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Quantitative CT measurements: the effect of scatter acceptance and filter characteristics on the EMI 7070

Rob B Merritt† and Stafford G Chenery‡

Medical Physics Unit and Department of Radiation Oncology, McGill University, 1650 Cedar Avenue, Montreal, Quebec, Canada H3G 1A4

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Abstract. Non-linearities in projection values on computed tomography (CT) scanners cause corresponding errors in derived Hounsfield unit attenuation measurements. Existing commercial machines have been refined for clinical usefulness but not necessarily for quantitative accuracy. In this report the results of measurements on the EMI 7070 fourth-generation scanner are presented and compared with theory with particular attention being paid to the scatter artefact and fan-beam energy profile. It is concluded that, irrespective of any quality assurance protocol, interpatient and interslice errors can be expected to range from 3 to 10% for water-equivalent materials and the intraslice positional dependence of the CT number can vary up to 5% for dense bone-like materials in a uniform phantom. It seems as though the principal cause of this variation is an increase in the effective x-ray energy in the fan beam with increasing distance from the central ray.

1. Introduction

Quantitative measurement of the local linear attenuation coefficient, or its correlates, has long been the goal in the development of computed tomography (CT). The marriage of computer technology with fundamental radiological physics has produced a digital representation with surprising accuracy and precision. However, non-linearities which occur in the projection values, such as beam hardening, partial volume, or scatter effects (Pullan *et al* 1981, Kijewski and Bjärngard 1978, Glover and Pelc 1980), continue to stand in the way of ideal quantitative measurement. In addition, many manufacturers prefer to avoid the technical difficulties involved in making accurate measurements and produce, instead, a machine more suited to the needs of the diagnostic radiologist than the medical physicist. The qualitative appearance of the final image, its clarity and absence of obvious artefact has become paramount over the quantitative accuracy, despite careful definition of the Hounsfield unit and considerable effort at formulating consistent universal quality assurance protocols by Brooks (1977) and White and Speller (1981). Careful scrutiny of the results can sometimes give insight into problems which exist in the imaging system and their origin. Corrective measures can then be undertaken. If these corrections are unsuccessful or impossible then bounds for existing errors along with the important influencing factors can be determined.

Despite the existing inaccuracies, quantitative CT has been used to advantage for the measurement of cortical bone mineral content (Revak 1980) as well as for iodine

† Present address: Department of Physiology and Biophysics, Dalhousie University, Halifax, Nova Scotia, Canada.

‡ Present address: Radiation Oncology Center, 5271 F Street, Sacramento, CA 95819, USA.

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wash-out studies (Fike *et al* 1982). In these cases, however, the large range of CT numbers helps minimise the effect of errors. More subtle measurements which could be of significant diagnostic value have been obscured by non-linearities, and the great promise of quantitative CT has largely been lost.

The errors caused by Compton-scattered photons, and the error caused by a variation of effective beam energy at different locations in the field of reconstruction are of special interest in this study. Scatter error has been discussed by Johns and Yaffe (1982) and Moran *et al* (1983). This is caused by inadequate collimation at the detector and results in an additional contribution to the measured signal from Compton-scattered photons originating in the irradiated subject. The error induced in the reconstructed image is proportional to the scatter to primary ratio and is most severe for highly attenuating materials. For large uniform subjects the scatter contribution is nearly uniform at all detectors and a first-order correction (subtraction of a constant) is often made by the data acquisition software. Large subjects will, however, have reduced primary transmission and a corresponding increased scatter error. Failure to correct for this scatter error can result in severe 'cupping'. Scanners most prone to this type of error are of the third- and fourth-generation rotate-only design. In the case of the fourth-generation stationary detector ring each detector is required to accept radiation from any chord of the reconstruction circle; consequently it cannot be collimated against in-slice scatter. A water-equivalent plastic phantom scan is used for a calibration on the EMI 7070 fourth-generation scanner to produce a calibration reference file for future scans. These files, along with software corrections, compensate for scatter when the subject is identical in scatter properties to the calibration phantom. The more the subject differs from the calibration phantom, however, the less adequate the scatter correction will be. It may even pose a source of error if the subject consists of a great deal of air and consequently gives rise to fewer scattered photons than expected, based on its outside dimensions which dictate the size of the circle of reconstruction and the calibration phantom size. This can occur, for example, in thoracic scans. As a simple test which demonstrates this effect, a 32 cm uniform water-equivalent phantom was placed off-centre in a 40 cm circle of reconstruction and was observed to have an average CT value as high as 70 HU in some locations. We suspect that this deviation from the normal zero value is caused by a lack of scatter relative to calibration conditions.

An additional source of error is the presence of the aluminium bow-tie filter on non-water-bath machines. The higher atomic number of aluminium causes hardening of the x-ray beam relative to that occurring in an equivalent path length through water. For peripheral rays in the fan beam the longer path length through the aluminium compensating filter hardens the beam relative to the central ray. This effect is not too severe, but, as will be shown in this paper, it is measurable and can cause CT number changes with position, especially for photoelectric absorbers such as bone.

2. Experimental details

Measurements were made on a CT disc phantom (RMI model 462 turned down to a diameter of 32 cm) into which inserts containing a variety of materials could be introduced at radial locations 0, 6 and 12 cm from the phantom centre (figure 1). The insert materials and their properties are summarised in table 1. Reagent-grade $CaCl_2$ and distilled water were used to make up the $CaCl_2$ solutions. Iodinated contrast material (Renografin-30, Squibb Canada, Inc) was diluted with distilled water to

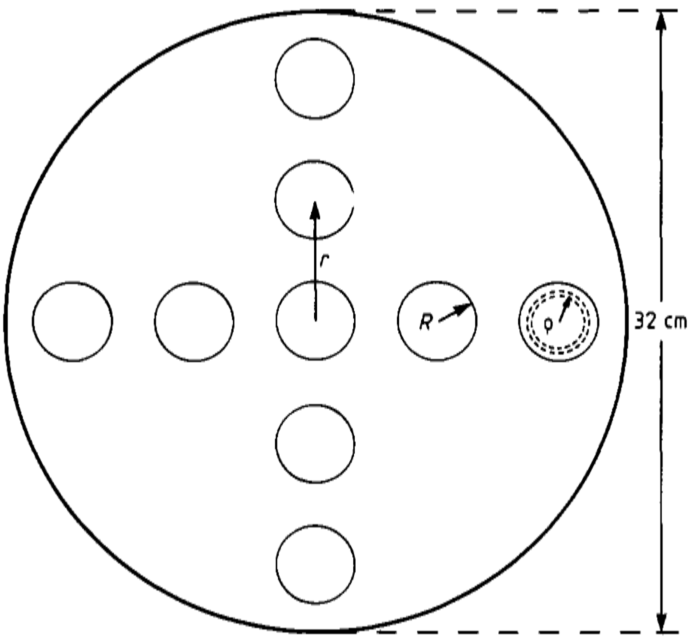

Figure 1. Geometry of solid water-equivalent phantom consisting of epoxy resin-base material WT/SR 6 (80.48% epoxy CB 4, 10% polyethylene, 5.77% $CaCO_3$, 3.75% phenolic microspheres) containing insert locations at three possible distances r from the phantom centre of 0, 6 and 12 cm. Inserts having diameters up to 2.54 cm were inserted into carrier plugs of phantom material in order to locate them on a common centre at each of the three positions. With the aid of the scanner statistical program average CT numbers were obtained for annular regions at radius ρ (variable) within each insert.

contain approximately 6% by weight of iodine to make up a solution which produced CT numbers in the 900 range. The Teflon used was of engineering quality (Crystaplex Plastic Ltd) and had a density of 2.15 g cm^{-3}. The liquids were retained in polyethylene walled syringes while special inserts were machined from the Teflon.

Measurements were made on an EMI 7070 scanner installed at the Montreal General Hospital. All measurements were made under identical operating conditions of 80 mA and 120 kV_p in a three-second scan. The data were collected with the aid of the statistical analysis routine supplied with the CT unit.

Table 1. Properties of materials used as phantom inhomogeneity inserts.

Material	Concentration (g/100 g)	Density (g cm^{-3})	Z_{eff}	Approximate CT number (HU)	Remarks
Air		0.000 12	7.8	−1000	Produces no scattered photons
Water		1.00	7.4	0	Calibration material
$CaCl_2$	0–10	1.0–1.1	7.4–8	0	Low CT number but strong photoelectric effect
	55	1.61	15.8	1000	Strong photoelectric effect as in bone
Iodinated contrast	6	1.00	23.9	1000	Very strong photoelectric effect
Teflon		2.16	8.4	900	Photoelectric effect similar to water

The phantom was centred in the 32 cm diameter circle of reconstruction and carefully aligned before each data session using phantom alignment inserts provided with it. CT numbers of insert materials were measured and the data were analysed to provide averages within annular regions. Next, these averages were plotted as in figure 2. Finally the value at the centre of the insert was obtained by regression to a fourth-degree polynomial.

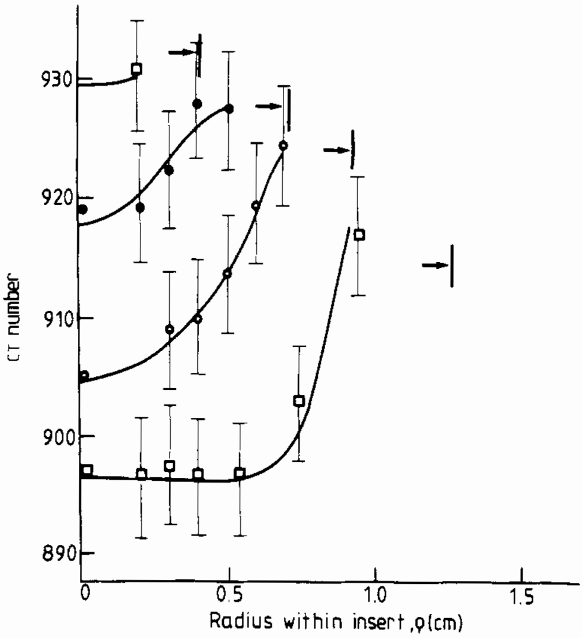

Figure 2. Data for four inserts of radius R, 0.4, 0.7, 0.92 and 1.27 cm containing $CaCl_2$ solution of concentration 55 g/100 g and located at the centre of the phantom ($r = 0$). The average raw CT number within the annular regions is plotted against the radius of the region from the insert centre. Error bars indicate the standard deviation of CT numbers within the annulus. Full curves are the best fourth-degree polynomial fit to the experimental data. The arrows indicate the location of the physical edge of each insert.

3. Results

Measurements were made for a set of inserts having a variety of radii in order to obtain the extrapolated zero-area value. This zero-area CT number is the limit of small perturbation from a uniform phantom and should be free of the beam hardening artefact and other non-linear errors in the projection data. It is chosen as the artefact free (*sic*) CT number in many quality assurance protocols and to produce corrected CT values in Hounsfield units it is scaled to the ideal Hounsfield scale for which the values for water and air are $H_w = 0$ and $H_a = -1000$. Figure 3 shows the centre values for the four cross sections of figure 2 plotted against the insert radius along with corresponding data obtained for the iodine solution and Teflon at radial locations of 0, 6 and 12 cm in the phantom. The slopes and intercepts of the best straight-line fits to the data shown in figure 3 are summarised in table 2 along with the corresponding values for air. As can be seen from the figure and from the table, the intercept values corresponding to inserts of zero radius vary as a function of location in the phantom. If these values were truly artefact-free no change with location should have been observed; a change is observed, however, and its magnitude increases with the effective atomic number of the insert material. This finding prompted us to investigate possible changes in the scanner effective energy corresponding to different radial locations within the circle of reconstruction.

The effective energy of the emergent photon spectrum can be derived with the aid of an iterative technique proposed by Brooks *et al* (1980) in which one calculates a

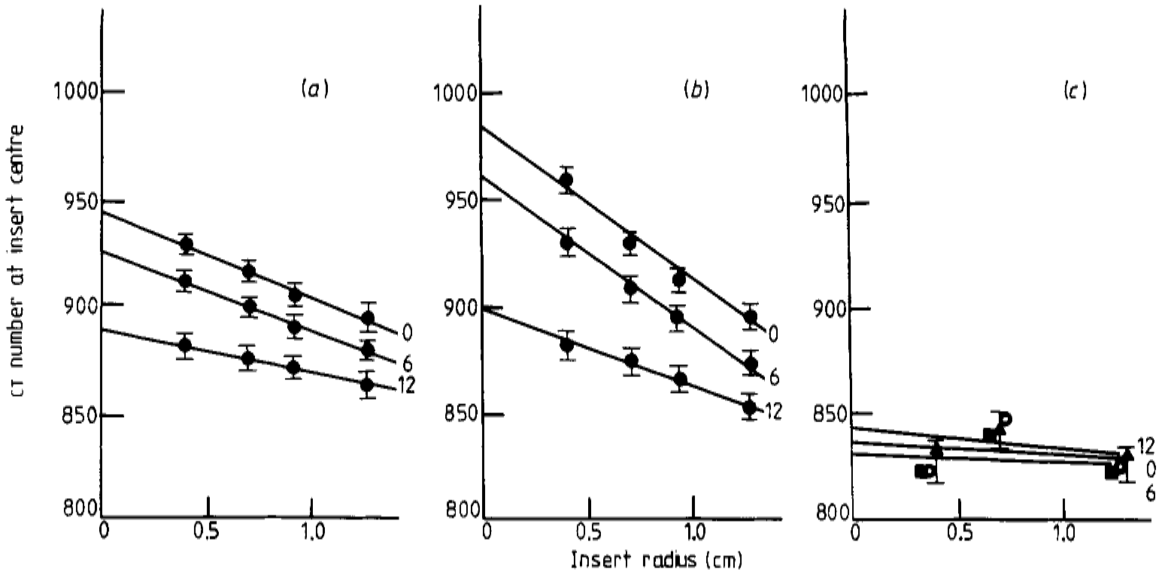

Figure 3. Variation of CT number with insert radius, material, and location, r (given on the curves in cm), from the phantom centre. The CT number at the centre of each insert is plotted against the insert radius for (a) 55 g/100 g solutions of $CaCl_2$, (b) 6 g/100 g solution of iodine, and (c) Teflon, for which open cricles refer to $r = 0$, full squares refer to $r = 6$ cm and full triangles refer to $r = 12$ cm. Error bars are the standard deviations of the fourth-degree polynomial regression data.

spectral factor β, the ratio of the photoelectric to Compton attenuation of water for the emergent photon spectrum. The effective energy is the monochromatic energy with the same value of β. β can be obtained from measurements of the CT numbers of weak solutions of a highly photoelectric attenuating substance. In our experiment the dilute solutions of $CaCl_2$ (table 1) were used to verify that the system response was linear and then a $CaCl_2$ solution of concentration 3 g/100 g was placed at the centre and at 6 and 12 cm off-centre to obtain the value of β from which the effective energy was deduced.

The results of this exercise are summarised in table 3. The effective energy increases with increasing distance from the centre of the circle of reconstruction and this trend is consistent with the direction of change in Hounsfield number observed for the concentrated $CaCl_2$ and iodine solutions. As the effective energy increases the attenuation experienced in these materials decreases owing to the $1/E^3$ dependence of photoelectric absorption. This leads us to expect a decrease in the CT number which is indeed observed.

Table 2. Sumary of insert data.

	Straight-line slope (CT number/cm)			Zero-radius intercept (CT number)		
Material	Centre	6 cm from centre	12 cm from centre	Centre	6 cm from centre	12 cm from centre
Teflon	-5 ± 5	-6 ± 5	-12 ± 5	836 ± 5	832 ± 5	844 ± 5
$CaCl_2$ (55 g/100 g)	-38 ± 5	-37 ± 5	-18 ± 7	945 ± 5	928 ± 5	889 ± 10
I	-72 ± 5	-70 ± 5	-32 ± 7	983 ± 5	960 ± 5	894 ± 10
Air	-31 ± 5	-25 ± 5	-22 ± 5	-912 ± 5	-920 ± 5	-938 ± 10

Table 3. Effective photon energy determination. CT numbers for water (H_w), air (H_a), and 3 g/100 g $CaCl_2$ solution (H_m), used to calculate effective photon energy (E_{eff}), at each possible insert location. Two sets of measurements were made at centre position.

	Centre		6 cm from centre	12 cm from centre
H_w	0.95	2.33	0.23	2.08
H_a	−916	−915	−923	−940
H_m	48.0	48.1	45.6	41.7
β	0.0844	0.0803	0.0783	0.0719
E_{eff} (keV)	72.0 ± 2		73.5 ± 2	76.0 ± 2.5

4. Discussion

The results presented above and previous work by Moran *et al* (1983) in this laboratory indicate that a combination of two processes is being observed: the change in effective energy across the x-ray fan beam and scatter reaching the detectors which is not adequately corrected for during data acquisition. Let us consider uncorrected scatter first. It has already been shown (Moran *et al* 1983) that the error due to uncorrected scatter depends upon the ratio of the scatter to the primary contributions to the measured intensity. This is expected to vary across the detector array exposed to the fan beam even without a change in scatter across that fan, simply because the primary detected intensity decreases with its obliquity of incidence at the detectors in the periphery. The obliquity effect alone will cause an increase in scatter error which will result in an apparent CT number of highly attenuating substances which is lower near the periphery of the phantom than at its centre. This effect will be balanced when the quantity of scatter detected near the centre of the fan is greater than that detected near the periphery. The data reported here for Teflon will be subject to this balance of scatter error, and its magnitude should be roughly the same for the iodine and the concentrated $CaCl_2$ solution as their CT numbers are similar. We see from table 2 and figure 3, however, that the change in CT number with position for Teflon is markedly less than the change observed for $CaCl_2$ and iodine. In fact, the net effect in Teflon is so small that it is obscured by experimental fluctuations which, we might add, we took great pains to minimise. It appears as though the magnitude of the balance of scatter error is insignificant compared with other sources of change with position which affect the iodine and $CaCl_2$ data. This is because scatter error should not depend upon the identity of the inhomogeneity but only upon the relative attenuation it produces, and since our inserts all produce the same range of CT numbers, one would expect no difference in the relative intercept values as a function of insert position for $CaCl_2$, iodine solution, or Teflon if scatter error were the only phenomenon being observed. We are therefore forced to look elsewhere for a plausible explanation of the observations shown in figure 3. In this study we have observed an increase in effective x-ray energy with radius in the circle of reconstruction. This increase would also be expected to produce a change in the CT number of a high atomic number inhomogeneity as observed in figures 3(a) and 3(b). To examine the relative magnitude of this effect, the expected CT number for the insert materials was calculated from a knowledge of their properties and the effective x-ray energy. In table 4, we compare this expected CT number with the observed zero insert radius data for iodine, $CaCl_2$ and Teflon at the 0 and 12 cm locations. It can be seen that the ratio of CT numbers which is predicted on the basis

Table 4. Comparison of calculated and observed CT numbers (HU). Observed data of table 2 using air and water values at position of interest corrected to the ideal Hounsfield scale: $H = (H_m - H_w)/(H_w - H_a)$.

Insert		Location		
		Centre	12 cm from centre	Ratio of CT numbers
Teflon	expected	1020	1008	0.99
	observed	909	897	0.99
$CaCl_2$	expected	1424	1310	0.91
	observed	1022	941	0.92
I	expected	1446	1275	0.88
	observed	1066	949	0.89

of the effective energy increase is very close to the ratio actually observed. No explanation for the difference between the expected and observed values can be offered at this juncture, however.

The explanation for the change in effective energy across the circle of reconstruction is simple. The aluminium bow-tie filter employed in the unit hardens the beam at the edges of the x-ray fan. This is because the path length through aluminium is greater for a specific ray located further from the centre. The actual average energy of such rays is considerably greater than what has been obtained in this study, however, because our study did not use a stationary beam. Our measurement of the effective beam energy has been made using the true conglomeration of rays which pass through the insert location during an actual scan. For inserts near the centre of the phantom ($r = 0$) this is always the central ray of the beam; however, for inserts near the phantom edge ($r = 12$ cm) rays from a large part of the fan supply the relevant analysing signal as the x-ray tube passes through that part of its orbit which allows rays from the defined fan beam to intersect the eccentric insert and register a signal at the appropriate detector. The insert is therefore analysed using rays having a range of effective energies and it is a complex average effective energy which we observe and which is clinically relevant. One might refer to this as the effective energy of analysis. What we have shown in this study is that this effective energy of analysis increases significantly with radius in the circle of reconstruction.

5. Conclusion

For practical purposes, software or hardware corrections to existing scanning systems are not feasible in a hospital setting without specific manufacturers' updates. The problem of scatter will be common to all fourth-generation scanners and has been observed as well on third-generation machines. While sophisticated scatter corrections are conceivable which account for position, size and shape of scattering material, no absolute correction seems possible. Nor, indeed, does it seem worthwhile since other more significant sources of error occur. The change of effective beam energy produced by the aluminium filter can induce changes in CT number of the order of 5 to 10% for some materials, notably strongly photoelectric absorbing ones such as bone, since the attenuation coefficient of calcium changes rapidly with photon energy in the range used. While this would be somewhat bulky, it may be possible to produce a bow-tie filter of similar effective atomic number to water from a dense plastic such as Delrin and thus eliminate the energy span. At least one manufacturer has done something similar to this. The bow-tie filter, after all, is a compensating filter which takes the place of the original water tank and is an attempt to equal out the path length in water.

The two errors mentioned above complicate the problems associated with photoelectric beam hardening corrections greatly since the presence of a range of energies, an additional contribution to the projection values, means that the three errors are inseparable. Most photoelectric beam hardening corrections rely on a path length calibration specific to the spectrum being used. If other systematic errors exist this path length calibration, usually taken under narrow-beam conditions, no longer strictly applies and contributions to the projection values cannot be sorted out. In the case of scatter a first-order correction is afforded by subtracting a constant scatter component from all projection values, but precise projection values are impossible to obtain and some position dependence of CT number will exist.

Finally, we would like to point out that the quantitative difficulties of these machines in no way disqualifies them from clinical usefulness. Third- and fourth-generation scanners are preferred in hospitals because their rotate-only design permits rapid three-second scans, and generally means less maintenance. Images are free from motion artefact and spatial resolution is usually good. With appropriate smoothing, clinical detail is usually more than adequate, and to the clinician this is the telltale mark of a superior machine.

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Résumé

Influence du rayonnement diffusé et des caractéristiques du filtre sur les études quantitatives pratiquées avec le scanneur EMI 7070.

Un défaut de linéarité dans les projections introduit des erreurs dans les mesures d'atténuation pratiquées à l'aide de scanneurs et à partir des nombres Hounsfield. Les appareils disponibles sur le marché ont été adaptés à l'utilisation clinique, mais pas nécessairement aux déterminations quantitatives précises. Dans ce travail, les résultats des mesures pratiquées sur le scanneur de quatrième génération EMI 7070 sont présentés et comparés aux données théoriques. Dans cette comparaison, on a porté une attention particulière aux artefacts dus au diffusé et aux variations d'énergie dans le faisceau. La conclusion des auteurs est que, quel que soit le protocole de contrôle de qualité utilisé, des erreurs peuvent être observées d'un malade à un autre ou d'une coupe à une autre. Les erreurs sur les nombres Hounsfield peuvent dans ce cas se situer dans l'intervalle 3 à 10% pour des matériaux équivalent-eau et celles dues à la localisation dans une coupe peuvent atteindre jusqu'à 5% pour des matériaux tels que l'os dense placé dans un fantôme uniforme. Tout se passe comme si la principale cause de cette variation était un accroissement de l'énergie effective des RX quand la distance par rapport au rayon central augmente.

Zusammenfassung

Quantitative CT-Messungen: der Einfluß von Streuakzeptanz und Filtereigenschaften auf das EMI 7070-System.

Die Nicht-Linearitäten von Projektionswerten bei computertomographischen (CT) Scannern sind der Grund für entsprechende Fehler bei Messungen der Schwächung in Hounsfield-Einheiten. Bestehende kommerzielle

Maschinen wurden zwar weiterentwickelt je nach klinischen Erfordernissen, aber nicht unbedingt in Richtung quantitativer Genauigkeit. In der vorliegenden Arbeit wird über Messungen an einem EMI 7070-Scanner der 4.Generation berichtet. Die Ergebnisse werden verglichen mit der Theorie, unter besonderer Berücksichtigung der Artefakte durch Streuung und der Energie-Profile des Fächerstrahls. Zusammenfassend kann gesagt werden, daβ man, unabhängig vom Qualitätssicherungs-Protokoll, zwischen einzelnen Patienten und zwischen verschiedenen Schnitten Fehler im Bereich von 3-10% für wasser-äquivalente Materialien erwarten kann. Die Lageabhängigkeit der CT-Zahl innerhalb eines Schnittes kann bis zu 5% schwanken für dichte knochenähnliche Materialien in einem gleichförmigen Phantom. Die Hauptursache dieser Schwankungen scheint eine Erhöhung der effektiven Energie der Röntgenstrahlung im Fächerstrahl bei wachsender Entfernung vom Zentralstrahl zu sein.

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