# RADIOTHERAPY PROCESS INTEGRATION USING A COMPACT PHOTON SOURCE TOGETHER WITH FLUENCE CONTROL AND PATIENT IMAGING

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Abstract

Process integration, combining diagnostic, simulation and therapy in a single instrument will overcome many limitations in cancer care. But it asks for a miniturization of the high energy photon source. Such effort could open unforeseen developments as mobile "global therapy" units. Parameter values for the photon source will be achieved with an X-band linac integrated with target and collimator as a shielded compact sub-unit.

## I. NEEDS AND OBJECTIVES

### *A. The needs (user's point of view):*

The therapeutic treatment consists of 3 phases: diagnostic imaging, simulation of the dose delivery conditions, dose delivery. It is done today with 3 different instruments with inherent problems such as induced incompatibility of materials (today without standards) and difficulties in coordinating the phases into a coherent process. The use of multiple, independent systems creates an incoherence resulting in faults of treatment efficacy and safety, and a waste of time and discomfort for the patient. And, no effective treatment exists today for concave shape of tumor volume (prostate).

One needs to *integrate the 3 phases of treatment into one system and instrument.* The advantages are in level with industrial effort required from the manufacturer and adaptation required from users: (1) the necessary control for a dynamic conformal therapy adapted to concave shape of tumor volume becomes possible, (2) *the precise and simple control position for each treatment* becomes feasible, (3) sources of errors due to use of different instruments and non medical data transfer are eliminated, (4) the global cost for the process is decreased as one instrument on one site replaces 3 instruments on 3 sites.

Such instrument insures 'unité de lieu' (same place whatever the phase) but does not require time limitation and the imaging resources can be used at different levels of sophistication for a typical sequence as: diagnostic (3 arbitrary time units) – simulation (later, 2 time units) – treatments (later, over several weeks, for ex. 15 x 1 time units). Then the instrument is used at 75% for therapy and at 25% for imaging preparations.

### *B. The objectives (technological point of view):*

A new concept for the delivery of dynamic conformal radiotherapy has been recently proposed under the "tomotherapy" name [1]. It arouses much interest as it proposes a combination in a single instrument of diagnostic and therapy. The proposal is to deliver therapeutic radation following the spiral scan method used on recent CT imaging units, as a modulated slice, while the patient moves through the gantry. This introduces a synergy between the large medical diagnostic engineering resources and the more modest medical therapy ones. However, weight and cost are large when the ratio of useful to available radiation is low as is the case in a "slice" delivery. Also the patient moves under a potentially harmful treatment.

One alternative keeps the conical radiation delivery and uses a multileaf collimator (MLC) and a pulse to pulse control [2]. See figure 1. This in turn leads to rather intricate 3D dosimetry. The emerging photon beam fluence is detected either behind the patient as is the case in portal imaging, or more simply before the patient at the MLC exit, as the patient view is available by perpendicular CT scan (meaning only a one-fourth rotation time delay). It acts as the key control for the treatment process. Dose delivery to a fixed patient combines slow longitudinal mechanical motions in the MLC [3] with quick azimuthal time modulation by pulse to pulse fast on/off electronic control. For treatment, patient view could be restricted to a slice (2D scan). For full diagnostic view the patient would move (3D helical scan).

Even if it is too soon to decide what is the solution best fitted to radiotherapy, it is clear that the association of a photon source with field fluence control and patient diagnostic imaging in a single global radiotherapy instrument is the key for more precise and simpler treatment process. This supposes a *miniturisation* of the source to make it much lighter and to allow its rotation on imaging rings of the fourth generation in continuous rotation. The necessary electrical power must be reduced to transfer through circular contacts. A first design presented below uses a 9.3 GHz linac to accelerate electrons at 6 MeV, a cooled target and a multi-leaf collimator (MLC).

### *C. Let us dream (a little) to future extensions*

Such a miniaturized instrument able to integrate the whole radiotherapy process could be used as a "global radiotherapy mobile unit" similar to the radiology mobile unit used today. The fixed protection necessary for the outside world would be replaced by a dedicated and eventually protected parking space with access limitation or `site.' Such an approach could benefit under-equipped regions. For example, 2 mobile units (for reliability) would move from site to site within a large district area around a central hospital.

## II. PHOTON SOURCE PARAMETERS VALUES

The following table summarizes the choices for the main parameter values. The photon megavoltage is modest, a necessary condition for compacity then low cost and general applicability (including optional mobile unit). In fact, it is this modest value compatible with many available beam input ports or with a basic generalization of therapy which allows the high frequency choice with dramatic effect on weight, cost, mobility. Before, conventional radiotherapy products were simpler (without imag-

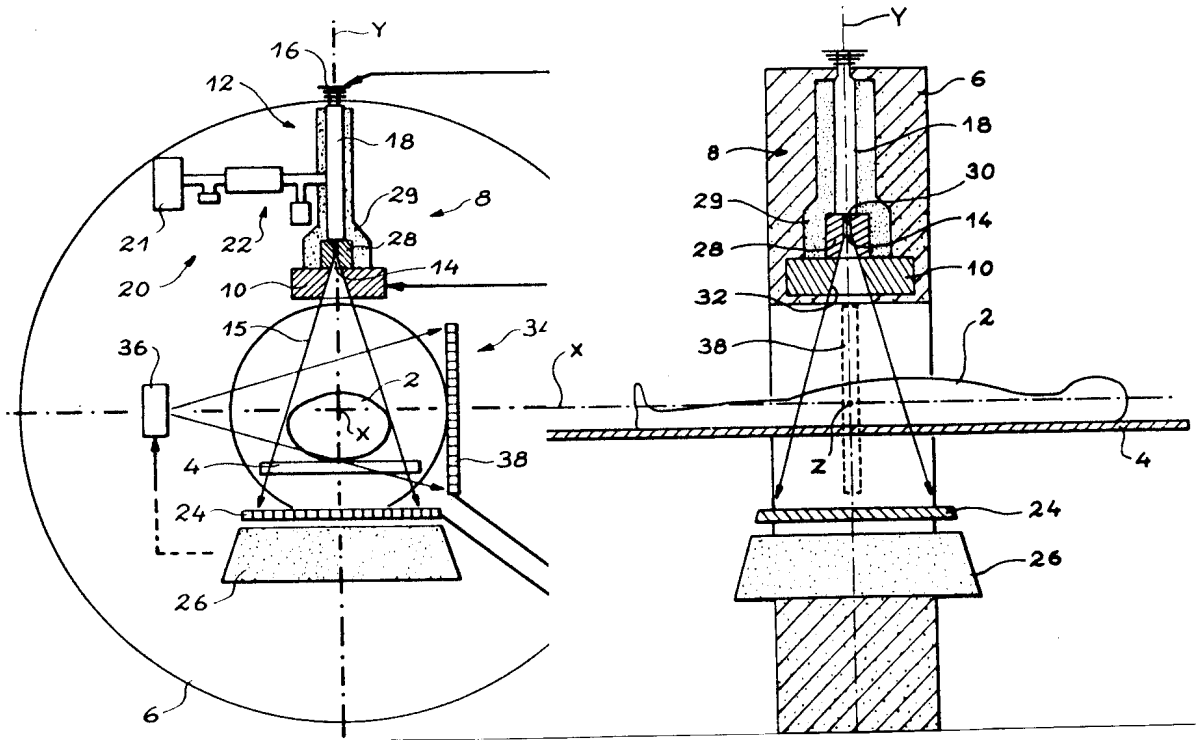


Figure 1. The 3D delivery case: the rotating ring which supports the photon source and the CT scan is seen in plane and in profile.

ing capabilities) but had to cover a whole range of energy up to 18 MeV, and this precluded RF higher frequency use.

| Fundamental parameters | Values |
|---|---|
| Treatment volume envelope radius | 200 mm |
| Patient clearance radius | 500 mm |
| Source to axis distance | 750 mm |
| Maximum external radius | 1300 mm |
| Electron energy (photon megavoltage) | 6 MeV |
| Electron beam peak/mean power | 0.5 MW/0.38kW |
| High Energy Photon Source weight | <200 kg |
| HEPS electrical power required | <3kW |
| Dose rate on-axis at isocentre | >4 Gy/mn |

## III. X-BAND COMPACT LINAC AND SHIELDED TARGET AND COLLIMATION

The predesign uses the following formulae:

The dose on-axis D in Gray/min for a mean electron current $i$ in $\mu$A at the energy V in MeV, delivered at the distance d in m, is empirically given by [4]:

$$D = 9 \times 10^{-}4 \times i \times \frac{V^{2.58}}{d^2}$$

The mean current $i$ in $\mu$A is related to the pulse length $\tau$ in $\mu$s and to the repetition frequency $F$ in Hz and to the peak current $I$ in A:

$$i = \tau \times F \times I$$

The peak current $I$ in A for the total peak power $P$ in MW and for the Joule power $P_J$ (necessary to create the electrical field leading to the energy gain) in MW and for the energy $V$ in MeV is:

$$I = \frac{P - P_J}{V}$$

The Joule power loss $P_J$ in MW leading to the energy gain $V$ in MeV for the shunt impedance per unit length $Z$ in M$\Omega$/m and for an acceleration length L in m is:

$$P_J = \frac{V^2}{ZL}$$

with $P$ = 1.2 MW at the section input (the magnetron delivering >1.3 MW), for $V$ = 6 MeV and $Z$ = 130 M $\Omega$/m at 9.3 Ghz (SUPERFISH code gives more for the profile chosen but surface roughness and coupling and dynamics lower it), and for L = 0.4m: $P_J$ = 0.69 MW, $I$ = 0.083 A.

At $\tau$ = 3 $\mu$s and $F$ = 250 Hz: $i$ = 62 $\mu$A.

With d = 0.75m and $V$ = 6 MeV: D = 10.1Gy/mn, more than 4Gy/mn required—and available even with field equalization (not used in the present project).

From the parameters defined above, one has to design the integrated electron linac plus the target plus the collimator, surrounded by a minimum shielding. The difficulties lie more in the integration than in any specific subcomponent. However, the magnetron, the triode gun, the beam capture and the beam radial control with minimum energy loss, the technology of target cooling presents challenges. The concept is to put the cooled target inside a rather thick shielding, then to shield more lightly the linac structure after assessment of the low stray X-rays production along it, by dynamic simulation to minimize beam loss

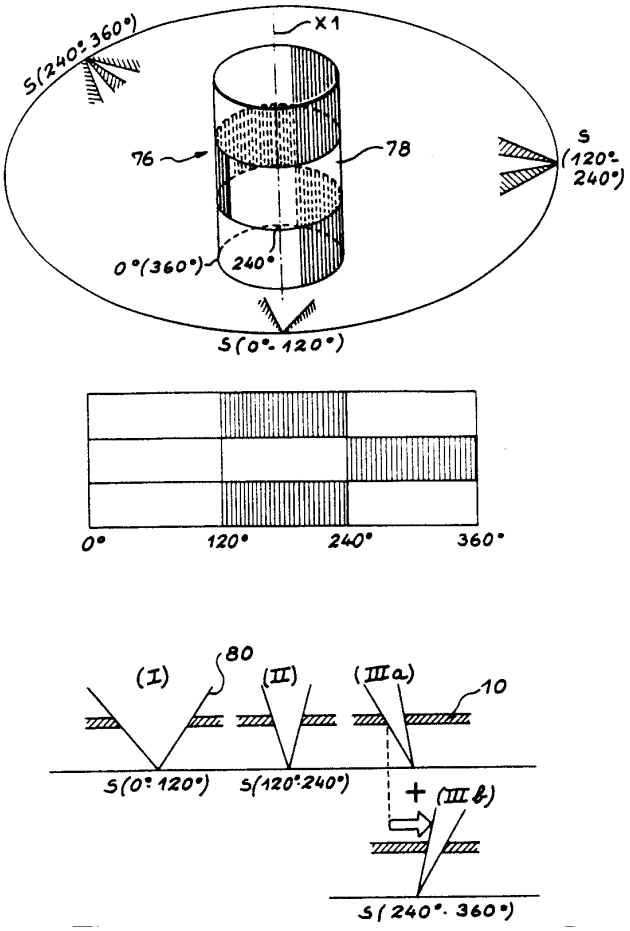


Figure 2. represents an oversimplified treatement of a cylinder mapped as rectangular 3 x 3 zones. The shaded ones are protected by proper collimation including at right summation (which requires then 2 turns around the cylinder).

before direct measurements. The MLC is set as near target as possible, to be part of the shielding and to be compact, but one is limited by the penumbra. The protected empty box between primary collimator and MLC upper boundary is used for optical simulation and optional beam flattening filter.

## IV. CONCLUSION

This approach offers a solution to catch up with the needs on two lines: (i) the very sophisticated instrument aimed to better conform to tumoral sites following multiports delivery techniques developed in centers of excellence, (ii) the very compact and relatively cheap instrument able to cope with the 3 treatment stages necessary to deliver safe radiotherapy. Eventually such an instrument could move in a truck to cover a large area when few medical centers are available.

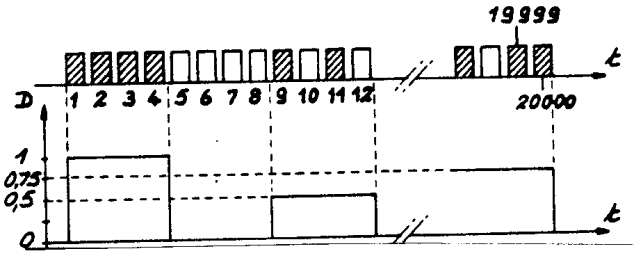


Figure 3. represents the fast pulse on/off control which leads to azimuthal dose modulation—here oversimplified to 4 possible levels only.

## References

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