

MONTE CARLO CLINICAL DOSIMETRY

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Received September 5th, 2001; received in a revised form February 27th, 2002; accepted March 11th, 2002

ABSTRACT

The choice of the most appropriate strategy for radiotherapy treatment is mainly based on the use of a planning system. With the introduction of new techniques (conformal and/or small fields, asymmetrical and non coplanar beams, true 3D calculation, IMRT) the trustworthiness of the algorithms used is questioned. An alternative verification procedure has become increasingly more necessary to warranty treatment delivery.

The reliability of the Monte Carlo method is generally acknowledged. However, its clinical use has not been practical due to the high CPU time required. During the last few years our objective has decreased CPU time by means of a new process distribution technique. This reduction has made it feasible, not only to apply physical dosimetry under special conditions, but also to use it in numerous clinical cases employing photon and electron conformal fields, in radiosurgery, and IMRT.

The procedure carried out is presented. Furthermore, conventional Treatment Planning System calculations are compared with the Monte Carlo simulations.

Key words: Monte Carlo simulation, clinical dosimetry, Treatment Planning System, process distribution.

INTRODUCTION

The usual way to implement a data set for calculations in a conventional treatment planning system (TPS) is to use dosimetric data measured with a water phantom in a linac room. However, in many cases the fields employed may be irregular or very small. In addition, detectors can have their limitations. Generally, the algorithms utilized in a conventional TPS provide results that are adequate in standard situations, but in some special cases new modifications of the algorithms and new empirical data must be considered [1-6].

The Monte Carlo (MC) method allows us to simulate the real particle transport using exact linac geometry [7]. In our case, we have used the MC from the beginning

of the process (the simulation of the linac) to the end (energy deposition at any anatomical voxel) [8].

To overcome the CPU time drawback we have considered the following statement: "as any particle history is independent, the work can be distributed between different CPUs". The first trials were carried out on two different Silicon Graphics computers: an O2 and a dual processor Octane workstation. Later, we implemented the code in a new 24 processor HP machine for which we have priority access, which however was a very expensive solution. Therefore we have developed a PC distributed process that permits us to obtain, at a low price, a crucial reduction in the CPU time. The multiprocessor technique is specially suitable

in Radiosurgery or Intensity Modulation, because the whole treatment can be divided into several beams which can be sent to different CPUs. With this tool we can verify, by means of the MC method, any physical or clinical dosimetry technique [9-11].

MATERIAL AND METHODS

Thanks to the cooperation with the Engineering School of the University of Seville we had access, during non-working hours, to a computer classroom with standard PCs connected to our server (Pentium III 500 MHz type). No software (MC code, Operative System, data, etc.) had to be installed. The computer consisted of a server and 36 CPUS (Pentium II, with 68 Mb of RAM).

Conventional (commercial) Treatment Planning Systems (TPS) used for comparison were: Levinger-Fisher (Radiosurgery), Helax (IMRT) and Focus (Conformal Photon beams).

We used linacs: one from Siemens (Primus) and two from Elekta (SL18 and SL18 with MLC).

We have implemented the OMEGA BEAM MC code distributed by Dave Roger from the NRC of Canada [12-14]. The Linac head was simulated using different BEAM geometrical modules (e.g. FLAT-FILT, CHAMBER, MIRROR, JAWS, MLC, etc), so that it was not necessary to generate any other additional geometry. Below the last module we added an air layer (using SLABS module) lying inferiorly on a linac reference point of a known distance to the target.

A Phase Space Data (PSD) was placed just below the MIRROR module. The PSD is a plane, which stores the position, angle of incidence and instant energy of all the particles in the beam. The file containing the PSD allows us to generate all the beam information from the target to the mirror.

The PSD file is unique for each energy as the geometry above the mirror is the same for all cases. This PSD was used as an input to simulate any field configuration including MLC, any additional collimator for radiosurgery or blocks for conventional conformal therapy. The result is a second PSD. With this PSD, we can study a physical situation in a phantom, obtaining dosimetric curves or isodose distributions. As a reduction variance technique, we used the uniform bremsstrahlung splitting (UBS) in order to maintain the same statistical weight, with 25 splitted photons resulting from each bremsstrahlung interaction. Neither the photon interaction forcing nor the range rejection were activated. The fact that all particles in the PSD have the same statistical weight is essential to relate monitor units to the number of particles.

The last part of the simulation was carried out using a DOSXYZ code, which produced a dose map. Each beam was simulated from the corresponding phase space file, starting below the air layer up to the phantom or patient, which was represented by a density distribution. This density distribution file was built from the CT information, taking under consideration 4 different materials (bone, tissue, lung, and air) and 4096 different density levels. Slice resolution was 512 x 512 pixels of 0.0853 cm, slices spaced every 0.3 cm. Patient densities were distributed in a total number of 256 x 256 x 80 voxels of 0.1706 x 0.1706 x 0.3 cm³ each. We used resolutions ranging from 0.1 by 0.1 by 0.1 cubic millimetre in the case of endovascular brachytherapy up to 4 by 4 by 4 cubic mm in the case of a simple PDD curve. The CT has been calibrated to correlate the Hounsfield number with the medium and physical density. Some treatment characteristics for CT and Phase Space Data are presented in table 1.

Tab. 1. Treatment characteristics for radiosurgery and IMRT simulated with FMC.

	PSD	CT
Radiosurgery	~ 70 beams	Slices: 50 - 80
		Voxels: 256 x 256x (1 mm RC – 1.6 mm Abdomen)
IMRT	~ 80 segments	4096 density levels

The DVH information was accessed by a BIOPLAN (a user-friendly windows-based software developed by one of the authors for biological evaluation of treatments) to perform the calculations of Normal Tissue Complication Probability (NTCP) and Tumour Control Probability (TCP). NTCP and TCP models implemented in BIOPLAN are described below.

NTCP values were estimated using the Lyman Kutcher-Burman Model [15] with Emami parameters [16]. As Emami data were extracted from 2-Gy/fraction clinical data, the tolerance doses (TD50) for radio-surgery cases were corrected to account for fractionation using an alfa/beta of 3 Gy (typical of normal tissues).

To compute TCP, the Poisson model was used [17-18] with clinically acceptable parameters ($\alpha=0.5 \text{ Gy}^{-1}$, $\alpha/\beta=4 \text{ Gy}$ and clonogenic cell density $\phi_{cl}=10^6/\text{cm}^3$).

RESULTS

In the last few years, we have been involved in the application of the Monte Carlo method to radiotherapy in a large variety of clinical cases. Some of them are presented in this paper. The first case represents treatment of an acoustic neuroma with two isocenters of 6 and 2 arcs (figure 1), respectively. When compared with a conventional TPS, there were not any significant differences in the 50% isodose (where the dose was prescribed) but some discrepancies could be found elsewhere (figure 2).

We also want to emphasise the differences in the penumbra of the beam between the Monte Carlo and a conventional TPS (figure 3).

The dose-volume histograms (DVHs) do not show significant differences between the MC and the conventional TPS in the target, but some discrepancies can be observed from a radiobiological point of view. Figure 4 shows the DVHs for the target calculated with the MC and the TPS, as well as the corresponding Tumour Control Probability (TCP) values. In this case, DVHs cross each other at approximately 12 Gy level, the MC predicting a larger cold region than the TPS as well as a larger region irradiated by higher doses. However, as TCP curves are not

symmetrical, even very small cold areas may make the TCP decrease drastically without being compensated for by large hot regions. This explains why, in this case, the TCP calculated from the MC calculations is lower (24.9%) than that associated with to the TPS (30.9%). Figure 5 shows the DVHs for the organ at risk, calculated again by the MC and the TPS, as well as the corresponding NTCP calculations [19-20].

In the case of a conformal photon beam for the prostate treated with 8 fields: 2 + 2 lateral opposite fields and 2 + 2 anterior-posterior fields (figure 6), it can be observed that in the target area the reproducibility is acceptable. However, an overdose in the femoral heads can be noticed in the MC simulation (figures 7 and 8).

Finally, a case of Intensity Modulation is presented. It was calculated using an Inverse-Planning by Helax. The optimised treatment is composed of 7 fields in a full arc of 360 degrees and 43 segments.

From the PSD obtained at the short distance from the jaws, every MLC configuration corresponding to each single segment was simulated by means of our PC distribution tool. The number of particles used in the resulting PSDs was made proportional to the corresponding MU number. Figure 9 shows the dose distribution differences between the TPS and the MC in a patient. A discrepancy between isodose curves can be observed, probably due to the pubic symphysis interface.

The actual times spent to simulate very complex radiosurgical treatments with 36 PC took less than one hour. Some very complex treatments with IMRT techniques were simulated and led to a total time of approximately 5 hours.

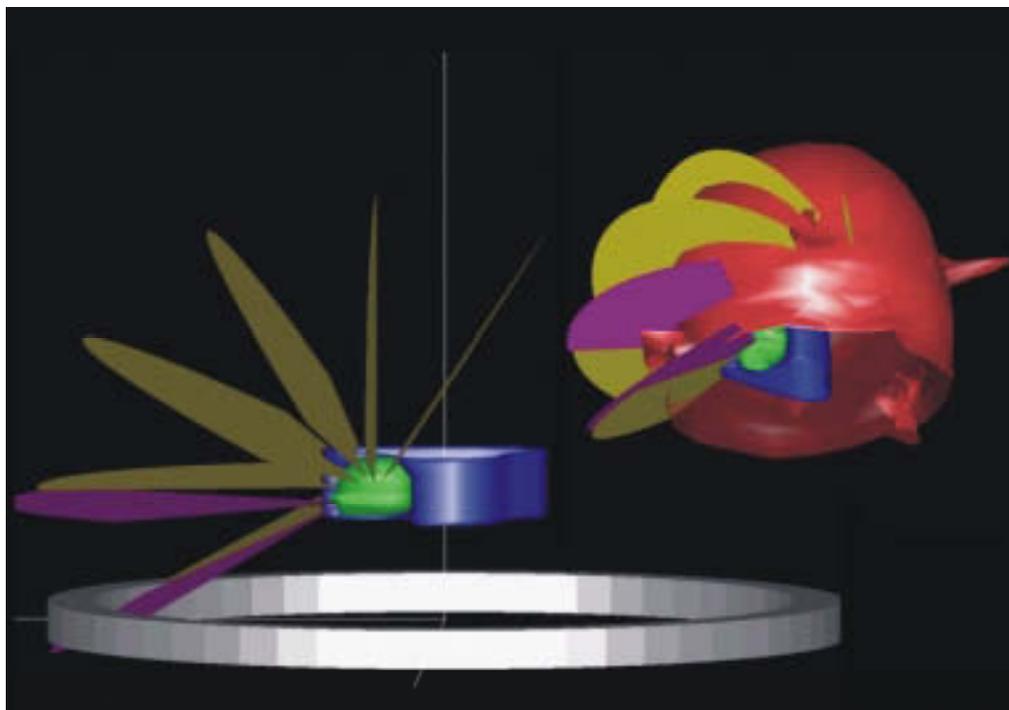


Fig. 1. Schematic of the acoustic schwannoma treatment consisting of eight arcs and two isocentre.

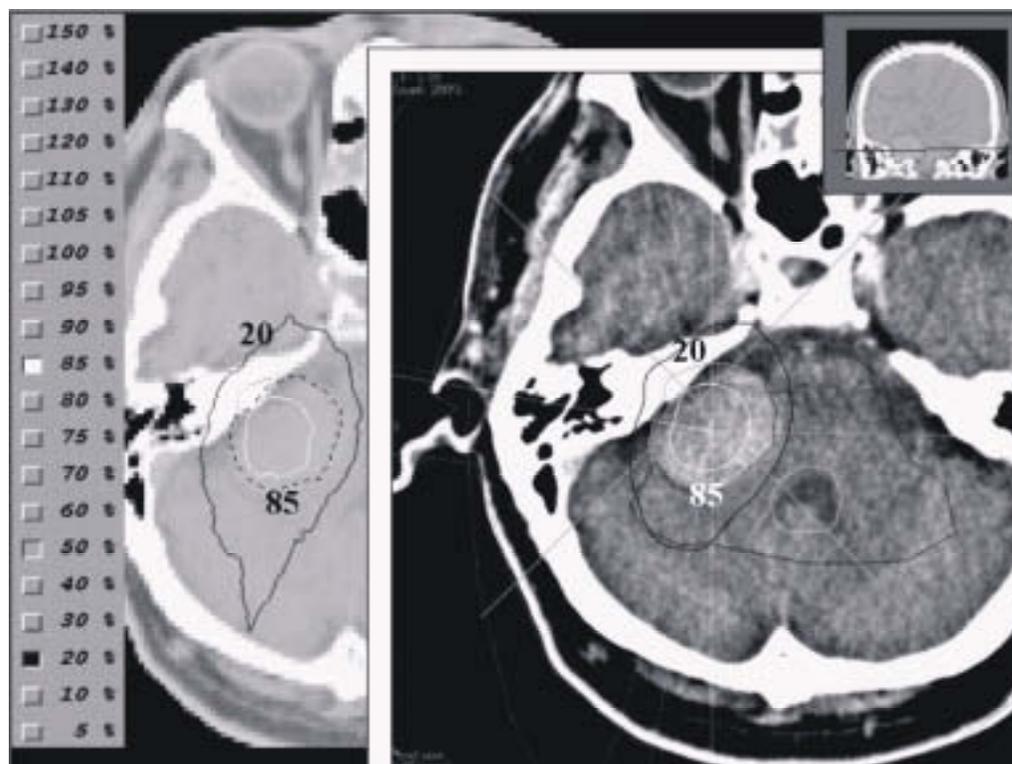


Fig. 2. Comparison of isodose distribution between Monte Carlo (left) and a conventional TPS (right) for an acoustic schwannoma treatment at the target level.

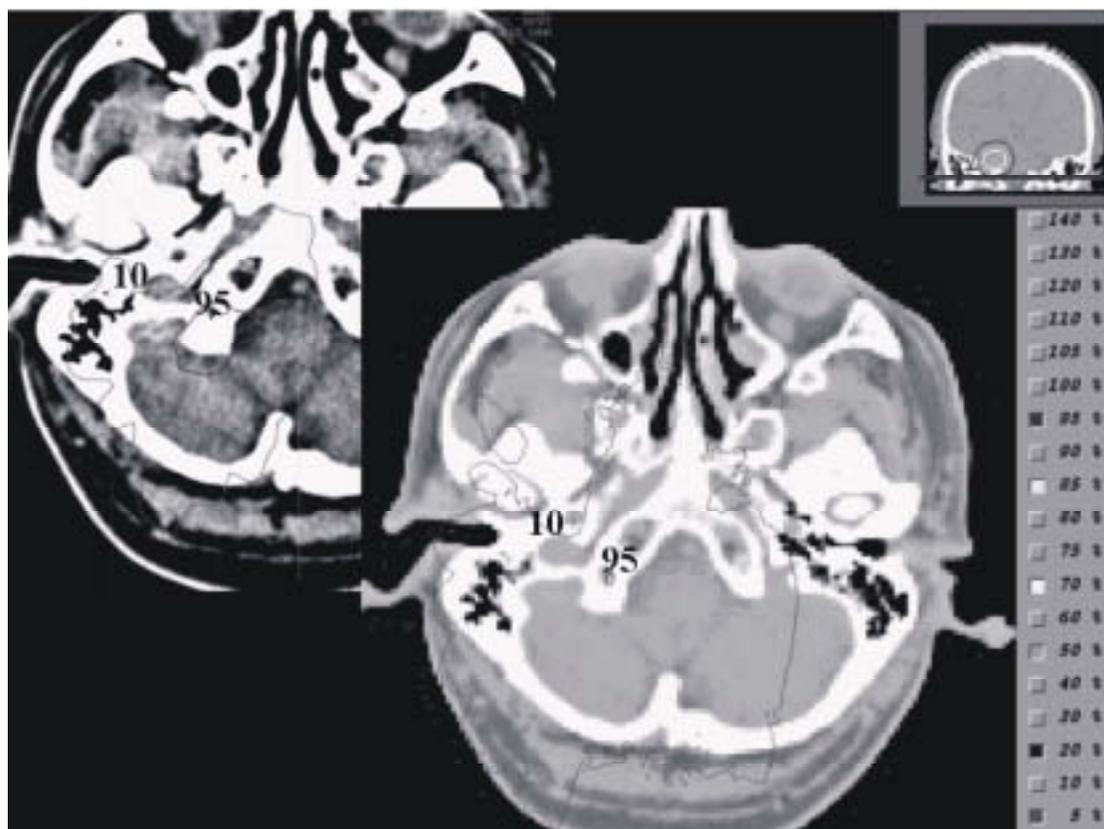


Fig. 3. Comparison of isodose distribution between Monte Carlo (right) and a conventional TPS (left) for the acoustic schwannoma treatment at the field edge.

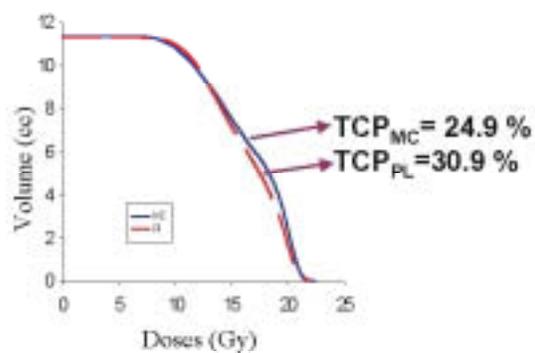


Fig 4. DVH of the target (acoustic schwannoma) showing the differences in TCP when calculated with Monte Carlo (MC) and a conventional planning system (PL).

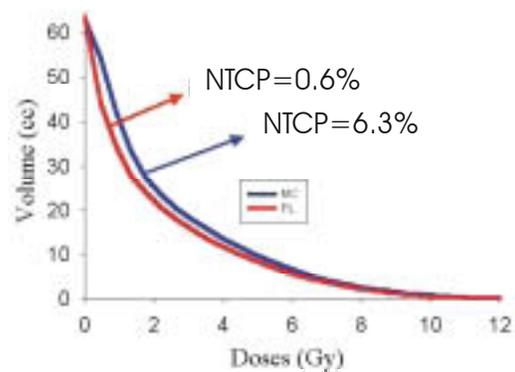


Fig. 5. DVH of the organ at risk (brain stem) showing the differences in NTCP when calculated with Monte Carlo (MC) and a treatment planning system (PL).

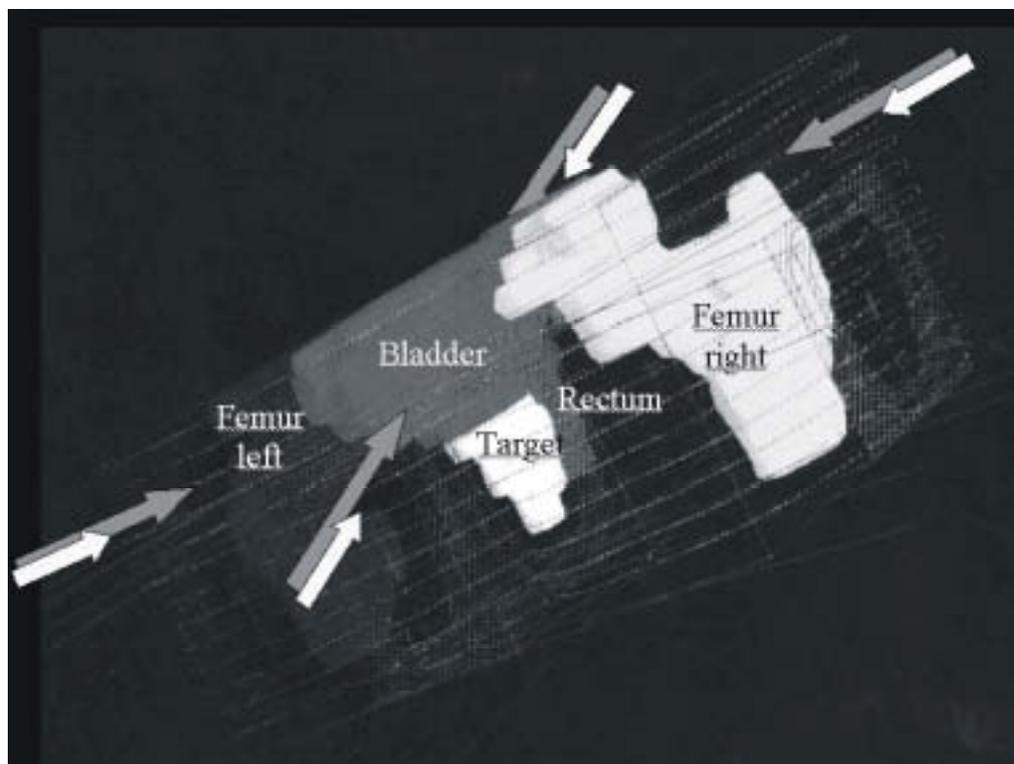


Fig. 6. Schematic of the treatment showing the 4 + 4 technique.

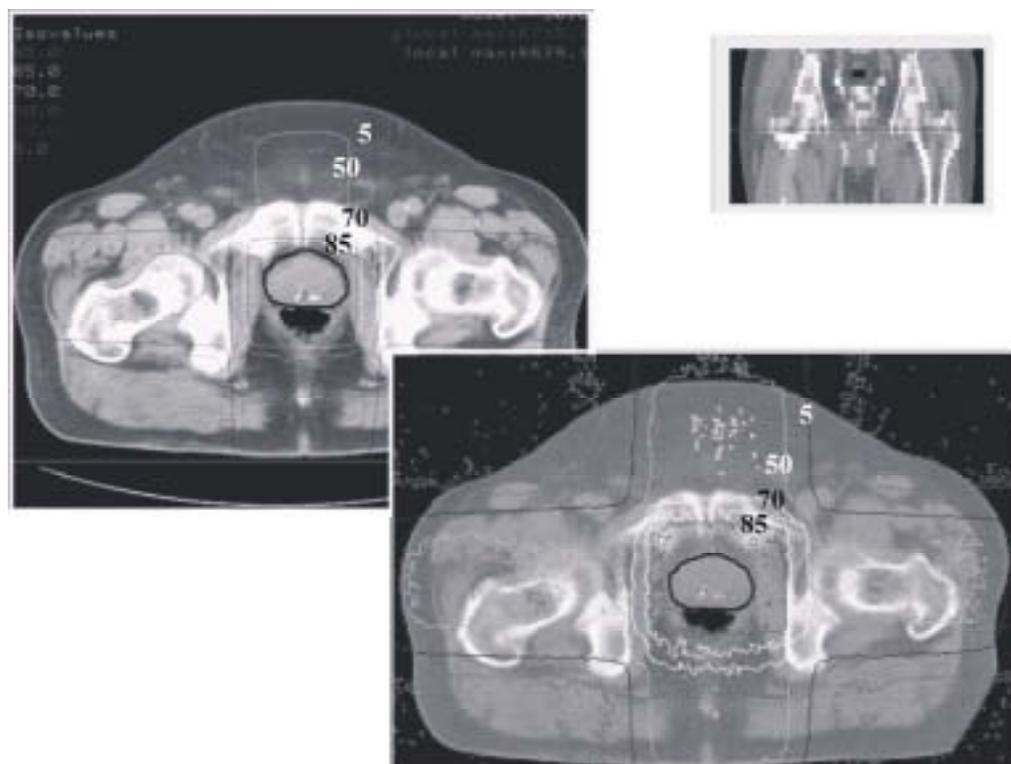


Fig. 7. Dose distribution in the axial plane. The isodoses line are similar around the target. However, differences can be found between Monte Carlo (down) and a conventional TPS (up) on the 50% isodose.

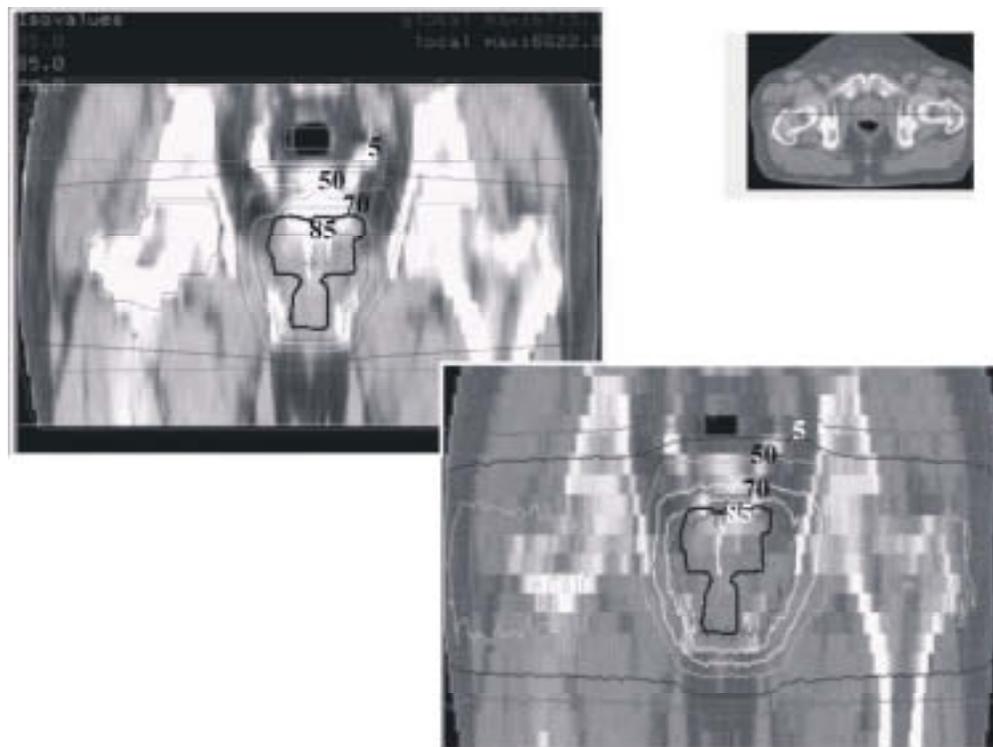


Fig. 8. Dose distribution in the coronal plane. As like in figure 7, some differences can be found between Monte Carlo (down) and a conventional TPS (up) on the femoral heads.

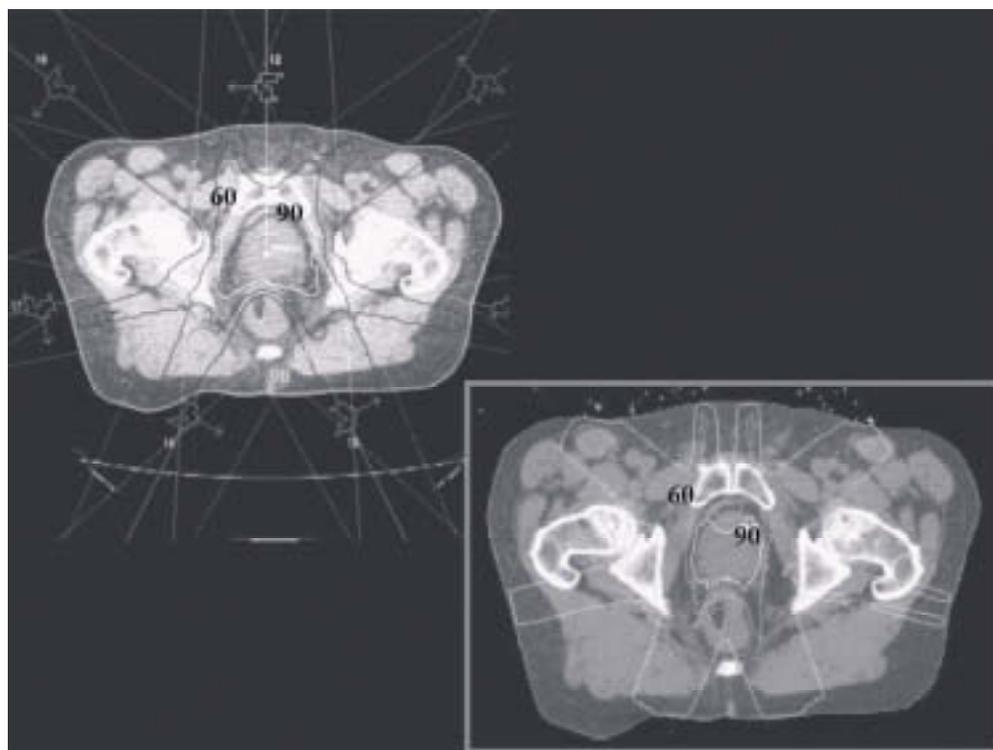


Fig. 9. Dose distribution comparison between a conventional TPS (up) and Monte Carlo (down) for an IMRT treatment. Some differences can be observed. It is noticeable the influence of the pelvic bone interfaces.

CONCLUSIONS

With this new procedure developed by our group we can confirm that the CPU time is adequate for MC verification. Considering this point of view the distributed process is perfectly suitable for MC calculations especially in radiosurgery and IMRT.

In this sense the distributed process is very suitable for MC calculation, especially in radiosurgery and IMRT.

The MC method represents a powerful tool for verification and conduction of procedures of mainly in atypical situations.

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