THE USE OF MONTE-CARLO CODES FOR TREATMENT PLANNING IN EXTERNAL-BEAM RADIOTHERAPY

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Introduction

Monte Carlo simulation of radiation transport is a very powerful technique. There are basically no exact analytical solutions to the Boltzmann Transport equation. Even the "straightforward" situation (in radiotherapy) of an electron beam depth-dose distribution in water proves to be too difficult for analytical methods without making gross approximations such as ignoring energy-loss straggling, large-angle single scattering and bremsstrahlung production. Monte Carlo is essential when radiation is transported from one medium into another. As the particle (be it a neutron, photon, electron, proton) crosses the boundary then a new set of interaction cross-sections is simply read in and the simulation continues as though the new medium were infinite until the next boundary is encountered.

Radiotherapy involves directing a beam of megavoltage x rays or electrons (occasionally protons) at a very complex object, the human body. Monte-Carlo simulation has proved invaluable at many stages of the process of accurately determining the distribution of absorbed dose in the patient. Some of these applications will be reviewed here (Rogers *et al* 1990; Andreo 1991; Mackie 1990).

Applications in Radiotherapy Dosimetry

Concerning the determination of the absorbed dose at a reference point in a water phantom irradiated by a reference beam, Monte-Carlo simulation has been an essential tool e.g. in accurate calculations of the stopping-power ratio $s_{\mathrm{med,det}}$ (Nahum 1978; Andreo 1988, 1990) for megavoltage beams and the backscatter factor, B for kilovoltage x-ray qualities (Knight 1996; Klevenhagen *et al* 1996). Such computations involve numerical integration over the spectrum of photons or electrons at a given depth in an irradiated uniform medium; this requires the power of Monte-Carlo simulation due to the complex changes in the radiation spectrum with depth. All *Codes of Practice* for absolute dose determination in radiotherapy beams now use M-C generated $s_{water,air}$ for both megavoltage photons and electrons (Thwaites *et al* 1996; IAEA 1987, 1997, 2000). Thus everyday radiotherapy practice all over the world has benefited directly from Monte-Carlo.

M-C can also compute directly the dose ratio D_{med}/D_{det} for the exact geometry of the detector at a certain position in a medium. This is just one specific application of the general case of determining the dose in an heterogeneous medium. Such simulations can then be interpreted to yield the perturbation factor for a particular detector (Nahum 1996), assumed to fulfill approximately *Bragg-Gray* conditions (e.g. Ma and Nahum 1991), by comparing D_{med}/D_{det} with $s_{med,det}$. Such simulations are computationally inefficient as only a small fraction of the particles incident on the phantom surface traverse the small dosimeter. *Correlated sampling* (CS) can be exploited to reduce the *variance*; one single set of histories is *split* at the depth where the geometry of the medium and detector first differ (Ma 1992). The computation of the ratio D_{det1}/D_{det2} can then be highly efficient; one example is the effect of a change in central-electrode material in an ion chamber (Ma and Nahum 1993). The behaviour of small LiF TLDs in the form of 1-cm rods and chips in radiotherapy beam qualities has also been investigated (Mobit 1996) by exploiting CS to yield D_{water}/D_{LiF}. Figure 1, taken from EGS4 simulations by Mobit *et al*. (1998), shows very clearly how the dose ratio, here for Calcium Fluoride TLD discs of 0.9 mm thickness, is equal to the (μ_{en}/ρ)-ratio, i.e. the *large* detector result, at low photon energies (i.e. very short electron ranges) but approaches the stopping-power ratio at high energies where the greater electron ranges result in *Bragg-Gray* detector behaviour.

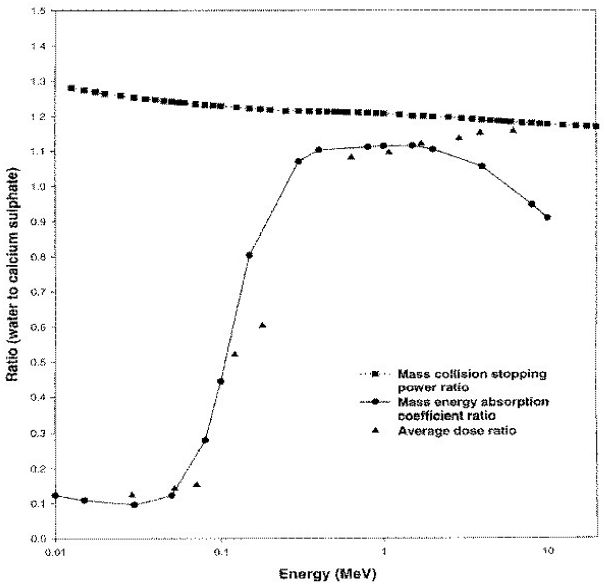

Figure 2. Calcium sulphate TLD discs, 0.9 mm thick, in MV and kV x-ray beams: comparison of the Monte Carlo derived average dose ratio, water to $CaSO_4$, with the mass collision stopping-power and mass energy-absorption coefficient ratios, as a function of the mean photon energy.

Figure 1. The ratio $D_{water}/D_{Calcium\ Fluoride}$ for TLD discs 0.9 mm thick, over a broad range of photon-beam qualities computed directly using the EGS4 code and compared to the two extreme idealised cavity-theory results (reproduced from Mobit *et al* 1998).

Monte-Carlo based radiotherapy treatment planning

The computation of the dose distribution in the inhomogeneous “geometry” of a patient undergoing radiotherapy is an obvious candidate for M-C simulation (Nahum 1988). Up to the present time, most radiotherapy treatment planning systems (TPS), i.e. the software, involve *algorithms*, of varying degrees of sophistication; these "correct" the measured distribution in a water phantom for the irregular patient skin surface, and inhomogeneities, most notably bone, air passages and cavities, and lung tissue. In high-energy (or *megavoltage*) x-ray beams, electron transport close to interfaces of different density is an intractable problem for any *analytical* method (Bielajew 1994; Mohan 1997).

In the case of electron beams the failure of analytical methods (including 3D pencil-beam convolution) to model the effect of density differences on electron scatter is even more serious than for photon beams (Mackie *et al* 1994). *Conformal therapy*, by which is meant the modification of beam directions, apertures and possibly other properties to cause the region, in 3D, of high-dose to *conform* to the shape of the target (Tait and Nahum 1994) causes special difficulties for analytical i.e. non-MC algorithms due to irregular field shapes, Multileaf Collimators, small-field stereotactic techniques and most notably due to the recent emphasis on so-called *Intensity-Modulated* radiotherapy (IMRT), very small beam elements, each of a different *intensity* i.e. fluence rate. Such very small photon-beam fields do not exhibit *charged-particle equilibrium* (CPE) on the central axis (Solberg *et al* 1995) due to the range of secondary electrons exceeding the dimensions of the beam cross section. Figure 2 shows what happens when a narrow photon beam crosses a broad low-density inhomogeneity: due to electron transport *away from* the central axis, which is uncompensated by transport *towards* the axis, the dose in the air region falls and there is then a *re-buildup* in the water beyond. The discontinuity in dose at the first water/air interface is a consequence of the water/air stopping-power ratio (this is **not** unity); M-C always yields the absorbed dose in whatever the medium is, whereas all the current analytical methods effectively yield *equivalent* water dose (Siebers *et al* 2000).

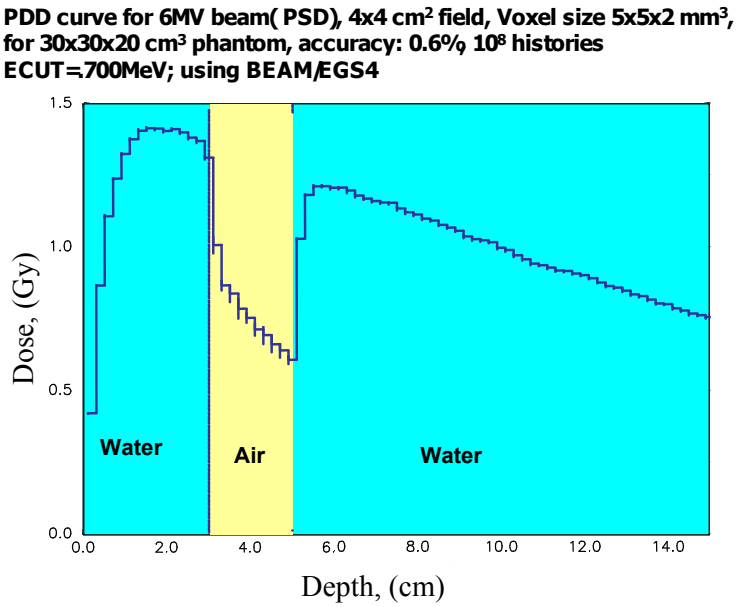

Figure 2. Loss of equilibrium on the central axis due to a low-density region in a narrow photon beam clearly predicted in a Monte-Carlo simulation involving electron transport (Cephas Mubata, priv. comm).

The only factor that has delayed the application of MC methods to the computation of dose distributions in radiotherapy is the amount of computer-processing power necessary; some 15 years ago this was estimated to be of the order of 100s of hours for a photon-beam radiotherapy treatment plan (Nahum 1988). In the last five years or so the picture has changed radically; very fast processors (which cost < $1000 rather than 10-100 times this) are now available – principally PCs. Furthermore, there have been extensive efforts to simulate the full geometrical detail of the passage of radiation through the so-called *treatment head* (target, flattening filter, monitor chamber, collimating jaws etc.) in modern linear accelerators for radiotherapy (e.g. Mohan 1988). There is an impressively comprehensive EGS4 *usercode* called *BEAM* for the purpose of detailed treatment head modelling (Rogers *et al* 1995). *BEAM* can produce a so-called *phase-space file* of co-ordinates (particle type, energy, direction, position) e.g. above the level of the devices which define the beam collimation for the particular patient undergoing treatment. The final patient-specific simulation is then (re-)started from this *phase-space* file. This approach is illustrated schematically in Figure 3.

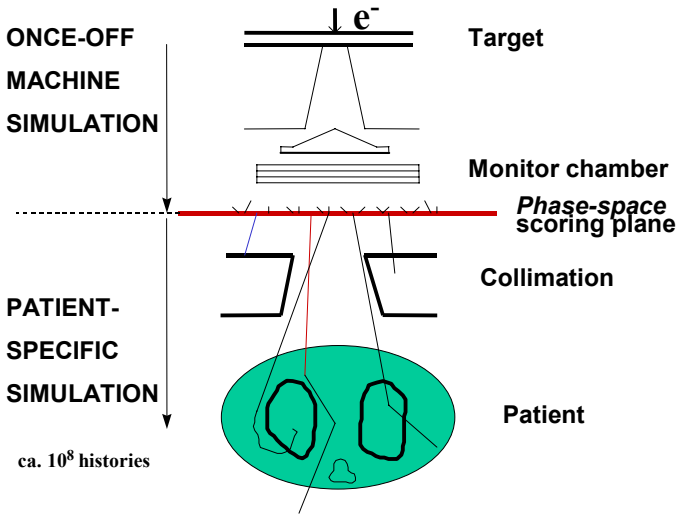

Figure 3. Schematic illustration of the two-step approach to Monte-Carlo simulation in radiotherapy patient dose computation.

For electron beams, runtimes of the order of a few minutes for 1-2% uncertainty (1σ) with a small voxel size (1-2 mm) were demonstrated by Neuenschwander *et al.* (1997) using the *macro-MC* (MMC) approach. In the case of megavoltage *photon* beams (the "bread and butter" of radiotherapy all over the world) of the order of 2-10x10^8 histories are required to fulfill the above specifications on voxel size and statistical uncertainty, essentially independent of the number of beams, (cf. ≈10^7 for electrons). Through a combination of ingenious *variance reduction* techniques, mostly concerned with speeding up (secondary) electron transport simulation, and generally using between 10 and 30 state-of-the-art PCs *in parallel*, i.e. each CPU executes a certain number of histories independently of all the others, several radiotherapy physics research groups (De Marco *et al* 1998; Wang *et al* 1998; Sempau *et al* 2000; Li *et al* 2000; Kawrakow *et al* 1996) have demonstrated that photon-beam treatment plans can be calculated in acceptable runtimes of the order of an hour or less. Articles documenting essentially perfect agreement between measurements and MC simulation in both homogeneous (i.e. water) and heterogeneous phantoms (e.g. Rando standard man) irradiated by radiotherapy beams are now appearing at a rapid rate in the medical physics research journals (e.g. Wang *et al* 1999). This excellent agreement includes not only *relative* dose distributions but also *absolute* dose determination; the latter is often referred to, using radiotherapy jargon, in terms of *Monitor Units* (MUs) and *Output Factors* (OFs) (Verhaegen *et al* 2001).

Two *clinical* examples of Monte-Carlo calculated dose distributions, shown in terms of so-called *isodoses* lines connecting points of equal dose, and superimposed on CT sections through the patient under study, are now given in the next two figures. Figure 4 shows an electron-beam treatment plan; the effect of the inevitable statistical noise is seen as a slight unevenness in the isodoses. The extension of the irradiated volume into the low-density lung is clearly shown. Figure 5 is a dramatic illustration of the difference between calculated dose distributions for a narrow photon beam, incident here on the extremely heterogenous anatomy of the head and neck region; in the figure on the left we see the smooth dose

isodoses produced by the so-called *pencil-beam* algorithm, widely employed in commercial radiotherapy treatment planning systems (TPS), whereas on the right the true behaviour of such a narrow photon beam in heterogeneous terrain is displayed. In fact, it is precisely such grossly distorted dose distributions that comprise the input of so-called *inverse planning* algorithms which are used to deduce the fluence-profile modulation patterns subsequently delivered in the I-M radiotherapy technique. Fortunately this unsatisfactory situation is about to change as a result of the fact several companies are implementing Monte-Carlo simulation into their patient dose calculation systems and also adapting MC to model today's increasingly complex treatment delivery techniques e.g. using MLCs to shape fields to the tumour shape and to modulate the fluence across the beam to reduce the irradiation of so-called *organs at risk*.

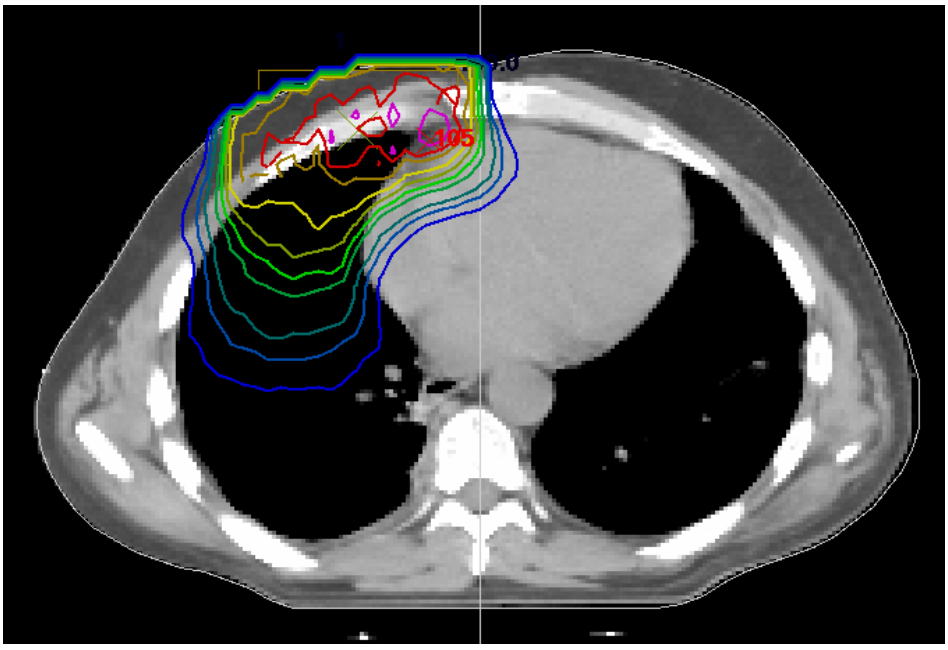

Figure 4. Electron-beam treatment plan in the lung with the dose distribution shown as *isodose* lines: 16MeV beam; 10x10 cm^2 field, calculated with EGS4/BEAM using full *phase-space data* from Varian 2100C linear accelerator; 5x5x5 mm scoring voxels, $2x10^6$ histories, CPU time: 16mins on DEC 500MHz, uncertainty (1σ): 1.5% (Cephas Mubata, private communication).

Will it be possible to *prove* that the increased accuracy of patient dose computation via Monte-Carlo influences the *clinical* outcome of radiotherapy? The rapidly developing subject of so-called *Biological Modelling* i.e. estimation of the probability of tumour (local) control (TCP) and of the probability of complications (unwanted side-effects) to normal tissue (NTCP) using the detailed dose distribution in the tumour and the organs-at-risk (Cattaneo *et al* 2001; Nahum and Sanchez-Nieto 2001; Buffa and Nahum 2000) is likely to give us this possibility.

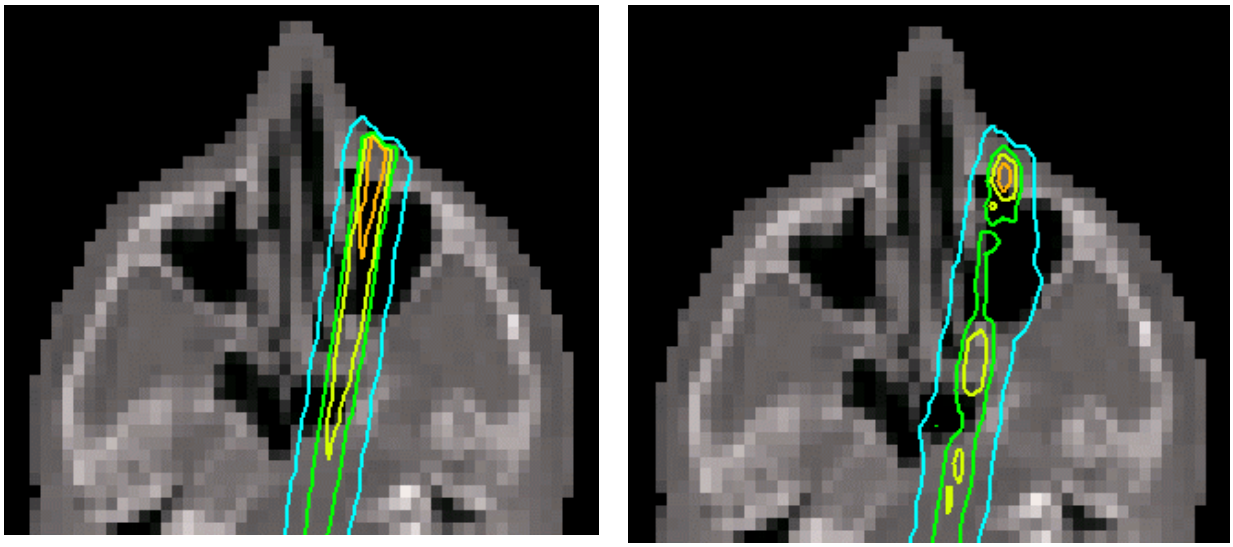

Figure 5. A narrow (megavoltage) photon beam incident on the anatomy in the head (represented by a computer tomograph image). Lhs: doses computed using a pencil-beam algorithm; rhs: Monte-Carlo simulation (Charlie Ma, private communication).

CONCLUDING REMARKS

Monte-Carlo simulation of radiation transport has also played an essential role not only in radiotherapy with external beams but also in several other branches of bio-medical physics e.g. in imaging and therapy in so-called *nuclear medicine* (i.e. unsealed sources of radioisotopes administered to the patient) (Zaidi and Sgouros 2003), and imaging generally with ionising radiation, principally kilovoltage x-ray beams (e.g. Flampouri *et al* 2002).

Ultimately it is thanks to the pioneering work of Martin Berger (Berger 1963) in developing the condensed-history approach to charged-particle Monte Carlo that we can now look forward to truly accurate determinations of the complex 3D dose distributions in patients undergoing external-beam radiotherapy.

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