The Monte Carlo Method: Versatility Unbounded In A Dynamic Computing World
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MONTE CARLO INVESTIGATION OF AN EBT SCANNER'S FAN-SHAPED BEAM

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ABSTRACT

Electron beam tomography (EBT) provides good quality imaging necessary to assess, e.g. the seriousness of coronary artery disorders. Effective dose to the patient, related to radiation risk, is also important in such examinations. Monte Carlo (MC) simulation enables obtaining conversion coefficients to derive effective dose from the CTDI (computed tomography dose index), a dosimetric quantity easily measured with a pencil ionization chamber. For appropriate MC simulation, information is required on relevant parts of the scanner's geometry and on the radiation quality of its fan-shaped beam. In EBT a narrow electron beam aimed at a tungsten anode ring around the patient creates a fan-shaped beam of X rays. This paper describes the modification of the SOURCE subroutine in the code MCNP-4C to deal with the fan beam moving along a 210° arc in the Siemens Evolution XP scanner. Furthermore it is described how, based on matching MCNP results to manufacturer's data and measurements on dose distribution in a transverse slice of a CT dosimetry phantom, the shape of the photon intensity profile along the fan should be chosen. Unlike often-applied beam shape filtering in conventional CT, the best match is obtained here with a uniform distribution of photon intensity over the fan angle.

Key Words: Monte Carlo; Dosimetry; Diagnostic Radiology; Electron Beam Tomography

1 INTRODUCTION

Computed tomography (CT) has become a standard tool in diagnostic and interventional radiology. By rotating an X-ray tube around the patient and measuring attenuation of the radiation in the body with a ring of detectors an image of the tissues in a transverse plane is obtained. Moving the patient along the longitudinal axis through the CT scanner results in a series of images (transversal slices) of an anatomical region, containing necessary information for medical diagnosis or guidance during an intervention. Despite developments to increase the speed of scanning (0.5 s per image is possible nowadays), conventional CT still is less suitable for imaging moving organs like the heart. In contrast, electron beam tomography (EBT), a technology already conceived in the 1980s, largely reduces motion blur by very short rotation

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time and starting rotations at a constant position in the heartbeat cycle (triggering). In an EBT scanner a narrow beam of electrons is emitted from an electron gun located at distance on the scanner's longitudinal axis. By electromagnetic steering the electron beam is made to move along the mantle of a cone-shaped vacuum chamber around this axis and ending at a tungsten target ring (anode). At the place of impact X rays are generated, which form a fan-shaped beam rotating in the plane of the ring and irradiating the patient lying in the center. As one rotation can be completed in 100 ms, even at fast rates (up to 120 beats per minute) the patient's heartbeat can be used to trigger the scanning cycles. More information on principles and applications of EBT can be found on the World Wide Web [1].

Coronary calcium is a good marker for coronary artery diseases like arteriosclerosis and EBT seems an excellent imaging technique to detect and quantify coronary calcifications [2-4]. To such purpose a Siemens Evolution XP EBT scanner (Imatron Inc., San Francisco, USA) was used at Erasmus Medical Center in a cardiology study comprising a large number of adults. Next to obtaining information of medical interest it is important to know what will be the radiation burden to the patients undergoing the examination. The risk of late radiation effects is best estimated by the effective dose, a weighted sum of organ doses [5]. As organ doses are difficult to measure they are usually derived by applying dose conversion coefficients (DCC) to a physical quantity that can be measured, e.g. the computed tomography dose index (CTDI). Basically, this quantity is the average dose (free in air) on the central longitudinal axis of the CT scanner (the center of rotation, COR) due to a single rotation, and can be measured with a pencil ionization chamber. DCC can be obtained through Monte Carlo (MC) simulation of radiation transport. In simulations, both CTDI and organ doses to the patient can be calculated. The latter require a suitable model of the patient's anatomy, which usually is approximated by a mathematical phantom [e.g. 6] with simple (2nd order) mathematical functions to describe organ shapes. More realistic "voxel" phantoms are being developed currently [e.g. 7] but their use is not yet widespread.

Next to the patient, the characteristics of the radiation field and relevant parts of the geometry must be known for proper MC simulation of DCC. The present paper investigates the description of the beam of the EBT scanner used for the mentioned coronary calcium study at Erasmus Medical Center. Calculated and measured quantities are compared to verify beam quality parameters. It also describes the modification of the MC code, which was necessary to enable simulation of the moving X-ray source in CT and EBT scanners. (In contrast to CT where the X-ray source goes full circle, in EBT the circling electron beam generates the fan beam of X rays on a circle segment (210º arc) only.) Other investigators simulating CT have omitted the motion but used as an approximation a number of fixed line sources, equally spaced on a circle and parallel to the central longitudinal scanner axis (length equal to the slice thickness) [8]. A moving fan-shaped beam, however, is not very difficult to implement.

2 METHODS

2.1 Simulations

Measured data suitable for verification of the radiation quality of the simulated beam comprise the half value layer (HVL) and a dose distribution in a standard CT body dosimetry phantom, both supplied by the manufacturer of the EBT scanner.

2.1.1 HVL

The HVL is the thickness of an aluminum sheet, which reduces the intensity of a radiation beam to half its undisturbed value. It is determined by inserting sheets of increasing thickness between the source and a detector under prescribed conditions [9], which can be translated straightforward into an MC simulation if the X-ray spectrum is known, and interpolating the measured or simulated readings. The manufacturer states an HVL of 10.4 mm Al.

2.1.2 Dose distribution in a CT dosimetry phantom

A CT body dosimetry phantom is a polymethylmetacrylate (PMMA) cylinder of 32 cm diameter and 15 cm length [10]. Cylindrical cavities (0.6 cm diameter) are present, running parallel to the central axis, with centers at 1 cm distance from the edge of the cylinder and 45º spacing. When the phantom is placed on the central axis of a CT or EBT scanner the cavity positions can be named after the points on a compass. N(orth) and S(outh) for top and bottom positions, corresponding with the belly and back of a supine patient. E(ast) and W(est) for the left- and right-hand side, respectively. NE, NW, SW and SE are the positions in between. Also in the center of the phantom a cavity is present. A pencil ionization chamber can be put in any cavity to measure dose. The manufacturer specifies the values 2.0, 9.2, 12.8 and 9.2 mGy/100 mAs at the N, E, S and W position, respectively. At the center of the phantom 3.0 mGy/100 mAs is measured. These values hold for one rotation at 130 kV and for a slice thickness of 10 mm halfway the length of the phantom.

The measured dose distribution can now be expressed in percentage relative to the central dose. A similar distribution can be calculated by MC simulation. In first instance the parameters that determine the beam quality are kept at their nominal value. If necessary they can be changed to match measured and calculated dose distributions.

In the MC simulations kerma (dose) in air in the central volumes of the cavities of the CT phantom is calculated, corresponding with the use of a 3 cm^3 pencil ionization chamber. The elemental composition of PMMA is 8.05% H, 59.99% C and 31.96% O by weight, at density 1.17 g/cm^3. The composition of air (density 1.2×10^{-3} g/cm^3) is 75.5% N, 23.2% O and 1.3% Ar by weight.

2.2 The MC Code

The general purpose MC code MCNP, version 4C [11] was used with standard data libraries and default settings on a DEC Alphastation XP900 workstation operating under UNIX. The "F6 tally" for photons was used, recording dose per emitted source particle averaged over a target volume employing a factor of 1.6022×10^{-10} to convert default MeV/g output to Gy. Per simulation run 10^7 starting photons were considered, yielding estimated statistical uncertainty of 0.3%-0.9% in dose results.

2.3 User-Supplied Subroutine to Simulate a Moving CT/EBT Source

Besides providing a number of types of particle source frequently needed, MCNP offers the possibility of entering a user-supplied source. This enlarges flexibility, but requires recompilation of the fortran code. Modifications to the code can be introduced by means of a patch file. The listing of such a patch file, with which to create code in the subroutine "source.f" of MCNP for the present application is given in Appendix A.

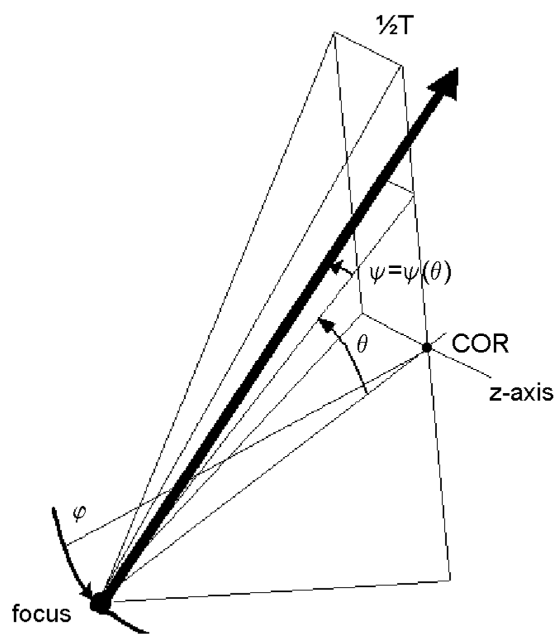

Figure 1. Description of the photon source for the present EBT scanner. The direction into which a photon is emitted falls at random within the fan shaped beam, the thickness of which is T at the center of rotation (COR) on the longitudinal axis of the scanner (z-axis). In a plane perpendicular to this axis, first the position of the focus is selected on the anode ring ($-15^o \leq \varphi \leq 195^o$). Then, in the same plane, angle θ (between ±15º) is chosen within the fixed fan angle of 30º and taking account of the (measured) intensity distribution along the fan. Finally, the angle ψ is chosen in perpendicular direction. The maximum of ψ is determined by the collimation to slice thickness T, and it also depends on θ.

To simulate a rotating fan-shaped beam, first a point is selected on a circle (CT) or circle segment (EBT) according to a uniform probability distribution, thus defining angle phi or the temporary position of the X-ray source (Fig. 1). Alternatively, one fixed position may be chosen, e.g. to simulate the so-called scanview or scoutview mode available on some CT scanners. Next, an angle (theta) to the left or right of the line connecting the point (focus) to the circle's center (COR) is selected within the fan angle and according to a certain intensity distribution (distribution 2). An angle (psi) in the perpendicular plane is selected uniformly within the width of the beam, which is determined by the desired slice thickness T. Particle energy is selected from the X-ray spectrum (distribution 1). Distributions 1 and 2 must be given in the MCNP input file. This also holds for geometrical variables (like r: focus-to-COR distance; z: slice position with respect to the patient; psi: angle for slice thickness) which are passed on in the common block arrays rdum and idum. The value of idum(6) determines whether CT or EBT is simulated (source on full or partial circle, respectively). The value of rdum(4) determines whether the focus moves or is fixed for scan- or scoutview.

2.4 The EBT Scanner

2.4.1 Geometry

A transversal cross section of the EBT scanner is schematically shown in Fig. 2. For the simulations only the most essential elements of the EBT scanner are modeled. Basically, an air-filled cylinder about the scanner's longitudinal axis (z-axis) is considered, which is just a little taller than the reference patient of 170 cm and has a radius equal to the distance from the X-ray source to the COR (90 cm). The X-ray source is a (moving) point source on the mantle of this cylinder. The anode ring itself is not modeled. (In reality, there are four tungsten anode rings close together as the EBT scanner can be used in multi-slice mode, i.e. the electron beam is aimed to complete the circular paths on the four rings in sequence before the table with the patient is moved for a next set of four images. This speeds up the scan procedure even further. In the present study, however, single slice mode is selected, employing one anode ring only before table movement.) The detector ring is also omitted. The collimators exist of two cylinders sharing the same z-axis with inner diameter of 60 cm and 1 mm thick. The cylinders can be moved towards or away from each other to set the desired slice thickness of 1.5, 3, 6 or 10 mm at the COR. Then, both cylinders can be moved simultaneously along the z-axis to assume a desired slice position on the patient or the CT dosimetry phantom. The plane in the middle of the gap between the cylinders is the central plane of the fan-shaped beam. Particles are fired from the X-ray source (focus) within the diverging boundary of the fan. Particles entering the mantle of the cylindrical collimators are killed, except in a few simulation runs when the influence of (back) scatter from the environment was evaluated. In those cases the thickness of the collimators was, rather arbitrarily, enlarged to 20 cm and the material density set to 1.2 or 2.4 g/cm^3.

2.4.2 X-ray spectrum and intensity distribution along the fan-shaped beam

The EBT scanner is operated at 130 kVp. The X-ray spectrum was provided by the manufacturer and is shown in Fig. 3.

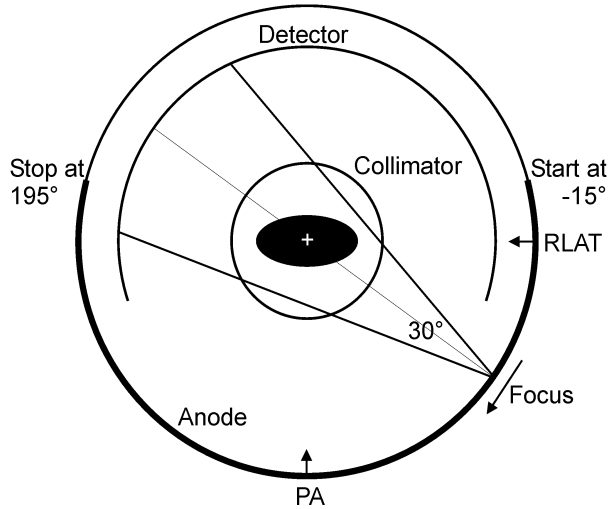

Figure 2. Schematic view of the exposure geometry in the scan plane, perpendicular to the EBT scanner's longitudinal axis (z-axis). At some distance behind this plane a narrow beam of electrons is generated and, by electromagnetic forces, steered to make a circular movement along the 210º arc of the anode ring (tungsten). One sweep at single slice mode (SSM) takes 116 ms, i.e. 100 ms of exposure when traveling along the arc and 16 ms to go from stop to start position. Hitting the target, the electrons cause a fan beam of X rays (30º top angle) emanating from the focus, which turns around the center of rotation (COR, shown as a white cross) to expose an object (phantom or patient, represented by the black elliptical cross section). The focus-to-COR distance is 90 cm. The transmitted radiation is collected on the stationary ring of detectors (67.5 cm radius) and is used for image reconstruction. The width of the fan beam is controlled with ring shaped brass collimators, which can be moved in longitudinal direction to set the desired slice thickness (0.15, 0.3, 0.6 or 1.0 cm). The focus position at 0º, i.e. to the left of the patient, is called the right lateral (RLAT) beam. The focus position at 90º, i.e. at the back of the patient, is called the postero-anterior (PA) beam.

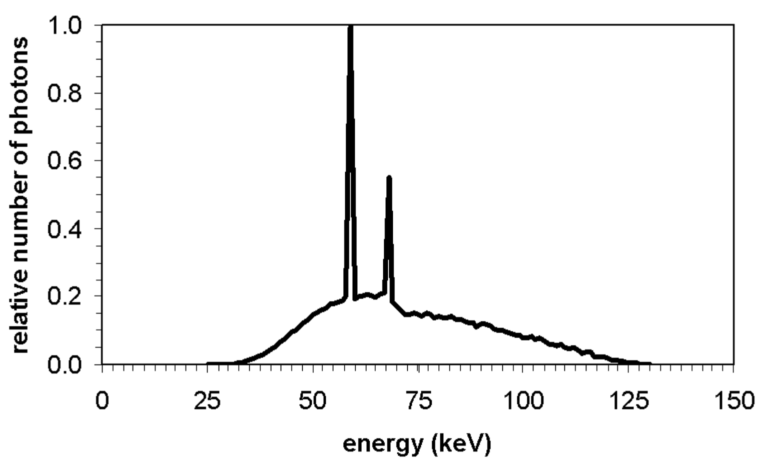

Figure 3. The 130 kVp X-ray spectrum of the EBT scanner, Siemens Evolution XP as provided by the manufacturer, Imatron Inc. The nominal HVL is 10.4 mm Al.

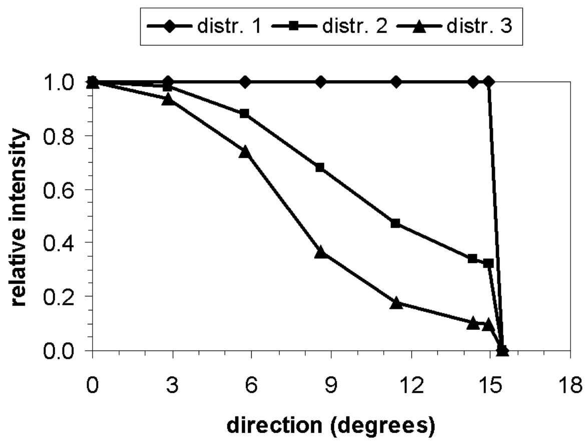

Figure 4. Assumed dose profiles across the fan shaped beam. The relative beam intensity is either uniform (distribution 1) or decreases to 25% (distribution 2) or 10% (distribution 3) with increasing angle θ (see Fig. 1) towards the edges of the fan. Distributions 2 and 3 correspond with dose measurements performed with two different conventional CT scanners [12].

Measurements of dose profiles along the fan of beams of various conventional CT scanners have revealed that loss of intensity occurred towards the edges of the fan beam [12]. Non-uniformity usually is intentional. Manufacturers introduce so-called beam-shaping filters to adjust the photon intensity to the shape of the patient (thick in the center, thinner towards the periphery). As for the Evolution XP scanner no information was available on this subject other than 0.38 mm stainless steel filtering being applied, in the simulations three assumed dose profiles of increasing non-uniformity are considered (Fig. 4). The first profile has a uniform distribution. The other two profiles are based on experience with conventional CT scanners.

3 RESULTS AND DISCUSSION

Using the X-ray spectrum of Fig. 3 for simulated HVL determination with MCNP a value of 10.7 mm Al is derived. This corresponds rather well with the HVL value of 10.4 mm Al given by the manufacturer, the deviation being less than 3%.

It is more difficult to satisfactorily match the calculated and measured dose distributions. Table I shows the results for various cases. Case 0 is the measured dose distribution expressed as percentage of the central dose in the CT dosimetry phantom. The other cases concern calculated results.

In Cases 1 through 3 the X-ray spectrum of Fig. 3 is used in combination with one of the intensity distributions shown in Fig. 4. Compared with Case 0 the uniform distribution of photon intensity across the fan beam (distribution 1) yields the best result (Case 3), indicating that beam shaping effects do not apply to EBT. Still, the peripheral doses then are too low with respect to the central dose, especially toward the north side of the phantom (relative deviations from -9% to -24%). East-West symmetry is as expected.

In principle, an error in the width of the beam may be of influence. This is examined in Cases 4 and 5, where considerably smaller or wider than 30º nominal fan angle is assumed. The size of the fan angle appears to make hardly any difference with regard to the dose distribution of Case 3.

The above results seem to suggest that the X rays in the simulation are a bit too hard, i.e. there are relatively too many high-energy photons in the spectrum. Indeed, a downshift in energy yields improvement of the calculated dose distribution. In Case 6 and Case 7 the spectrum of Fig. 3 is shifted, respectively, 5 keV and 10 keV to the left. Uniform intensity distribution across the fan is assumed. The relative deviations with respect to the measured distribution are reduced to -3% – -22% and +8% – -19%, respectively. Another consequence of softening the spectrum by the downshift in energy, however, is that the calculated HVL decreases. For Case 6 and Case 7 the HVL becomes 9.4 mm Al and 7.9 mm Al, respectively, which is too low with respect to 10.4 mm Al specified by the manufacturer.

Table I. Dose distribution at the peripheral locations on the CT phantom's central transversal plane, expressed as percentage of central dose and (between brackets) as fraction of percentage according to factory measurements. 10 mm slice, one rotation.

Case / Position	N	NE	E	SE	S	SW	W	NW
0. Imatron data	67	-	307	-	427	-	307	-
1. Intensity distrib. 3	25 (0.38)	70	198 (0.65)	279	296 (0.69)	280	197 (0.64)	71
2. Intensity distrib. 2	38 (0.57)	96	220 (0.72)	313	342 (0.80)	313	218 (0.71)	96
3. Intensity distrib. 1	51 (0.77)	122	242 (0.79)	349	389 (0.91)	349	244 (0.79)	122
4. Intensity distrib. 1 fan angle 24º	51 (0.77)	122	245 (0.80)	351	392 (0.92)	351	242 (0.79)	122
5. Intensity distrib. 1 fan angle 36º	51 (0.77)	121	243 (0.79)	349	392 (0.92)	350	243 (0.79)	122
6. Intensity distrib. 1 shift spectrum -5keV	52 (0.78)	128	258 (0.84)	373	415 (0.97)	374	261 (0.85)	130
7. Intensity distrib. 1 shift spectr. -10 keV	54 (0.81)	140	287 (0.93)	416	462 (1.08)	418	290 (0.94)	142
8. Intensity distrib. 1 scatter body present	55 (0.83)	126	244 (0.80)	350	390 (0.91)	350	246 (0.80)	126
9. Combination of Cases 6 and 8	57 (0.85)	133	260 (0.85)	374	416 (0.97)	375	263 (0.86)	134

Scatter from the environment may cause presence of more photons of lower energy than accounted for in the emitted X-ray spectrum. Many elements of the EBT scanner, which cause scatter in reality, are not considered in the simulations. In Case 8 the spectrum of Fig. 3 is used again with uniform intensity distribution across the fan. A scatter body is introduced now, i.e. the thickness of the wall of the cylindrical collimators is increased to 20 cm and the density of the material is set to a factor of 1000 or 2000 larger than the value for air. Unfortunately, it is not simple to place objects like a detector ring because they would obstruct the primary beam, and differently from different focus positions. Thus, in Case 8 there is scatter only from objects not directly in the beam. Little influence of scatter is noticed but, especially at the North position, the contribution to the peripheral dose is definitely increased.

Finally, Case 9 being a combination of softer spectrum (Case 6) and scatter (Case 8) yields a calculated dose distribution with relative deviations of -2% to -15% from the measurements.

Although no information on the reliability of the measured dose distribution is stated, the uncertainty of similar measurements with a pencil ionization chamber by other investigators varied from 3.5% to 24% when irradiating a 3 mm thick slice of the CT dosimetry phantom in EBT [13]. These literature conditions were also simulated (3 mm slice, spectrum of Fig. 3, any of three intensity distributions according to Fig. 4). As can be seen from the results (Fig. 5) the calculated dose distribution for uniform intensity across the beam is in very reasonable agreement with the measurements.

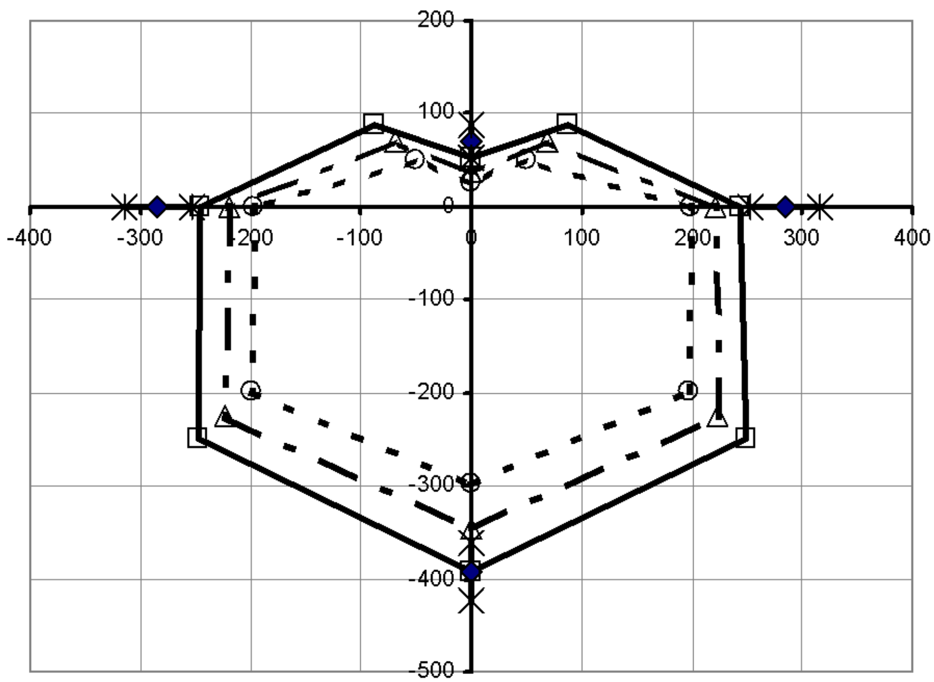

Figure 5. "Radar plot" showing the dose at the various peripheral positions (North clockwise via East through Northwest) on the CT body dosimetry phantom's central transversal plane, expressed as percentage of the midpoint dose, for one rotation and 3 mm slice thickness. <u>Diamonds</u>: measured data with <u>crosses</u> indicating ±1 SD uncertainty ranges [13]. <u>Squares</u>, <u>triangles</u> and <u>circles</u> pertain to intensity distributions 1 through 3 across the fan, as shown in Fig. 4. X-ray spectrum as in Fig. 3.

4 CONCLUSIONS

MC simulation can be a useful tool to do “experiments” on the computer and obtain insight into the physics and into what variables should be considered or can be ignored in a radiation problem. This paper has examined beam quality parameters of an EBT scanner to verify that manufacturer’s specifications are correct, i.e. yield dose values in agreement with measurements. Only then this beam can be used in MC simulations to calculate reliable conversion coefficients to estimate effective dose to patients. First, a method has been developed to properly simulate the movement of the fan beam within the scanner. The subroutine SOURCE for MCNP can be used for EBT and CT in general. From the present investigation it can be concluded that the X-ray spectrum supplied by the manufacturer of the EBT scanner can be used in combination with a uniform photon intensity profile across the beam. No beam shaping effects were found. The accuracy requirements for radiological protection allow uncertainties as met in this investigation. Like for conventional CT there seems to be no need to model the geometry and materials of the scanner in greater detail.

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APPENDIX A

Listing of entrees in a patch file required to create subroutine "source.f" describing a user-defined particle source in MCNP, which simulates a rotating fan-shaped beam of a CT or EBT scanner. (See also Fig. 1.)

```
*define dec,lp64,unix,cheap,pointer,plot,mcplot,gkssim,xlib,xs64
*/
*ident soCT_EBT
*d,so.11
      data i/0/
      save i
c     General CT source, focus goes full circle or part of circle (EBT) or
c                          remains at a fixed position (=scanview)
c     this source subroutine needs input-cards, distr. 1, 2, rdum and idum:
c     distribution 1: erg   (energy of starting particle, MeV)
c     distribution 2: theta (angle in rad in fan-direction)
c     rdum r z psi phi
c     idum cel_1 cel_2 cel_3 cel_4 cel_5 idum(6)
c     where r: focus-COR dist.(cm), psi: half angle of beam in z-dir. (rad),
c      z: slice position (cm), phi: scanview angle (rad, 0=RLAT, pi/2=PA, etc)
c          (nb., scanview if phi is less than 6.5, else rotating source)
c          cel_i with i=1,2,3,4,5 are the cell numbers containing the source;
c          if idum(6)=100 the focus will move less than 2 pi around phantom
c
c     the next if statement is used to initialize,
c     always check sampling, print ksd(i,j) and psf(i,j)
      if (i.eq.0) then
      if (ksd(lksd+12,1).eq.0) ksd(lksd+12,1)=7
      if (ksd(lksd+12,2).eq.0) ksd(lksd+12,2)=7
      if (ksd(lksd+21,1).eq.0) ksd(lksd+21,1)=1
      if (ksd(lksd+21,2).eq.0) ksd(lksd+21,2)=1
      if (ksd(lksd+19,1).ne.0) call norma(1)
      if (ksd(lksd+19,2).ne.0) call norma(2)
      end if
```

```
      wgt=1
      if (rdum(4).lt.6.5) then
       phi=rdum(4)
      else
       if (idum(6).eq.100) then
        phi=(14*rang()-1.)*pie/12
       else
        phi=2.*pie*rang()
       end if
      end if
      xxx=rdum(1)*cos(phi)
      yyy=rdum(1)*sin(phi)
      zzz=rdum(2)
      call smpsrc(theta,2,ix,fr)
      psi=atan(cos(-theta)*tan(rdum(3)))
      psi=-psi+2.*psi*rang()
      uuu=-cos(psi)*cos(phi-theta)
      vvv=-cos(psi)*sin(phi-theta)
      www=sin(psi)
      jsu=0
      do 10 i=1,5
        icl=namchg(1,idum(i))
        call chkcel(icl,2,j)
        if (j.eq.0) go to 20
   10 continue
      call expire(1,'source',
     1 'source is not in any cells on the idum card.')
   20 call smpsrc(erg,1,ix,fr)
      tme=0.
```