A method for calculating the dose length product from CT DICOM images

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Objective: The dosimetric calculations in CT examinations are currently based on two quantities: the volume weighted CT dose index (CTDI_{vol}) and the dose-length product (DLP). The first quantity is dependent on the exposure factors, scan field of view, collimation and pitch factor selections, whereas the second is additionally dependent on the scan length.

Methods: In this study a method for the calculation of these quantities from digital imaging and communication in medicine (DICOM) CT images is presented that allows an objective audit of patient doses. This method was based on software that has been developed to enable the automatic extraction of the DICOM header information of each image (relating to the parameters that affect the aforementioned quantities) into a spreadsheet with embedded functions for calculating the contribution of each image to the CTDI_{vol} and DLP values. The applicability and accuracy of this method was investigated using data from actual examinations carried out in three different multislice CT scanners. These examinations have been performed with the automatic exposure control systems activated, and therefore the tube current and tube loading values varied during the scans.

Results: The calculated DLP values were in good agreement ($\pm 5\%$) with the displayed values. The calculated average CTDI_{vol} values were in similar agreement with the displayed CTDI_{vol} values but only for two of the three scanners. In the other scanner the displayed CTDI_{vol} values were found to be overestimated by about 25%. As an additional application of this method the differences among the tube modulation techniques used by the three CT scanners were investigated.

Conclusion: This method is a useful tool for radiation dose surveys.

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CT has greatly evolved since it was first introduced into clinical practice. The technological innovations in CT scanner hardware and software have led to the introduction of many new clinical applications of CT in diagnosis and therapy (CT-guided interventions), making CT the examination technique with the largest contribution to the cumulative patient doses from radiographic examinations [1].

From the time of its establishment, CT was considered as a relatively high-dose imaging technique [2]. For this reason, the calculation of patient dose has been for many years a subject of research. Special dosimetric quantities (see Appendix A), protocols and dedicated software have been designed for this purpose, and many scientific papers concerning patient doses from CT examinations have been published in the international literature. Currently the quantity that is being used as an indicator of patient dose from CT examinations is the dose-length product (DLP). The effective dose can be calculated from the DLP using conversion coefficients that have been proposed for specific routine examinations or by using special software [2, 3].

Modern multislice CT (MSCT) scanners offer a direct display of the volume-weighted CT dose index (CTDI_{vol})

and the DLP [4]. In older CT scanners where no direct display was available, these quantities could only be calculated retrospectively provided that the technical parameters of the examination (such as the tube potential, tube current, exposure time, scan field of view, pitch and scan length) were known. In examinations where the acquisition parameters were constant throughout the scan, the technical parameters could be derived from any one of the images and the scanned length could be deduced from the table position indication of the first and last image provided that a hard or a soft copy of the examination was available. The same procedure is also applicable in the case of MSCT scanners and can be used to verify the accuracy of the DLP values displayed. However, in the case of helical scanning it must also be taken into account that because of the overscan the scanned length is always larger than that planned [1, 2, 5]. Owing to the overscan, the displayed DLP values should always be greater than those calculated using the planned scan length.

However, for CT examinations performed using automatic exposure control systems, the retrospective calculation of DLP is difficult and time-consuming, since the tube current continuously changes during the scan [1, 2, 4, 6]. The same applies for interventional CT-guided procedures, where a quite large number of series are usually acquired at different anatomical positions and with different technical parameters each time, making the calculation of DLP a very cumbersome task [7].

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In this study an automated method for the calculation of the DLP from DICOM CT images is presented, whose purpose is to allow the objective audit of patient CT doses (in terms of DLP) using information from the DICOM header. This method was based on software that was developed to enable the automatic extraction of the DICOM header information of each image into a spreadsheet containing calculation algorithms. The applicability and accuracy of this method was investigated using data from actual examinations carried out in three different MSCT scanners.

Methods and materials

This study was carried out in three CT facilities, equipped with three different MSCT scanners, a Brilliance 6 (Philips Medical Systems, Cleveland, OH), an Asteion 4 (Toshiba Medical Systems, Otawara, Japan) and a CT/e Dual Plus (GE Healthcare, Milwaukee, WI); henceforth referred to as scanners A, B and C, respectively. In all scanners the $CTDI_{vol}$ and DLP values were displayed on the CT graphical user interface immediately after setting the technical parameters of the scan and prior to scanning.

All scanners were equipped with automatic exposure control (AEC) systems with which the X-ray tube current was modulated during scanning in helical mode, utilising the information obtained from the scout scans concerning the attenuation characteristics of the anatomy that was scanned [4, 6]. The AEC systems of scanners A, B and C are named respectively DoseRight-DOM, Real EC and Auto mA. With the DoseRight-DOM the user can select only the highest mAs value allowed. With the Real EC and Auto mA, the user can select from a number of preset options the tube modulation mode that will be used. Depending on this selection, high mA values can be used in order to minimise the image noise, low mA values can be used to minimise the patient dose at the expense of increased image noise or mA values that offer a compromise between image noise and patient dose can be used.

In each one of these scanners, a number of axial scans were performed with the CT table empty (no phantom or patient was used) using all the scan field of views (SFOV) and collimation selections available, with a tube potential of 120 kV and a constant tube loading of 100 mAs. In this way the $CTDI_{vol}$ indications displayed were actually equal to the normalised weighted CTDI values ($nCTDI_w$). These values were recorded and a table of $nCTDI_w$ values for the available options of SFOV and collimation was constructed for each scanner, henceforth referred to as the $nCTDI_w$ (SFOV, collimation) table.

As was mentioned above, in the case of helical scans, owing to the overscan, the DLP values displayed should always be larger than the product of $CTDI_{vol}$ and the planned scan length. The contribution of the overscan to the displayed DLP value in each scanner was obtained using the following methodology. A number of helical scans were performed (with the CT table empty as before) using two of the most common examination protocols employed in clinical practice, that is the chest and the abdomen/pelvis examination protocols, manual mAs selection (the AEC systems were deactivated) and different scan lengths ranging from the minimum allowed up to a scan length of about 40 cm. For each scan, the

displayed $CTDI_{vol}$ and DLP values and the technical parameters of each protocol were recorded. The displayed DLP values were plotted against the scan lengths and the intercept of the best fit lines, that is the DLP for scan length equal to zero ($DLP_{(0)}$), was attributed to the overscan. All the recorded values for the $CTDI_{vol}$, DLP and $DLP_{(0)}$ were normalised per 100 mAs to derive the $nCTDI_{vol}$, $nDLP$ and $nDLP_{(0)}$ values, respectively. The parameters of the chest, abdomen and pelvis CT examination protocols are shown in Table 1 and the results of the procedure described above are summarised in Figure 1.

For each scanner, DICOM images of the axial and helical scans described above were stored on a CD-ROM in DICOM format in order to examine the information displayed on the DICOM headers. For this purpose the respective CT manufacturers' DICOM viewers, which are burnt on the CD-ROM along with the images, were used, as well as the eFilm Workstation 3.0 (Merge Healthcare, Merge Technologies Incorporated, WI). Using the DICOM viewers, the tags (data elements) containing information relevant to the technical parameters of each image were identified.

A software program working in the Windows environment (Windows 2000 or later) was developed using C++ (C++ Builder 2009, Embarcadero Technologies, CA), which enables the automatic extraction of the information contained within the DICOM headers of each image. When starting this program (henceforth referred to as Dicom Info Extractor) for the first time, the set-up option has to be used in order to specify the technical parameters of interest by typing in an arbitrary name for the description of each parameter and the corresponding tags' addresses. After the set-up list has been saved, the user can then locate the directory containing the DICOM images of a specific CT examination, extract the DICOM header data and save them in a Microsoft Office Excel spreadsheet format. The program was customised for each scanner and the technical parameters defined for each one along with the respective tags' addresses are given in Table 2.

Figure 1. The results of the dose-length product (DLP) indications (normalised per 100 mAs) vs planned scan length for the three multislice CT (MSCT) scanners are given. Apart from data points the best fit lines (solid and dashed lines) and the respective equations are also given.

Table 1. The technical parameters of the routine chest, abdomen and pelvis examination protocols are presented

Scanner—Protocol	Scanning mode	FOV	Collimation	Slice width (mm)	Reconstruction interval (mm)	Pitch factor	Pre-set mAs	Pre-set CTDI _{vol}	nCTDI _{vol} (nCTDI _w) (mGy)	nDLP ₍₀₎ (mGy)
A—Chest	Helical	Body	4 × 6	7.5	3.7	1.2	240	13.3	5.54 (6.65)	34.6
A—Abdomen, pelvis	Helical	Body	4 × 6	7.5	3.7	0.9	180	13.3	7.39 (6.65)	44.4
B—Chest	Helical	LL	4 × 3	5	5	1.25	150	13.0	8.67 (10.8)	30.3
B—Abdomen, pelvis	Helical	LL	4 × 3	5	5	1.25	187.5	16.3	8.67 (10.8)	30.3
C—Chest, abdomen	Helical	Large	2 × 7	7	7	1.50	160	8.3	5.19 (7.63)	14.1
C—Pelvis	Helical	Large	2 × 7	7	7	1.50	200	10.4	5.19 (7.63)	14.1

The tube potential in all these examination protocols was 120 kV. The nCTDI_{vol} values of these protocols are compared with the respective nCTDI_w values (in parentheses). The contributions of the overscan with each examination protocol are also given in the last column (normalised per 100 mAs). FOV, field of view; CTDI, CT dose index; DLP, dose-length product.

The software for calculating DLP from the DICOM image data was developed using Microsoft Office Excel spreadsheets and embedded functions. The calculation method was based on the assumption that the DLP can be considered as the sum of the contributions of the DLP fractions that correspond to each one of the reconstructed images plus the amount of DLP that is due to the overscan and is analysed in the following paragraphs:

The nCTDI_{vol} of any type of scan can be derived from the nCTDI_w using the following equation:

$$nCTDI_{vol} = \frac{nCTDI_w}{pitch} \quad (1)$$

Since the nCTDI_w is dependent on the SFOV and collimation selection, the program uses the relevant information from the DICOM data of each image to automatically pick the appropriate value from the nCTDI_w (SFOV, collimation) table that will be used in

Equation 1. While in this study the nCTDI_w (SFOV, collimation) tables were constructed using the displayed values in the axial scans performed for this purpose, relevant data for a number of other scanners can be found in the Impact group website [3]. However, measured data could be used as well.

The contribution of each reconstructed image to the DLP (normalised per 100 mAs) was termed nDLP_{img} and can be calculated from the following equation:

$$nDLP_{img} = nCTDI_{vol} \times \frac{|\text{reconstruction interval}|}{10} \quad (2)$$

The reconstruction interval values are given in mm and were divided by 10 in order to calculate nDLP_{img} values in units of Gy.cm as usual. The absolute value was taken because depending on the scan direction the reconstruction interval may be a positive or a negative number. The reconstruction interval was used instead of

Table 2. The addresses of the tags relevant to the identification of the examination and the technical parameters required for the calculation of dose-length product (DLP) values are given for each one of the scanners included in this study

Parameter/scanner	A: Brilliance 6 (Philips)	B: Asteion 4 (Toshiba)	C: CT/e Dual Plus (GE)
Study ID	0 × 0020, 0 × 0010	0 × 0020, 0 × 0010	0 × 0020, 0 × 0010
Study date	0 × 0008, 0 × 0020	0 × 0008, 0 × 0020	0 × 0008, 0 × 0020
Series number	0 × 0020, 0 × 0011	0 × 0020, 0 × 0011	0 × 0020, 0 × 0011
Scan mode (axial, spiral)	0 × 01F1, 0 × 1001	0 × 0018, 0 × 0022	0 × 0018, 0 × 0022
Image number	0 × 0020, 0 × 0013	0 × 0020, 0 × 0013	0 × 0020, 0 × 0013
Slice location	0 × 0020, 0 × 1041	0 × 0020, 0 × 1041	0 × 0020, 0 × 1041
Scan FOV diameter	0 × 0018, 0 × 0090	0 × 0018, 0 × 0090	0 × 0018, 0 × 0090
Collimation	0 × 01F1, 0 × 104B	0 × 7005, 0 × 1008	0 × 004B, 0 × 1001
Image slice thickness	0 × 0018, 0 × 0050	0 × 0018, 0 × 0050	0 × 0018, 0 × 0050
Reconstruction interval	0 × 0018, 0 × 0088	0 × 7005, 0 × 1022	— ^d
Pitch ^a	0 × 01F1, 0 × 1026	0 × 7005, 0 × 1023	0 × 0043, 0 × 1027
Tube potential (kV)	0 × 0018, 0 × 0060	0 × 0018, 0 × 0060	0 × 0018, 0 × 0060
Tube current (mA)	0 × 0018, 0 × 1151 ^b	0 × 0018, 0 × 1151	0 × 0018, 0 × 1151
Exposure time (ms)	0 × 01F1, 0 × 1027^c	0 × 0018, 0 × 1150	0 × 0018, 0 × 1150
Exposure (mAs)	0 × 0018, 0 × 1152	0 × 0018, 0 × 1152	— ^e

The tags that are not the same for all scanners are given in bold. ID, identification; FOV, field of view. ^aFor all scanners this tag does not appear when the image is acquired in axial mode. ^bFor this scanner the mA values remain constant when automatic exposure control (AEC) is used despite the fact that exposure (mAs) values are different. Therefore, they do not represent the actual mA but rather the maximum mA value with deactivated AEC, for the pitch value, exposure time and maximum mAs value selected for the acquisition. ^cFor this scanner the exposure time in helical mode is given in units of sec. In axial mode the exposure time appears at the tag 0 × 0018, 0 × 1150 as in the other two CT scanners, in units of ms. ^dThe reconstruction interval was not found within the digital imaging and communication in medicine (DICOM) header information. It was derived from the difference of the slice location values of successive images. For all exams included in the study it was equal with the slice thickness. ^eThe exposure (mAs) was not found within the DICOM header information. It was derived from the product of the mA value and the exposure time value (expressed in seconds).

the reconstructed slice width, because this is what actually determines the number of images that correspond to a given scan length.

Therefore, for each scan series the DLP can be calculated from the sum of the contributions of all images plus the contribution of the overscan, that is:

$$DLP_c = \sum_{i=1}^k \left[({}_nDLP_{img})_i \times \frac{mAs_i}{100} \right] + {}_nDLP_{(0)} \times \frac{\text{average mAs} \left(\begin{array}{c} \text{first and last image} \\ \text{in each scan series} \end{array} \right)}{100} \quad (3)$$

In the above equation DLP_c stands for the calculated DLP, ${}_nDLP_{img}$ stands for the contribution (normalised per 100 mAs) of each one of the “k” images of the scan series to the DLP_c and mAs_i stands for the mAs value in the DICOM header of i th image. In the last term of the equation it can be seen that the contribution of overscan is calculated by multiplying the ${}_nDLP_{(0)}$ by the average mAs of the first and last slices of each helical scan series, since when AEC is used these will be usually different. For the purposes of this study, a single value of ${}_nDLP_{(0)}$ was used for each scanner. For scanners B and C the ${}_nDLP_{(0)}$ values used are shown in the last column of Table 1, while for scanner A the average of the two ${}_nDLP_{(0)}$ values shown in Table 1 was used. However, it must be stressed that the amount of overscan is dependent on the pitch and collimation selection (as can be seen in Table 1 for scanner A) and therefore for better accuracy in the calculation of DLP from the DICOM data, an ${}_nDLP_{(0)}(\text{pitch, collimation})$ table should be constructed as it was done for the ${}_nCTDI_w$.

The average $CTDI_{vol}$ of each scan series was calculated by the following equation:

$${}_{av}CTDI_{vol} = \text{Mean}_{i=1}^k \left[({}_nDLP_{img})_i \times \frac{mAs_i}{100} \times \frac{10}{|\text{reconstruction interval}|} \right] \quad (4)$$

It is straightforward that if more than one scan series is performed during an examination, the total DLP_c will be the sum of the DLP_c values of all scan series.

The three versions of this program (one for each scanner) were used to extract the DICOM data of the experimental scans carried out with constant mAs and the DICOM data from actual examinations carried out in these CT scanners using the chest and abdomen/pelvis examination protocols and the AEC option activated. Concerning the CT examinations in actual patients, the $CTDI_{vol}$ and DLP indications displayed after the end of the scans had been manually recorded for comparison with the calculated results.

Results

Validation of the DICOM-based calculation method

In Table 3 the DLP_c and ${}_{av}CTDI_{vol}$ values calculated from the DICOM data are shown for a sample of 10 randomly selected examinations carried out on each scanner (with the AEC systems activated) in comparison with the displayed values of DLP and $CTDI_{vol}$. Concerning the DLP_c it can be seen that the DICOM data-based calculation method produces results which are in good agreement (within $\pm 5\%$) with the displayed DLP values. For scanners A and C the same was also valid for the ${}_{av}CTDI_{vol}$ and $CTDI_{vol}$ values, whereas for scanner B the ${}_{av}CTDI_{vol}$ was on average $74 \pm 6\%$ of the displayed $CTDI_{vol}$ values.

The deviation of the ${}_{av}CTDI_{vol}$ from the displayed $CTDI_{vol}$ value observed for scanner B was attributed to the overestimation of the $CTDI_{vol}$ by the CT scanner software. Indeed, for scanner B the displayed $CTDI_{vol}$ multiplied by the planned scan length produced a DLP value that was 5–41% larger than the displayed DLP (even without accounting for the overscan contribution), indicating clearly that the displayed $CTDI_{vol}$ was overestimated.

It must be noted that in scanner A the tube current automatically increases in proportion to the increase in pitch factor (in order to maintain a constant noise level) and for this reason the $CTDI_{vol}$ is equal to $CTDI_w$ for all pitch selections [1]. Therefore, when selecting a tube loading of 200 mAs, this corresponds to axial scanning with a tube current of 267 mA (assuming that the default exposure time of 0.75 s is used). When selecting a pitch factor of 0.9 or 1.2 the respective tube loadings (with the AEC deactivated) automatically become 180 mAs and 240 mAs and the respective tube current values 240 mA and 320 mA. It must be noted, however, that these tube current and tube loading values could be seen only in the DICOM header information. In the CT monitor the tube loading value shown was always 200 mAs.

Using the method for testing the behaviour of the AEC systems

In Figures 2, 3 and 4, typical examples of the tube loading (mAs) variations with respect to the scanned anatomy, are shown for scanners A, B and C, respectively. It must be clarified that the mAs variations shown in these figures are due to the tube current (mA) variations since the exposure time per rotation remained constant. The slice location was matched to the patient anatomy shown on the scout image using the reference line tools available in eFilm Workstation. The mAs variation pattern observed in scanner A was quite different from that exhibited by scanners B and C. In scanners B and C the mAs values were greatly reduced when moving from the shoulders to the lungs area and started to rise at the interface of the heart and the abdominal region and they usually retained high values up to the end of the pelvis. This variation pattern is indeed quite similar to that exhibited in a similar figure included in McCollough et al [4]. On the other hand, in scanner A the mAs was not significantly reduced in the lung area and consequently the rise at the

Table 3. Comparison of displayed and calculated CTDI_{vol} and dose-length product (DLP) values for a sample of 10 exams in each scanner sorted according to the DLP value. CT dose index (CTDI) values are in mGy and DLP values in mGy cm

Scanner no.	Examination	Displayed values		Calculated values		DLP _c /DLP	avCTDI _{vol} /CTDI _{vol}
		DLP	CTDI _{vol}	DLP _c	avCTDI _{vol}		
A—1	Chest, abdomen and pelvis	799.2	11.30	764.5	11.04	0.96	0.98
A—2	Chest, abdomen and pelvis	775.0	11.10	745.4	10.80	0.96	0.97
A—3	Chest, abdomen and pelvis	769.4	11.40	740.3	11.18	0.96	0.98
A—4	Chest, abdomen and pelvis	759.5	11.30	730.0	11.06	0.96	0.98
A—5	Chest and abdomen	580.8	11.20	563.9	10.95	0.97	0.98
A—6	Chest and abdomen	520.8	11.50	501.2	11.20	0.96	0.97
A—7	Abdomen and pelvis	507.1	11.20	491.1	10.70	0.97	0.96
A—8	Chest	443.5	11.60	428.5	11.22	0.97	0.97
A—9	Chest	396.9	10.70	376.6	10.29	0.95	0.96
A—10	Chest	394.8	11.90	375.5	11.59	0.95	0.97
B—1	Chest, abdomen and pelvis	981.5	20.20	992.4	16.05	1.01	0.79
B—2	Chest, abdomen and pelvis	922.3	19.50	932.5	14.48	1.01	0.74
B—3	Chest	545.6	19.50	556.2	13.69	1.02	0.70
B—4	Chest, abdomen and pelvis	310.3	7.80	314.9	5.59	1.01	0.72
B—5	Chest	234.7	12.30	245.5	7.48	1.05	0.61
B—6	Chest	231.8	11.00	236.9	8.12	1.02	0.74
B—7	Abdomen and pelvis	228.6	7.80	232.5	6.43	1.02	0.82
B—8	Chest	208.5	9.10	219.6	6.91	1.05	0.76
B—9	Abdomen and pelvis	151.0	5.20	151.5	3.75	1.00	0.72
B—10	Pelvis	83.9	5.20	82.4	4.00	0.98	0.77
C—1	Abdomen and pelvis	376.2	8.14	371.0	8.30	0.99	1.02
C—2	Abdomen and pelvis	324.9	8.14	330.6	8.30	1.02	1.02
C—3	Abdomen and pelvis	323.2	7.33	327.8	7.69	1.01	1.05
C—4	Abdomen and pelvis	290.7	8.14	301.6	8.30	1.04	1.02
C—5	Abdomen and pelvis	284.3	9.67	281.1	9.89	0.99	1.02
C—6	Chest	222.3	8.14	226.0	8.30	1.02	1.02
C—7	Chest	191.7	7.02	194.9	7.03	1.02	1.00
C—8	Chest	168.1	6.16	170.2	6.06	1.01	0.98
C—9	Chest	153.9	8.14	156.3	8.30	1.02	1.02
C—10	Abdomen and pelvis	131.8	4.88	134.5	4.86	1.02	1.00

abdominal region was not as prominent, while additionally a slight decrease was always observed around the end of the pelvis region. As a result of the similar mAs variation patterns observed among all patients examined on scanner A and the fact that the protocols' default tube loading value of 200 mAs was always used, the displayed CTDI_{vol} values shown in Table 3 do not present any significant differences among different patients. The minimum and maximum CTDI_{vol} values recorded for scanner A differ by only 11%, while for scanners B and C the maximum to minimum CTDI_{vol} value ratios were 3.9 and 2.0, respectively. The small CTDI_{vol} variations among patients observed in scanner A may be because in this specific CT scanner only the DoseRight-DOM option was available. The DoseRight ACS option, which also utilises information from the scout scan to obtain mA modulation along the Z-axis, was completely deactivated (unavailable). Detailed information about these two AEC modules can be found in Lee et al [6].

In contrast to scanner A, in scanners B and C the CTDI_{vol} exhibited significant differences among patients but the mAs variation pattern was similar. In Figure 3a,b the mAs variation observed in two chest, abdomen and pelvis examinations performed in scanner B is shown. It is obvious that the examination of Figure 3a has been performed with a low-dose AEC mode whereas the examination of Figure 3b has been performed with a high-dose AEC mode in a patient having a lateral diameter in the waist area of about 10 cm larger than the patient shown in Figure 3a. Despite the differences in

AEC mode selection, in both patients the mAs was dramatically reduced in the lungs. The only difference in mAs variation pattern between Figure 3a and b is that in Figure 3b the mAs remained constant in the whole abdominal and pelvis area and equal to the maximum mAs value allowed because of large body size of the specific patient. A similar mAs variation pattern to that seen for scanner B (Figure 3b) was also observed for scanner C (Figure 4).

Figure 2. The mAs variations for the examination of patient A—1 of Table 3 performed in scanner A are shown with respect to the slice location and superimposed on the scout scan image.

Figure 3. (a) The mAs variations for the examination of patient B—4 of Table 3 performed in scanner B are shown with respect to the slice location and superimposed on the scout scan image. (b) The mAs variations for the examination of patient B—1 of Table 3 performed in scanner B are shown with respect to the slice location and superimposed on the scout scan image.

The relative dose reduction or increase obtained with the use of AEC for the examinations studied can be easily calculated by simply replacing in the calculation spreadsheets the actual mAs with the preset constant mAs value that is used in the respective examination protocol of each scanner. For example, for the examination shown in Figure 2, a pitch factor of 0.9 was used and therefore the mAs value that would be used if the AEC system was deactivated is 180 mAs. The respective DLP_c value would be 832.9 mGycm instead of 764.5 mGycm and therefore the use of AEC offered an 8% reduction. Similarly, for the examination shown in Figure 4, if the chest examination was performed with 160 mAs and the abdomen/pelvis examination with 200 mAs, the respective DLP values would be 226 mGycm and 297 mGycm instead of 170.2 mGycm and 281.1 mGycm, respectively. Therefore, the respective dose reductions offered by the AEC system were 25% and 5%, and overall for both

examinations 14%. However, if for the abdomen/pelvis examination a value of 160 mAs was used, the respective DLP value would be 237.6 mGycm. Therefore, in this case the AEC increased the dose by 18%, with an overall reduction for both chest and abdomen/pelvis examinations of just 3%.

It must be clarified that the aforementioned dose comparisons have been calculated based on the DLP differences. The respective differences in effective dose will not be the same since the conversion factors from DLP to effective dose are actually a function of the anatomical area irradiated. Furthermore, it should always be kept in mind that the purpose of the AEC systems in CT is not to reduce the patient dose but to optimise the mAs used during acquisition so as to ensure that all the images of a given patient are of approximately the same quality [4, 6]. Therefore, the dose to a large patient may actually increase with the use of the AEC but in return the image quality will be improved.

Figure 4. The mAs variations for the examinations of patient C—5 and C—8 of Table 3 performed in scanner C are shown with respect to the slice location and superimposed on the scout scan image. This was a chest, abdomen and pelvis examination that was performed with two scans, one for the chest examination and one for the pelvis–abdomen examination.

Discussion

Although all modern CT scanners provide a display of $CTDI_{vol}$ and DLP values, these values are not always stored within the examination archives and therefore, if not manually recorded somewhere, they will be lost. Indeed, only in scanner B could the DLP values be retrieved after the end of the examination, since they were retained within the examination record. Manually recorded DLP and $CTDI_{vol}$ may not always be reliable, since the radiation technologist may write down wrong numbers, omit the DLP values of some parts of the examination or assign the sum of the DLP values of a two- or three-phase examination as the DLP value of a single-phase examination.

The method presented in this study provides medical physicists with a useful tool for patient dose survey purposes, since it is free from transcription errors and produces reliable estimates of both DLP and $CTDI_{vol}$ values. This method can be used in conjunction with a DICOM viewer, such as the eFilm Workstation, to correctly identify

the scan series concerning different anatomical areas or different examination phases and calculate the DLP for each individual scan series, anatomical area or examination phase, as well as the cumulative DLP for all the examinations performed in the same patient.

The Dicom Info Extractor was developed for extracting data from CT Dicom images; however, it can be easily customised for use with other digital diagnostic modalities, such as digital radiography and digital mammography systems, for patient dose survey purposes. Similar DICOM-based methods have been presented in the recent literature concerning digital radiography and mammography systems [8–10]. However, to our knowledge, no study exists in the literature for calculating the DLP and CTDI_{vol} in CT examinations from DICOM data.

It must be also noted that the Dicom Info Extractor that was presented in this study can be used to facilitate the applicability of other published methods for calculating the DLP, effective and skin dose in CT-guided interventional procedures [7, 11]. These methods required the manual transcription of the technical parameters from the DICOM headers into the calculation spreadsheets, a task that was rather tedious and time-consuming.

The main purpose of this study was to develop a method for calculating the DLP from DICOM data and validate its accuracy in difficult cases, such as when the tube loading is changing during the scan. In this study the comparison of displayed and calculated DLP values was made using the displayed CTDI_{vol} of axial scans as a basis, but it is obvious that measured CTDI_w values could be used instead. This would be useful for calculating the actual DLP values in cases where the displayed CTDI_{vol} and consequently the displayed DLP values were for some reason inaccurate [2]. Furthermore, as can be seen in Figures 2–4, this method is also useful for evaluating the behaviour of AEC systems incorporated in modern CT scanners.

Conclusion

In this study a method was developed for calculating the DLP values and the CTDI_{vol} in CT examinations. This is a useful tool for radiation dose survey purposes, given the fact that in Greece as well as in many other countries of the European Union such dose surveys are compulsory for establishing the local and national diagnostic reference levels in CT examinations. Apart from patient dose surveys, this method is useful for studying the behaviour of different AEC systems and for calculating the relative dose decreases or increases with respect to constant mAs examination protocols.

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The Dicom Info Extractor is available at <http://www.infomed.gr/en/links.aspx>

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Appendix A

Definitions of CTDI

The CT dose index (CTDI) was first introduced in the era of single-slice CT scanners and it was defined as the integral of the dose profile $D(z)$ from a single axial scan along a line perpendicular to the tomographic plane (z -axis) divided by the product of the nominal slice thickness (T):

$$\text{CTDI} = \frac{1}{T} \int_{-\infty}^{+\infty} D(z) dz \quad (\text{A1})$$

For the case of multislice CT scanners where N slices of thickness T are acquired during a single axial scan, two definitions of CTDI were introduced:

$$CTDI_{FDA} = \frac{1}{NT} \int_{-7T}^{+7T} D(z) dz \quad (A2)$$

$$CTDI_{100} = \frac{1}{NT} \int_{-50mm}^{+50mm} D(z) dz \quad (A3)$$

The most often used CTDI quantity is the $CTDI_{100}$, which is measured using a pencil-type ionisation chamber with an active length of 100 mm, both in free air and within two cylindrical polymethylacrylate (PMMA) phantoms of 16 and 32 cm diameter, simulating the head and body of a patient, respectively.

The $CTDI_{100}$ as measured with the ionisation chamber positioned in free air at the centre of rotation is referred to as $CTDI_{air}$. As $CTDI_c$ and $CTDI_p$ are defined respectively the $CTDI_{100}$ values measured with the ionisation chamber within the centre and four positions (12, 3, 6 and 9 o'clock) in the periphery (1 cm for the surface) of the two centrally positioned head and body phantoms.

Definition of $CTDI_w$

The weighted CTDI ($CTDI_w$) is used for approximating the average dose over a single slice and is defined by the following equation, separately for the head and the body phantoms:

$$CTDI_w = \frac{1}{3}CTDI_c + \frac{2}{3}CTDI_p \quad (A4)$$

where for $CTDI_p$ the average of the four roughly equal $CTDI_p$ values measured in the periphery of the phantom is used.

Definition of $CTDI_{vol}$

The volume-weighted CTDI ($CTDI_{vol}$) is used to account for helical scanning and it is defined by the following equation:

$$CTDI_{vol} = CTDI_w \cdot \frac{NT}{I} = \frac{CTDI_w}{pitch} \quad (A5)$$

where NT is the total nominal collimation width and I is the table travel per rotation during a helical scan (pitch factor = $\frac{I}{NT}$).

Definition of DLP

Dose-length product (DLP) is used to calculate the dose for a series of scans or a complete examination and is defined by the following equation:

$$DLP = \sum_{i=1}^N (CTDI_{vol})_i \cdot L_i \quad (A6)$$

where i represents each one of the individual scans of the examination that cover a length L_i of patient anatomy.