Effects of bone- and air-tissue inhomogeneities on the dose distributions of the Leksell Gamma Knife® calculated with PENELOPE

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Abstract
Monte Carlo simulation with PENELOPE (version 2003) is applied to calculate Leksell Gamma Knife® dose distributions for heterogeneous phantoms. The usual spherical water phantom is modified with a spherical bone shell simulating the skull and an air-filled cube simulating the frontal or maxillary sinuses. Different simulations of the 201 source configuration of the Gamma Knife have been carried out with a simplified model of the geometry of the source channel of the Gamma Knife recently tested for both single source and multisource configurations. The dose distributions determined for heterogeneous phantoms including the bone- and/or air-tissue interfaces show non-negligible differences with respect to those calculated for a homogeneous one, mainly when the Gamma Knife isocentre approaches the separation surfaces. Our findings confirm an important underdosage (∼10%) nearby the air-tissue interface, in accordance with previous results obtained with the PENELOPE code with a procedure different from ours. On the other hand, the presence of the spherical shell simulating the skull produces a few per cent underdosage at the isocentre wherever it is situated.

1. Introduction

GammaPlan® (GP) (Elekta 1996) is a computer-based treatment dose planning system designed to calculate the dose distributions of the Leksell Gamma Knife® (GK) for stereotactic radiosurgery of certain brain diseases. Like almost all radiosurgery planning systems, GP is quite simple. Using a standard set of beam data, the dose distributions in patients are calculated by adding those corresponding to each one of the 201 beams of the GK actually present in
each particular treatment (Wu et al 1990, Wu 1992). GP assumes homogeneous target media and tissue heterogeneities are not taken into account (Yu and Sheppard 2003).

However, in stereotactic radiosurgery, Solberg et al (1998) have pointed out a remarkable disagreement between Monte Carlo (MC) results and those predicted by the usual planning systems, in the case where inhomogeneous phantoms are considered.

In the investigation of dose perturbations produced by heterogeneities, MC has showed up as a useful tool, mainly because it accounts, in an adequate way, for the lack of electron equilibrium near interfaces. For the GK, Cheung et al (2001), using the EGS4 MC code, have found discrepancies up to 25% in the case of extreme irradiation conditions, mainly near tissue interfaces and dose edges. This contrasts with the sub-millimetre accuracy with which the GK operates (Elekta 1992).

In this paper we have investigated the effects of bone- and air-tissue interfaces on dosimetric calculations involving the GK. To simulate the GK, a simplified geometry model of the source channels is considered to perform the calculations. This model was proposed in Al-Dweri et al (2004) and it is based on the characteristics shown by the beams after they pass through the treatment helmets. It has been shown that the collimation system of each source channel acts as a ‘mathematical collimator’ in which a point source, situated at the centre of the active core of the GK source, emits photons inside the cone defined by the point source and the output helmet collimators. If a homogeneous target phantom is considered, this simplified model of the GK produces doses in agreement with those found if the full geometry of the source channel is considered, with those calculated by other authors with various MC codes and with the predictions of GP, for both a single source (Al-Dweri et al 2004) and different multisource (Al-Dweri and Lallena 2004) configurations.

In this work we want to use the simplified geometry model of the GK to calculate doses in the case of heterogeneous target phantoms, including bone- and air-tissue interfaces. Simulations have been performed by using the 2003 version of PENEOLOPE (Salvat et al 2003). We compare our findings for the 201 source configuration with those obtained by Cheung et al (2001) with EGS4 and by Moskvin et al (2004) with PENEOLOPE (version 2001). Different situations of the GK isocentre (both far and near the interfaces) are considered.

2. Material and methods

2.1. Leksell Gamma Knife® model

To study the effect of the heterogeneities, we have used different configurations of the phantom depicted in figure 1. It is chosen to be a sphere with 80 mm radius made of water except for the two shadow regions. Region 1 is a cube with sides of 30 mm and with its centre at 50 mm from the centre of the phantom as shown in the figure. It is considered to be made of material \( m_1 \) which can be air (‘a’), to simulate the maxillary or frontal sinuses, or water. Region 2 consists of a 5 mm width spherical shell with its external surface situated at 5 mm from the phantom surface. We have considered this shell made of material \( m_2 \) which can be either bone (‘b’), to simulate the skull, or water (‘w’). The different phantom configurations have been labelled as \( \mathcal{P}_{m_1m_2} \). With this notation, \( \mathcal{P}_{ww} \) labels the homogeneous phantom. The origin of the coordinate system is chosen to be at the centre of the phantom, as indicated in the figure. In the figures below, the different regions relevant to the calculations will be shown in gray scales and labelled with the corresponding number in italic.

As mentioned above, each one of the 201 sources of the GK are simulated according to the simplified geometry model which is described in detail in Al-Dweri et al (2004). It consists of a point source emitting the initial photons in the cone defined by the source itself.
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and the helmet outer collimators. The coordinates of the 201 point sources can be found in Al-Dweri and Lallena (2004). They are distributed in the $z < 0$ region and correspond to the situation in which the isocentre of the GK coincides with the centre of the phantom. For the simulations described below, in which the isocentre is situated at different positions, these coordinates must be shifted. The position of the isocentre appears explicitly in the figures as $[x_I, y_I, z_I]$, with the values of the coordinates in mm.

Due to the fact that the distribution of the sources is not completely uniform, no cylindrical symmetry is shown by the system. Thus, the doses we have calculated depend on the three Cartesian coordinates, $D^{(m_1m_2)}(x, y, z)$. The superscript refers to the materials of the particular phantom $P_{m_1m_2}$ considered in the simulation. Throughout the paper, the values of the coordinates are given in mm.

Cheung et al (2001) used a phantom similar to our $P_{wb}$. It included the 5 mm wide bone shell at 5 mm from the phantom surface as in our case, but with a full diameter of 180 mm. This phantom cannot be positioned inside the treatment helmets in such a way that the isocentre of the GK approaches the skull interface. That is why in our simulations we have chosen the phantom described above, which is slightly smaller. In any case, we have compared our results with those of Cheung et al by performing simulations with their phantom, which we label $P_{wb}$. To do that we have calculated the quantity

$$D_{\text{norm}}(x, y, z) = \frac{D^{(wb)}(x, y, z)}{[D^{(wb)}(x, y, z)]_{\text{max}}},$$

which corresponds to the dose obtained for $P_{wb}$ divided by its maximum.

A first evaluation of the effects of the different interfaces has been obtained by calculating the relative differences

$$\Delta_{ww}^{(m_1m_2)}(x_I, y_I, z_I) = \frac{D^{(m_1m_2)}(x_I, y_I, z_I) - D^{(ww)}(x_I, y_I, z_I)}{D^{(ww)}(x_I, y_I, z_I)}$$

between the doses obtained at the isocentre for the heterogeneous, $P_{m_1m_2}$, and homogeneous, $P_{ww}$, phantoms.
Table 1. PENELOPE tracking parameters of the materials assumed in our simulations. $E_{\text{abs}}(\gamma)$ and $E_{\text{abs}}(e^-, e^+)$ stand for the absorption energies corresponding to photons and electrons and positrons, respectively.

| Materials         | Air     | Bone and water |
|-------------------|---------|----------------|
| $E_{\text{abs}}(\gamma)$ (keV) | 1.0     | 1.0            |
| $E_{\text{abs}}(e^-, e^+)$ (keV) | 0.1     | 50.0           |
| $C_1$             | 0.05    | 0.1            |
| $C_2$             | 0.05    | 0.05           |
| $W_{cc}$ (keV)    | 5.0     | 5.0            |
| $W_{cr}$ (keV)    | 1.0     | 1.0            |
| $s_{\text{max}}$ (cm) | $10^{35}$ | $10^{35}$     |

In addition, we have calculated the quantity

$$d^{(m_1m_2)}_{ww}(x, y, z) = \frac{D^{(m_1m_2)}(x, y, z)}{[D^{ww}(x, y, z)]_{\text{max}}}$$

in order to analyse the differences observed in the dose profiles calculated for the different phantoms.

2.2. Monte Carlo calculations

PENELOPE (version 2003) (Salvat et al 2003) was the MC code used to perform the calculations. PENELOPE permits the simulation of the coupled transport of electrons and photons, for an energy range from a few hundred eV up to 1 GeV, for arbitrary materials. PENELOPE provides an accurate description of the particle transport near interfaces.

Photons are simulated in PENELOPE in a detailed way. Electrons and positrons are simulated by means of a mixed scheme which includes two types of events: hard events, which are simulated in detail and are characterized by polar angular deflections or energy losses larger than certain cut-off values, and soft events, which are described in terms of a condensed simulation based on a multiple scattering theory (Salvat et al 2003). The tracking is controlled by means of the four parameters $C_1$, $C_2$, $W_{cc}$ and $W_{cr}$, as well as the absorption energies. All these parameters must be fixed for the materials present in the geometry considered in the simulation. Table 1 shows the values we have assumed in our simulations. In addition we have fixed the parameter $s_{\text{max}} = 10^{35}$ in all the simulations performed.

The initial source was selected by sampling uniformly between the 201 sources. Initial photons were emitted with the average energy 1.25 MeV and uniformly in the corresponding emission cone.

The number of histories followed in each simulation was $3 \times 10^8$. This permitted us to maintain the statistical uncertainties under reasonable levels. The uncertainties given
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Table 2. Composition of the materials assumed in the MC simulations performed in this work. The values correspond to the weight fraction of each element in the material. Also the densities are quoted. The three materials have been generated with the code material included in the PENELOPE package and correspond to the numbers 104, 119 and 277, respectively, in the material database of the MC code.

|       | Air    | Bone   | Water  |
|-------|--------|--------|--------|
| H     | 0.52790| 0.111894 |
| C     | 0.000124| 0.19247 |
| N     | 0.755267| 0.01603 |
| O     | 0.231781| 0.213111| 0.888106 |
| Mg    | 0.00068 |
| P     | 0.01879 |
| S     | 0.00052 |
| Ar    | 0.012827|
| Ca    | 0.03050 |
| Density (g cm\(^{-3}\)) | 0.0012048| 1.85 | 1.0 |

throughout the paper correspond to \(1\sigma\). In many of the figures, the error bars do not show up because they are smaller than the symbols used.

3. Results

3.1. Comparison with EGS4 calculations

First of all, we have compared our results with those obtained by Cheung et al. (2001) using the EGS4 code. They considered the 18 mm helmet for two situations of the isocentre: \(I[0, -66, 0]\) and \(I[0, 0, -69]\). Their results (solid curves) are compared with our findings (open squares) in figure 2, where the values of \(D_{\text{norm}}(x, y, z)\), as given by equation (1), are plotted for the three Cartesian axes and the two positions of the isocentre. As we can see, both calculations are in good agreement for the two cases considered.

In the figure, the predictions of the GP, quoted by Cheung et al. (2001), are also included (dashed curves). For the isocentre at \(I[0, -66, 0]\), a discrepancy between MC simulations and GP results is observed in the far negative \(y\) region (see medium left panel). This is due to the fact that the GP does not take into account the interfaces and assumes all tissues to be uniformly represented by water. The same situation is not observed when the isocentre is at \(I[0, 0, -69]\) (see right lower panel), because in that case the dose is roughly zero before reaching the interface (for \(z \sim -80\) mm).

3.2. Comparison with GammaPlan predictions

Figure 3 shows a comparison of different simulations performed for the phantom \(P_{w}\) (open squares) with GP predictions of Hamad and Mherat (2005) (dashed curves). In the upper panels the isocentre is situated at \(I[0, 34, 0]\) and the profiles along the \(y\)-axis are shown for the 18 and 4 mm helmets. The isocentre is situated at \(I[0, 66, 0]\) in the medium panels where the profiles along the \(x\)- and \(z\)-axes are shown for the 14 mm helmet. Finally, the profiles along the \(y\)-axis are shown in the lower panels, for the 14 and 8 mm helmets and the isocentre situated at \(I[0, 70, 0]\).

As we can see, the simulation for the water phantom produces results in very good agreement with the GP predictions. Below (see section 3.4), the effects of the interfaces in these cases will be analysed and it will be clear that GP cannot describe these effects.
3.3. Effects of the tissue inhomogeneities on dose at the isocentre

Now we analyse the results obtained for different positions of the isocentre of the GK, paying special attention to the situations in which the isocentre is close to the interfaces.
First, we investigated the effects of tissue inhomogeneities on the doses calculated at the isocentre. We varied its position by fixing the coordinate $y_I$ at different values ranging from $-70$ mm to $70$ mm and maintaining $x_I = z_I = 0$. The results obtained for the
Figure 4. Relative differences \( \Delta_{ww}^{m_1 m_2}(0, y_I, 0) \) in percentage, between the doses calculated at the isocentre for heterogeneous and homogeneous phantoms (see equation (2)). The upper panel corresponds to the phantom \( P_{wb} \); the medium panel represents the results in the case of the heterogeneous phantom \( P_{aw} \), and, finally, in the lower panels the phantom \( P_{ab} \) has been considered. We have plotted the results for the 18 mm (left panels) and 8 mm (right panels) helmets. Similar results are obtained for the 14 and 4 mm helmets.

As we can see, the presence of the bone spherical shell (upper panels) produces a reduction of the dose at the isocentre with respect to that obtained for the homogeneous phantom. This reduction is observed at practically any position of the isocentre, being \( \sim 3\% \) for the two helmets. A higher perturbation in the dose is observed when the isocentre is situated exactly at the bone-water interface, \( y_I = \pm 70 \text{ mm} \). In this case, the reduction in the dose rises to 5% for the 18 mm helmet and it is even larger for the 8 mm one. It seems evident from our results that the effect of the full skull, which we simulate here by means of the bone spherical shell, is not negligible at all.
To have an idea about the origin of this effect, we have evaluated the reduction in the dose due to the bone inhomogeneity in a very simple case. We have considered a photon pencil beam coming from the source and reaching the phantom and we have calculated the dose at the isocentre neglecting scattering photons. In the case where the phantom $P_{wb}$ is considered, the dose at the isocentre is proportional to (see e.g. Berger (1968))

$$D^{(wb)}(x_I, y_I, z_I) \propto \left( \frac{\mu_{en}}{\rho} \right)_w \frac{E_0}{4\pi r^2} \exp\left[ -\mu_a s_4 - \mu_w (s_3 + s_2 + s_0) \right],$$  \hspace{1cm} (4)

where $(\mu_{en}/\rho)_w$ is the mass energy absorption coefficient of water at the initial energy of the photons $E_0$, and $\mu_a, \mu_w$ and $\mu_b$ are the attenuation coefficients of air, water and bone, respectively, at the same energy. The values $s_i$ correspond to the length of the trajectory segments travelled in the region $i$ of the phantom, and thus

$$r = s_4 + s_3 + s_2 + s_0$$  \hspace{1cm} (5)

is the distance from the source to the isocentre. If we consider the phantom $P_{ww}$,

$$D^{(ww)}(x_I, y_I, z_I) \propto \left( \frac{\mu_{en}}{\rho} \right)_w \frac{E_0}{4\pi r^2} \exp\left[ -\mu_a s_4 - \mu_w (s_3 + s_2 + s_0) \right],$$  \hspace{1cm} (6)

We are interested in the fraction of both doses, which is given by

$$\frac{D^{(wb)}(x_I, y_I, z_I)}{D^{(ww)}(x_I, y_I, z_I)} = \exp\left[ -(\mu_b - \mu_w) s_2 \right].$$  \hspace{1cm} (7)

For the photon energy considered in our simulations, $E_0 = 1.25$ MeV, the attenuation coefficients can be calculated easily (see Hubbell and Seltzer (2004)) and one obtains $\mu_b = 0.11174$ cm$^{-1}$ and $\mu_w = 0.06323$ cm$^{-1}$. On the other hand, the length $s_2$ can vary from source to source, depending on the position of the isocentre. If the phantom is centred with respect to the helmet, that is if the isocentre is at $I[0, 0, 0]$, $s_2 = 5$ mm for all the sources. In this case the dose ratio is 0.976, and a reduction of 2.4% is found. This is the minimum reduction found for all positions of the isocentre. By varying them in the interval ($-70$ mm, 70 mm) in the three directions, we sample the full volume of the phantom, $s_2$ ranges between 0.5 cm and 2.7 cm and the reduction due to the bone inhomogeneity varies between 2.4% and 12.2%. These results indicate that, as we have found in our simulations, a few per cent reduction in the dose at the isocentre is expected due to the bone inhomogeneity, independently of the position of the isocentre.

The air-tissue interface (central panels) produces a slight increase (1–2% at most) in the dose at the isocentre, in comparison with that found for the homogeneous phantom, when it is situated far from the separation surface. When the interface is approached the relative difference $\Delta_d^{(m/m)}$ increases, the dose at the isocentre for the inhomogeneous phantom is $\sim 5\%$ larger than that obtained for the homogeneous phantom and this occurs until a point very close to the separation surface is reached. Once the isocentre is situated at this position, the dose calculated for the heterogeneous phantom reduces strongly with respect to that of the homogeneous one. This reduction is $\sim 15\%$ in the inner side and $\sim 10\%$ in the outer side of the air cube.

A similar situation is observed when both interfaces are present (lower panels) except for a general decrease of the dose at the isocentre whatever $y_I$ is. This shift towards smaller doses is due, as said before, to the presence of the skull, as we can see clearly in the region $y_I < 0$, far from the air interface, which shows a behaviour rather similar to that plotted in the upper panels. One should point out the fact that the overdosage observed in the region around $y_I = 30$ mm for the $P_{aw}$ phantom is largely reduced when, in addition, the bone shell is considered.
Figure 5. Dose ratio at the isocentre for the doses calculated with the \( P_{ww} \) and the homogeneous \( P_{ww} \) phantoms, as a function of the distance from the isocentre to the air-tissue interface and for the 8 mm helmet. Black squares are our results. Open squares and circles are those of Moskvin et al (2004) for two different areas of their phantom.

Figure 6. Values of \( d_{(m1m2)}^{(ww)}(0, y, 0) \) in percentage, as given by equation (3), when the isocentre is situated at \( I[0, 34, 0] \). The profiles in the \( y \)-axis are shown for the four helmets. The squares correspond to the homogeneous phantom \( P_{ww} \). Solid curves have been obtained with the phantom \( P_{wb} \). Dashed curves represent the results in the case of the heterogeneous phantom \( P_{wb} \). Dotted curves refer to the phantom \( P_{ab} \).
3.4. Effects of the tissue inhomogeneities on dose profiles

The larger effects observed appear when the isocentre is situated near an air-tissue heterogeneity. In order to analyse in detail the dose in this situation, we have calculated
the quantities \( d_{m1m2}^{(m3)}(x, y, z) \), as given by equation (3), for two positions of the isocentre: \( I[0, 34, 0] \) and \( I[0, 66, 0] \). In these two positions the isocentre is at 1 mm distance from the inner and outer sides of the air cube (region 1), respectively. Some results are plotted in figures 6 and 7.

Figure 6 shows the profiles along the \( y \)-axis for the four helmets and for the isocentre at \( I[0, 34, 0] \). Therein the squares correspond to the homogeneous phantom \( P_{ww} \), while solid, dashed and dotted curves have been obtained with the phantoms \( P_{wb} \), \( P_{aw} \) and \( P_{ab} \), respectively. If only the bone is considered (solid curves), a reduction in the plateau region including the maximum dose is observed. This is the same reduction previously discussed for the dose at the isocentre.

On the contrary, the presence of the air-tissue interface (dashed curves) produces a strong reduction of the dose on the 'air' side (the right side in this case) of the interface and an enhancement of the dose profile on the 'water' side (the left side in this case) of the separation surface. These effects are better seen for the 18 mm helmet. The main effect of the simultaneous consideration of both interfaces (dotted curves) is to cancel the overdosage on the left side of the interface. The results here obtained are very similar to those plotted in figure 6 of the work of Moskvin et al (2004). These large differences in the dose produced by the air-tissue heterogeneities cannot be neglected.

Figure 7 depicts the results obtained for the 14 mm helmet when the isocentre is situated at \( I[0, 66, 0] \). The different curves correspond to the same phantoms as in the previous figure. Here the profiles along the three Cartesian axes are plotted. Two facts deserve a comment. First, appreciable in the upper and lower panels, is the reduction of the dose produced by the presence of the air-tissue interface (dashed and dotted curves). Also, the comparison between both curves gives us an idea of the additional diminution produced by the bone shell. Second, the
strong overdosage produced if the air-tissue interface is not taken into account is again evident (see medium panel), but also remarkable is the reduction in the dose observed in region 2 when the bone shell is considered (solid curve).

To complete our analysis, we show in figure 8 results similar to those in figure 6 but for the isocentre situated at [0, 70, 0], that is exactly at the bone-tissue interface. Apart from the reductions observed in the dose nearby the air-tissue interface, here the effect of the bone shell in region 2 is, as expected, stronger than in the previous case.

To finish we point out that, as can be observed in figures 6 and 8 and in the medium panel of figure 7, the dose in the air-water interface is, in all cases, ~50% of the maximum dose obtained for the homogeneous phantom. This result is in agreement with the findings of Moskvin et al (2004).

4. Conclusions

In this work we have investigated the dosimetry of the GK in the case of heterogeneous phantoms by considering a simplified source model for the single source channels. Calculations have been done by using the Monte Carlo code PENELOPE (version 2003) for the configuration including 201 unplugged sources and for different positions of the isocentre of the GK.

The use of the simplified model produces results for the dose profiles at the isocentre which are in agreement with previous calculations done with EGS4, whereas they show discrepancies with the predictions of the GP, mainly at the interfaces.

In general we can say that the presence of typical tissue inhomogeneities produces an underdosage with respect to the results obtained when a homogeneous phantom is considered. This happens for almost all the positions of the isocentre of the GK. This underdosage can reach values larger than 10% in the vicinity of air-tissue interfaces. The only exception to this conclusion occurs when the isocentre is situated at a distance of a few millimetres from the air-tissue separation surface, where an overdosage is produced. However, this overdosage is very small if, in addition to the tissue inhomogeneity, the bone inhomogeneity is also considered.

We have analysed the doses deposited in phantoms including bone and air inhomogeneities and we have found non-negligible discrepancies with the doses obtained in the case of the water homogeneous phantom. In this respect it is worthwhile to mention that an air inhomogeneity simulating the maxillary or frontal sinuses gives rise to large modifications of the dose profiles.

We have found a reasonable agreement with previous calculations performed by Moskvin et al with the PENELOPE code in the case of the air-tissue interfaces.

In what refers to the bone-tissue inhomogeneity representing the skull, our results show a ~3% underdosage at the isocentre, with respect to the doses calculated for the homogeneous phantom. This effect can be observed wherever the isocentre is situated.

The discrepancies observed between the results obtained for heterogeneous and homogeneous phantoms suggest that GP predictions must be corrected in order to take care of the air- and bone-tissue inhomogeneities, mainly in those cases in which the interfaces are present near the target area.

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