Digitally Reconstructed Radiographs (Drr) in Radiotherapy: Natural Contrast Enhancement For Virtual Simulation.

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ABSTRACT

As a tool for image reconstruction treatment planning systems offer the possibility to generate sagittal, coronar, and arbitraryly orientated slices on the basis of primary transversal slices or spiral scans. These additional slices can be used for calculation of dose distribution in radiotherapy. For better review of dose distribution and beam directions according to patient's anatomy and divergent beam geometry new views are reconstructed by ray tracing from the source through the patient: DRR (digitally reconstructed radiography). This technique is named "virtual simulation" with reference to the conventional technique by using a specific X-ray-equipment. Compared to conventional X-ray-recordings the contrast is reduced by reason of the different image content: absorption- or transmission values vs. density values (Hounsfield Units HU). This reduction of contrast depends on the X-ray-energy used for CT-scanning, and can be restituted by special filter functions, the so called LQO-filter [1]. According to HU units and X-ray energy an attenuation coefficient is added. The visual impression of reconstructed CT-scans then is comparable to conventional planar X-ray-images, especially for the bone-soft tissue contrast. This is demonstrated by some examples, showing excellent contrast and recognizability of details.

If the X-ray-energy for simulation is increased up to 5 MV (comparable to a 15 MeV-X treatment beam) the contrast in images is reduced progressively, as it is well known from therapy verification films with poor contrast.

INTRODUCTION

In radiotherapy CT-scans of patients are used for treatment planning: The HU can be recalculated into electron density basically used for the calculation of dose distribution. Modern treatment planning systems offer tools to reconstruct sagittal, coronar and arbitraryly orientated views of the patient for better visualisation and calculation of dose distribution.

In commercially available systems various concepts for contrast enhancement are offered: a. filter functions; b. window setting technique; c. suppression of information; d. modification of HU-number; e. maximum value method; f. path length method.

For more accurate assessment of dose distribution and treatment beam directions according to anatomic structures a ray-tracing-method is applied. As a result we have digitally reconstructed radiographes (DRR). The combination of DRR's and projected treatment fields is called "virtual simulation", according to the conventional method using a specific X-ray equipment, called "simulator". But in contrast to simulator images the CT-scan-reconstructions contain HU as information instead of absorption values. This is equivalent to an enormous reduction of contrast, depending on X-ray-energy. This factor varies from 2 to 20 when X-ray-energy decreases from 100 kV to 50 kV. This loss of contrast can be compensated for by a LQO-filter [1] : The contrast in CT-scans is shifted to that of normal X-ray-images taken with an appropriate X-ray-energy. The modified image seems to be optimal for the purpose of "virtual simulation", but not for dose calculation because of the nonlinear shift of the HU and the inherent beam divergence in the reconstructed image.

To increase contrast in DRRs the following changes in HU - units (or density values) are in use in commercial planning systems: total suppression of parts of volume information; geometrically orientated suppression of information; anatomically guided suppression of information; increase of HU-numbers for selected volumes; window setting technique.

MATERIAL AND METHOD

The principle of X-ray-imaging and CT-scanning is identical: during passing an object X-rays were absorbed or scattered; the transmitted part is used for exposure of a film or calculation of HU in case of a CT [2, 3]. This results in transmission values, visible as different grey values within the X-ray-image or HU characterizing density values within a CT-scan [3].

The ratio of contrast for X-ray-images vs. CT-scans can be estimated by:

$$r = \frac{\exp(\mu_2, E) / \exp(\mu_1, E)}{\left| HU_{M2} - HU_{M1} \right| \cdot 10^{-3}}$$

μ_1, μ_2 : attenuation coefficients for medium 1,2 [2, 5]

E : X-ray-energy

HU : Hounsfield unit

M_1, M_2 : medium 1, 2 .

For an X-ray-energy of 60 kV (or ≈ 20 keV_{eff}) [4] the contrast reduction bone/water is estimated to r ≈ 10. For higher energies this reduction factor decreases, r (^{60}Co) ≈ 0.6. This implies an image contrast smaller than the original value within the CT-scan. Portal images taken during radiotherapy with cobalt units or linear accelerators confirm this physical phenomena.

Tabulated values of linear absorption coefficients [5, 6, 7, 8] for equivalent photon energies [4] are used to calculate the transmission for particular röntgen energies and materials. The normalized inverse transmission T_i,

$$T_i = \exp(\mu(E_i, HU(\rho))),$$

shows in good agreement a linear-quadratic behaviour for a particular energy:

$$k_i = a_{1,i} + a_{2,i} \cdot HU + a_{3,i} \cdot HU^2,$$

k_i : correction factor for a voxel taking into consideration a particular energy interval (see Tab. 1)

$a_{1,i}$, $a_{2,i}$, $a_{3,i}$: coefficients

HU : Hounsfield units

μ : linear absorption coefficient

ρ : physical density

E_i : particular X-ray energy.

For simplification the spectrum of X-ray-energies of interest is divided into four intervals (see Tab. 1) :

Interval #	X-ray-energy [kV]	E_g (eff) [keV]	Coefficients a_1	a_2	a_3	Corr.-factor
00	35 – 40	22	31.234	-0.3414	0.0002929	k_{00}
0	45 - 50	25	4.976	-0.0337	$2.942 \cdot 10^{-5}$	k_0
I	55 - 65	28	1.763	-0.0073	$6.926 \cdot 10^{-6}$	k_I
II	70 - 80	30	1.227	-0.002269	$2.433 \cdot 10^{-6}$	k_{II}
III	85 - 100	40	1.032	$-2.0242 \cdot 10^{-4}$	$4.573 \cdot 10^{-7}$	k_{III}
IV	130	60	1.004	$1.5054 \cdot 10^{-4}$	$6.594 \cdot 10^{-8}$	k_{IV}
V	180	80	0.9998	$1.908 \cdot 10^{-4}$	$5.434 \cdot 10^{-9}$	k_V
VI	330	150	0.998	$1.814 \cdot 10^{-4}$	$-2.196 \cdot 10^{-8}$	k_{VI}
VII	450	200	0.9981	$1.7057 \cdot 10^{-4}$	$-2.436 \cdot 10^{-8}$	k_{VI}
VIII	900	400	0.9985	$1.3204 \cdot 10^{-4}$	$-2.059 \cdot 10^{-8}$	k_{VIII}

Table 1. *X-ray-energy-intervals and parameters.*

k_I : 55 – 65 kV;
k_{II} : 70 – 80 kV;
k_{III} : 85 – 100 kV;
k_{IV} : 130 kV.

The correction factors k_{00}, k_0 and k_V – k_{VIII} are just listed up to demonstrate the variations of contrast enhancement for bone material: low X-ray energies (< 55 kV) lead to high contrast values, high X-ray energies (> 130 kV) reduce the contrast.

For reconstruction of orthogonal or oblique orientated projections the HU – value of a voxel is multiplied by the appropriate k – value and according to the ray - tracing – method, summed up[3]:

$$p_{l,m,n} = \Sigma\, HU(x_l, y_m, z_n) \cdot \Delta_{xyz} \cdot k_{I,II,III,IV}$$

$p_{l,m,n}$: reconstructed grey value of a voxel within the new projection
x_l, y_m, z_n : coordinates of a voxel (original)
Δ_{xyz} : size of a voxel
$k_{I,II,III,IV}$: correction factor according to energy interval mostly used in röntgen simulation radiography.

The modified HU – numbers of the reconstructed image have to be adopted to the grey scale used in the planning system, i. e. HU(air) = -1000; HU(bone) ≈ 1500.

EXAMPLES

In radiotherapy simulation 73 kV X-ray-energy is used often. The image contrast is defined by different absorption coefficients of bone, soft tissue, fat, and air. In Fig.1 a lateral X-ray-simulator image of the head is shown. Clear differentiation of bone and soft tissue is possible. For treatment planning a set of CT-scans was taken. Because of the slice distance some details got lost when reconstructing the lateral DRR (Fig. 2) coinciding with the lateral X-ray-image. This loss of contrast can be observed by comparison with the conventional X-ray-image.

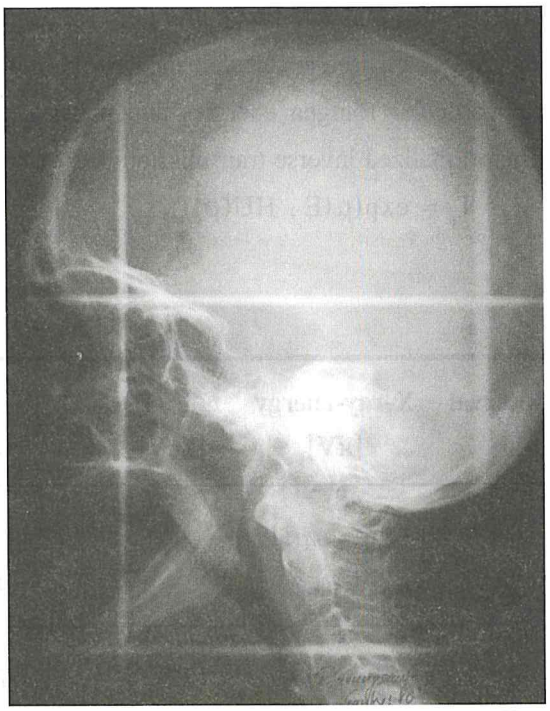


Fig. 1. *Simulation image (head, lat. view).*

The application of the correction factor k_{II} (70 – 80 kV: medium X-ray energy) restitutes and enhances the contrast (Fig. 3).

If low (55 – 60 kV) or high X-ray-energies (85 – 100 kV) have to be simulated, the correction factor k_I or k_{III} have to be applied respectively (see Tab. 1).

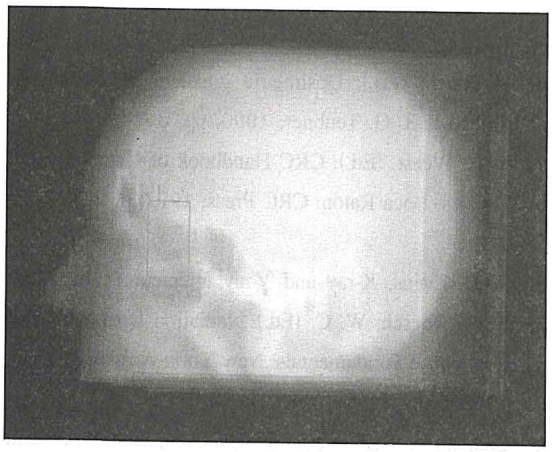


Fig. 2. *Digitally reconstructed radiograph.*

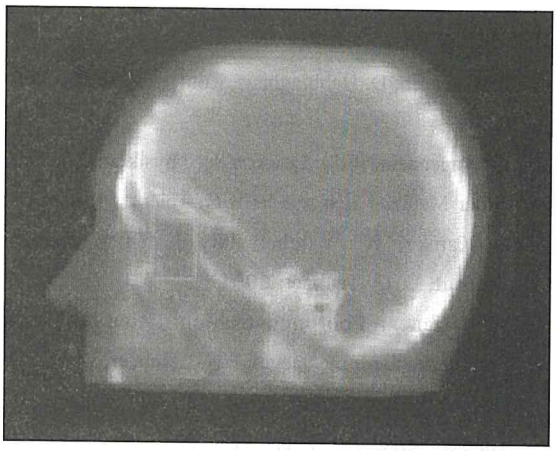


Fig. 3. *DRR incl. correction factor* k_{II} .

RESULTS AND DISCUSSION

The introduction of the so called "virtual simulation" into radiotherapy treatment planning procedure on the basis of sets of CT-scans makes it mandatory to enhance the image contrast. For better orientation and localisation, especially the soft tissue/bone contrast has to be increased. The application of LQO-filter-functions is not time consuming, when the multiplication of voxel values (HU) is done by precalculated matrix operations.

The preliminary use in routine showed, that the modified visual impression of the DRRs seems to be a very helpful tool in treatment planning. It has been successfully implemented in a commercially available treatment planning system [9] to support better visibility of landmarks and bone/soft tissue/lung contrast (s. a. Fig. 3).

The same algorithm can be used to calculate DRRs similar to verification images done by a treatment beam, i. e. 15 MeV-X: compared to CT-scans the image contrast is clearly reduced.

For better adoption of the simulated X-ray-energy a continous recalculation of the correction factor k is possible, instead of dividing the X-ray-spectrum into several intervals:

$$k = f_1 + f_2 \cdot HU + f_3 \cdot HU^2,$$

with: $f_i = f(a_i, E(\text{X-ray[kV]}))$, $i = 1, 2, 3.$

Further developments may lead to real-time calculation and visualisation of enhanced DRRs during changing the direction of view, strictly speaking horizontal and vertical rotation (gantry and table rotation or change of isocenter). This implies a 3D-presentation of images because of the different movement of voxels inside the CT-scan or body, respectively, according to the center of rotation.

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