volume, when using an energy below (left) the Kα-edge of gold, contrasted with the spectrum resulting from irradiation with a photon beam of 90 keV which is above this edge (right).

Figure 4 shows the absolute number of Auger or Coster–Kröning electrons (AE) generated in the target. This quantity depends on the atomic structure (local maximum at 90 keV) and the depth of the target in the medium (local maximum at 40 keV). The highest number generated occurs at 90 keV. However, the energy of the AE in the latter case is much higher than those generated with lower energy. This can be seen in figure 5 where the energy deposition of the various electrons is presented. Also note that in both cases the contribution of AE to the energy deposition process remains an order of magnitude lower. The largest part of the dose is deposited by photo-electric (PE) and knock-on electrons.

We plotted the absolute number of AE generated at the position of the GNP filled volume. Very low energy x-rays are not able to reach this position. For this reason we see no expression of the photo-activation threshold for the M-shell AE.
Figure 6. Graph depicting the variation of the absolute enhancement factor ($E_a$) as a function of the gold concentration. A second-order polynomial is fit to the values using a $\chi^2$-minimization.

Figure 5 provides a visualization of the energy deposition by the different types of electrons at two different energies. It is the energy of each electron that deposits energy in the medium. For this reason we see a continuous contribution from the Auger electrons as they gradually lose their energy. In the 60 keV beam only the L-shell electrons are generated, while for the 90 keV beam the K-shell Auger electrons augment the dose. Auger electrons can be generated in different ways. The general mechanism is an atomic relaxation after ionization either from a photon interaction or from a direct electron interaction. Both are plotted separately in the plots. A very small contribution is noted from Auger electrons from oxygen. Both graphs extend slightly above their nominal energy as the spectra used were quasi-mono-chromatic and thus have a finite spectral width.

### 3.2. Concentration

Figure 6 shows the absolute enhancement ratio as a function of the concentration of gold in the cell. $E_a$ was fitted with a second-order polynomial, where the quadratic factor is small with respect to the other factors, showing that the dose enhancement is quasi-linear in the sense that the quadratic term in the polynomial has a small contribution. When the second-order polynomial is denoted as $f(x) = ax^2 + bx + c$, the coefficients are $a = (-0.0195 \pm 0.0005)$, $b = (1.386 \pm 0.004)$ and $c = (1.003 \pm 0.003)$. Fixing the $c$-coefficient to 1 gives comparable results. If a linear fit is performed the slope is given by $(1.26 \pm 0.02)$, a number which can be used to easily predict the impact of concentration changes.

### 3.3. Biological effect

Figure 7 shows the yield per cell and per Gy of single strand breaks (SSB) and double strand breaks (DSB) as a function of energy, for the three volumes under consideration. The target
volume containing GNP reflects the atomic structure in the calculated yields. One can easily identify when Auger electrons are generated as a function of beam energy. The addition generates increases in DSB and lowers the contributions of SSB. However, the added tail of high-energy electrons mainly affects the DNA through SSBs. It is only for very low energies (10–20 keV) that the addition of GNP changes the ratio of SSB to DSB to give the dose deposition a more high LET character. For deeper lying structures there are no data available for 10 keV as the photons do not penetrate deep enough to generate a meaningful contribution.

The spectrum in the target does contain more low-energy electrons compared to the OARs. This advantage is negated by the additional tail of high-energy electrons as illustrated in figure 8 where a cumulative representation of the dose deposition in double strand breaks is shown for a beam energy of 60 keV.

4. Discussion

From the data available here, one can conclude that there exists a necessity to perform full Monte Carlo simulation based on these type of treatments. Not only does the enhancement change with energy, it is also imperative to monitor the concentration of the nano-particles during treatment. Before treatments can be started, methodologies need to be developed to monitor concentrations adequately. The monitoring of the concentration and the error of this specific measurement has a direct impact on the dosimetry as we find a relationship of concentration with dose which is close to unity. This implies that the uncertainty of the dosimetric planning is at least equal to the uncertainty of this measurement.

The choice of energy of the radiation beam for treatment does not seem to be a critical issue, as long as radiation reaches the intended volume. So, it might not be necessary to build expensive high output mono-energetic photon sources. However, to determine the concentration it might be necessary to do so albeit with a lower photon flux (Gambaccini et al 2001, Baldelli et al 2005). A possible alternative for measurement is the use of the same markers with nano-particles attached, but where the gold is replaced by a radioactive isotope (Roy and Lahiri 2006). Using the methodology developed in nuclear
medicine imaging techniques, the concentration and its variation can then be monitored over time. This approach gives up some of the advantages as the radioactive particles will deliver a dose in the manner we are trying to avoid. The dose, however, is only for imaging purposes and is more limited than when a therapeutic procedure is attempted. A further approach could be to measure concentration changes using other nano-particles attached to the same targeting agent and use non-ionizing techniques for visualization (Sun et al 2008). A drawback of this technique is that due to the use of different nano-particles, the uptake and concentration dynamics could be slightly different from the therapeutically enhanced targeting molecules.

The impact of Auger electrons seems to be limited as they consist of only a small fraction of the dose depositing electrons. Only at very low energies do they seem to have an effect in increasing the efficiency of the dose deposition. Using such low energies limits the use to superficial tumors, or warrants the use of intra-operative techniques or brachytherapy techniques, using low energy sources like $^{125}\text{I}$ or electronic brachytherapy sources (Rivard et al 2006).

The data presented here do not stand alone and can be compared to the experimental data provided by other authors who have also looked at the possible optimal energies to use for nano-particle-enhanced treatment using in vitro techniques as well as implementations through animal testing. Rahman and colleagues (Rahman et al 2009) used clinical superficial x-ray machines to irradiate the cells containing GNPs at different concentrations. Nominal energies of 80 and 150 kVp showed dose enhancement. In agreement with our calculations they show a dependence of cell survival on energy, whereby the 80 kVp beam is shown to be more effective. It is not clear what the exact spectra of both aforementioned beams were, as only nominal indications were given. In theory it could be possible to predict the overall dose enhancement of a poly-chromatic beam from the weighted sum of mono-chromatic beams. Also important to note is that Rahman and colleagues found increased cytotoxicity depending on the concentration, a factor not taken into account in our analysis.

Brun et al performed a comparable study also using different superficial energies characterized by the effective energy of the x-ray beam (14.8, 24.4, 29.8, 42.4, 49 and
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70.1 keV). The highest enhancement was noted at 49 keV, which could be considered as a 50 keV beam. This again agrees with our calculations.

In the papers discussed above the dose is determined in water and the enhancement factors are determined by estimating the increase in cell kill. In our approach we attempt to resolve the difference between the physical dose and the biological effect. Indeed figure 7 has the physical dose removed and shows that most of the enhancement proposed here is a consequence of the increased photon absorption rather than an effect of short-range Auger electrons.

Finally, an interesting approach is presented by Biston _et al_, where cis-platinum is introduced in rats with a glioma tumor (Biston _et al_ 2004). The tumor is subsequently irradiated with quasi-mono-energetic x-ray beams just below and above the K-edge of platinum ($Z = 78$). This group reports an increase of DSB damage at higher energy. This is in agreement with figure 7 where we see a substantial increase just above the K-edge. However, the group has not tested their hypothesis at other energies. From the same figure we see that there are other energies that provide a better DSB/SSB ratio. From figure 5 we see that in all cases, and also at the ‘photo-activation’ energy, the contribution of Auger electrons to the dose is an order of magnitude lower than the contribution from other sources. Therefore, it might not be necessary to have mono-chromatic sources and sources with a sufficiently high flux could be adequate. Proposals of converting conventional linear accelerators using low-Z targets come to mind as proposed by Robar (2006).

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