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Spectrum Optimization of Dual Energy CT Scan

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Abstract

Dual-Energy Computed Tomography (DECT) is known as a useful method to discriminate different objects with similar chemical properties. DECT measures the linear attenuation coefficient of the scanned object at two different x-ray energies. Using these two energy spectra, we can characterize the scanned objects by using the dependency of the linear attenuation coefficients on the x-ray energy and chemical characteristic of the material. Finding the optimum source spectrums for DECT method to attain the best material discrimination, is the aim of this study. Materials and Methods: The present study uses a simulation method to find the maximum differences between the HU values of the tissuesat 80 and 140 kvp which have the same attenuation properties, such as soft tissue, muscle, lung, and bone. IMASIM software is used to simulate different source spectrums by using of the different filters, such as aluminum, tin and copper, to produce the phantom images. The HU values of different tissues at different energies, are then measured and compared quantitatively. Results: The results show that the maximum differences between the HU values, pertain to cortical bone, for 80 and 140kvp when 2mm copper plus 0.4mm tin is used (45%). It has to be noted that the minimum differences (36%) between the HU values occur for cortical bone at 80 and 140 kvp and are recorded when the simulated source spectrums were made by 3mm Al plus 0.4mm tin filters. Conclusion: The results of this study demonstrate that the highest material discrimination may occur at 80 and 140kVp when 2mm Copper plus 0.4 mm tin filters are used as the added filters in front of the x-ray source with 140kVp.

Keywords: DECT, Filter, x-ray, Spectrum, Attenuation Coefficient

Introduction

It is known that the basis of image formation in CT is the average linear attenuation coefficient of the material on the path of the x-ray beam As is understood, the linear attenuation coefficient in the range of diagnostic radiology is a combination of the Compton scattering and photoelectric absorption as is demonstrated by Eq. 1 (Haghighi et al., 2015).

$$\hat{\mu}(V) = \hat{\mu}_{Comp}(V) + \hat{\mu}_{Photo}(V) \qquad (1)$$

Where $\hat{\mu}_{Comp}(V)$ and $\hat{\mu}_{Photo}(V)$ are the average attenuation coefficients due to Compton scattering and photoelectric absorption respectively (in cm^{-1}).

The probability of these phenomena are dependent on the chemical characteristic of the object, such as effective atomic number (Z_{eff}) and electron density (ρ_e), as well as the energy (E) of the x-ray photons. These relationships are presented in Eq. 2 for mono-energetic x-ray beam, (Haghighi et al., 2015)

$$\mu(V) = \rho_e\left[\alpha_0 f_{KN}(E) + \beta_0 f_{Ph}(E) Z_{eff}^{x}\right] \qquad (2)$$

Where the exponent x of the effective atomic number (in the photoelectric term) is also dependent on the material under study as is explained in Ref. Med Phys (2011), with $\alpha_0 = 66.62\times10^{-26} cm^2$, $\beta_0 = 54.7578\times10^{-18} cm^2$ and $f_{KN}(E)$, $f_{Ph}(E)$ are the coefficients which respectively to describe the energy dependency of the Compton scattering (Kelin-Nishin) and photoelectric effect. While the energy dependency of compton scattering is weak,as is shown in Eq. (3), the photoelectric effect is strongly dependent on the energy of the x-

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ray photon (Eq. 4),

$$f_{KN}(E) = \left(\frac{3}{4}\right)\left\{\left(\frac{1+\gamma}{\gamma^3}\right)\left(\frac{2\gamma(1+\gamma)}{1+2\gamma} - \ln(1+2\gamma)\right) + \frac{1}{2\gamma}\ln(1+2\gamma) - \frac{1+3\gamma}{(1+2\gamma)^2}\right\} \quad (3)$$

With γ= E/mc^2 where mc^2=511.8 keV (m is the mass of the electron and c is the speed of light in vacuum) (Haghighi et al., 2015).

$$f_{Ph}(E) = \left(\frac{I_0}{E}\right)^3 \quad (4)$$

In Eq. (4) I_0=13.5 eV is the ionization energy of the hydrogen from the first Bohr orbit.

X-ray photons emitted from x-ray source in diagnostic radiology such as CT has a broad range of energies in the source spectrum. The minimum and maximum energy of the x-ray photons are determined by the inherent and added filters and the voltage applied to the x-ray tube. The type of the interaction of x-ray photons with material is determined by the average x-ray energy in the source spectrum. As is explained in Eq.(5) the source spectrum is given by .

$$S(E,V) = \int S_0(E,V)\exp(-\int \mu_i l_i)d(E) \quad (5)$$

Where *S (E,V)*is the source spectrum with filtration (inherent and added filter) and , *$S_0(E,V)$* is the original x-ray photons without filtration. As can be seen in Eq. (5) the total attenuation coefficient of the inherent and added filter is, (Haghighi et al., 2015)

$$\int \mu_i l_i = \mu_1 l_1 + \mu_2 l_2 + \ldots \quad (6)$$

Where μ_1, μ_2, are the linear attenuation coefficients of the filter materials, l_1, l_2, are the thicknesses of the related filters.

The average energy is given by,

$$\hat{E} = \frac{\int S(E,V)Ed(E)}{\int S(E,V)d(E)} \quad (7)$$

DECT Method:

Scanning the same object by two different x-ray energies is the basis of Dual Energy Computed Tomography (DECT). DECT can be used to discriminate two materials with similar chemical properties due to the dependency of the attenuation coefficient on the x-ray energy. On the basis of the above explanations x-ray energy spectrum at each tube voltage is determined by the inherent and added filters as is seen in Eq. (5). The best material discrimination in DECT evaluation is dependent on the maximum energy separation between two x-ray energy spectra (Bushberg and Boone, 2011).

The idea of producing the desired contrast between two materials, with similar chemical properties, by two x-ray energy spectra (DECT) was known since the invention of the CT system.But this was not applied in CT scan at that time because of technical difficulties such as patient's involuntary movement (Johnson, 2012).

DECT image can be produced by different methods by modifying switching voltage (Johnson, 2012), and using different components like two layer detector (stacked detectors) (Johnson, 2012), two x-ray tubes perpendicular to each other's with two sets of detectors (DSCT) (Johnson, 2012), and one x-ray tube with split filter (Almeida et al., 2017).

In all of DECT methods, the maximum separation between two x-ray energy spectra is desired to produce the best discrimination (contrast) between objects with similar chemical characteristics. Each x-ray energy spectrum as is explained in Eq. (5) has the maximum intensity distribution around the average of x-ray spectrum. The average x–ray energy spectrum is called effective energy (Eq.(7)). Effective energy of the x-ray beam is equivalent to the mono-energetic one with the same x-ray attenuation coefficient properties.

Figure 1 shows the example of x-ray spectra at 80 and 140 KVP tube voltages when tube filters for both of these spectra were 3 mm Al. As Fig.(1) shows, the maximum energy is determined by the tube voltage and the average energy or effective energy occurs at 56keV and 76 keV respectively (Johnson, 2012).

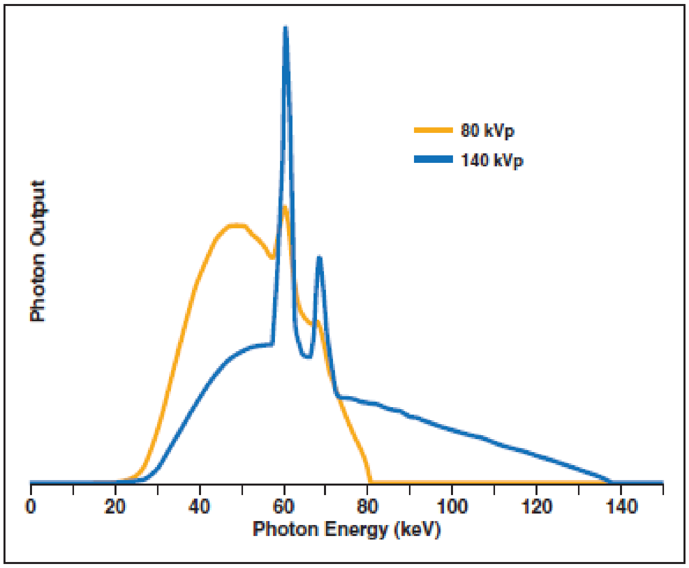


Figure 1. Simulated x ray spectra for 80 and 140 kVpbased on Monte Carlo (Johnson, 2012).

For optimization of X-ray spectrum in DECT, a number of studies have been done so far. A. N. Primaket al used additional filtration for dual-energy (DE) imaging, using a dual-source CT (DSCT) system. Their results showed that added filter can be used to determine material specifications by DE ratio method. The X-ray spectra, data acquisition, and reconstruction processes for a DSCT system Siemens Definition were simulated using information provided by the system manufacturer, resulting in virtual DE images. Different materials were used as filters, with atomic numbers between 40 and 83, in front of the x-ray source with 140kVp, to simulate different x-ray source spectra. DE ratio of Calcium Hydroxyapatite, iodine, and iron were calculated for 80/140 kVps. Their results showed that the DE ratio (80/140 kVp) increased for Ca, I, and Fe by increasing the atomic number of added filters (Primak et al., 2009).

In another study M.Saito optimized the x-ray spectra of DECT by balanced filter method. The author showedthat beam hardening and the tube-loading efficiency with balanced filters would be significantly more effective than conventional DECT for electron density measurement (Saito, 2011). Also, in another study, he used different filters in front of low kVp x-ray source in order to optimize source spectrum in DECT method (Saito, 2009).

The author concluded that electron density measurement by using tungsten filter with 0.14 mm thickness at low kvp is more accurate than 0.4mm tin filter for higher kVp (140kVp) in second generation of DECT (Saito, 2009).

Landry and Deblois used IMASIM software, designedbyFrankVerhaegen, (Verhaegen, 2013) in their study. It has to be noted that IMASIM software can simulate Radiography, CT scan and Radiation therapy up to 6 MV energy. One can use different x- Ray energy and different filters to produce different spectra, to be used for different phantoms .In this software RMI 465 phantom with 20 tissue mimicking inserts and materials is simulated as a template. User can change inserts and shape of the template phantom. Landry et.al found that the differences between Hounsfield Unit of simulated and empirical methods are within 2% (Verhaegen, 2013).

The other phantom which can be defined for IMASIM software is RMI 467. This phantom is made by Gammexcompany. It contains16 tissue mimicking inserts.RMI 467 phantom is used for electron density and effective atomic number (Z_{eff}) calculation by HU value measurement as is shown in Figure 2

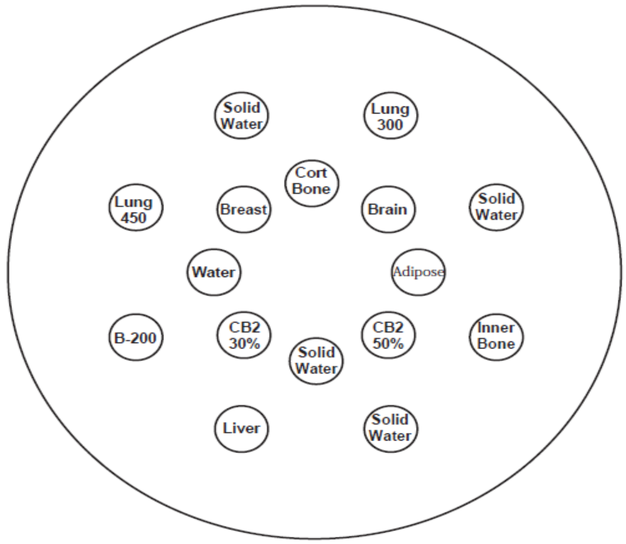


Figure 2. Pattern of Gammex467 phantom shows different material inserts

The aim of this study is to optimize X-ray source spectrum of DECT by simulation method. This study tries to simulate x-ray energy spectra for DECT by adding different filter materials with different thicknesses. This research will examine different combination of source spectra at two tube voltages to find the best material filter and its desired thickness to produce the maximum material discrimination through HU measurement at two different tube voltages (DECT).

Materials and Methods:

In the present study, IMASIM software[1] was used due to its availability as an open access source, user friendly and has a lot of choices to select different types of x-ray energy and different filters to simulate different source spectra. As it is mentioned, IMASIM software can help the user to simulate phantoms with different shapes and different material inserts .Also this software permits the user to select different types of detectors and image matrix (for example 256x 256 matrix).In this research RMI 467 phantom with 16 inserts was simulated. This simulated phantom has a cylindrical shape with 31cm diameter and its height is 5 cm. Each insert has a 28.5mm diameter as seen in Figure 2.the arrangement of material is such that to separate bone materials for diminishing artifact (Haghighi et al., 2011).

The X-ray sources with two X-ray energies, 80 and 140 kvp, are simulated by IMASIM. This selection is based Somatom Definition Flash DSCT scanner .Somatom Definition Flash CT Scanner uses 3mm aluminum (Al) and 3mm Al plus 0.4mm tin (Sn) respectively as added filters in DECT mode of this scanner. In this study, instead of the conventional filters uses in Somatom Definition Flash CT Scanner in front of x-ray source with 140kVp excitation voltage were replaced by Copper (Cu) and Titanium (Ti) filters. The thicknesses of added filters for 140kVp voltage are as follows; copper (Cu)from 0.5 to 2.5 mm and Titanium (Ti) from 0.5 to 3mm. Copperand Titanium filters were chosen owing to easy to produce thin sheets and low atomic number (near Aluminum) respectively.
The simulated spectra at 80 kvp, with conventional filter (3mmAl), and 140kVp, with different thicknesses of Cu (0.5 to 2.5mm) and Ti (0.5 to 3mm), were used to produce CT images of the RMI 467 phantom.

To simulate CT images of the phantom, IMASIM software was used with 256*256 matrix size, gadolinium oxysulfide (Gd2O2S)as a detector, and Filter Back Projection (with Cosine filter) algorithm to reconstruct images .as by Verhagen (Verhaegen, 2013).

The simulated axial slices of the phantom were used to measure the Hounsfield Units (HU) of the inserts. HU values were measured by selecting the suitable ROI at the center of each insert in such a way that it should not contaminate with neighboring structures.HU values of each insert at 140 kVp for different types of filter combinations (different thicknesses of Cu and Ti) were compared with the measured HU values of the same insert at 80 kVp (with 3mmAl) and 140kVp (with 3mmAl plus 0.4mmSn) source spectra produced by conventional filters.

The differences between HU of simulated conventional filters and Copper, Titanium with and without 0.4mm Tin were calculated by Excel to optimize best filters.

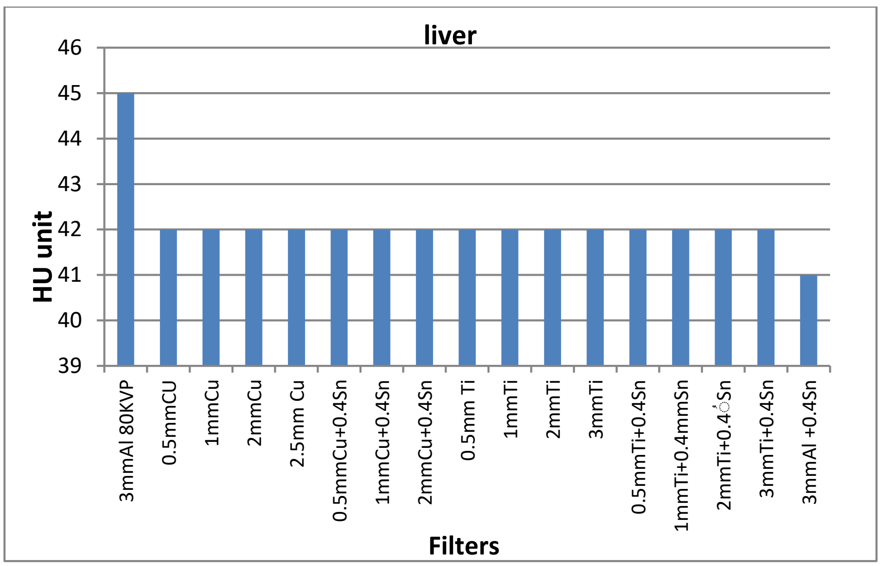
[1] http://www.medphys.mcgill.ca/~ImaSim/

Results

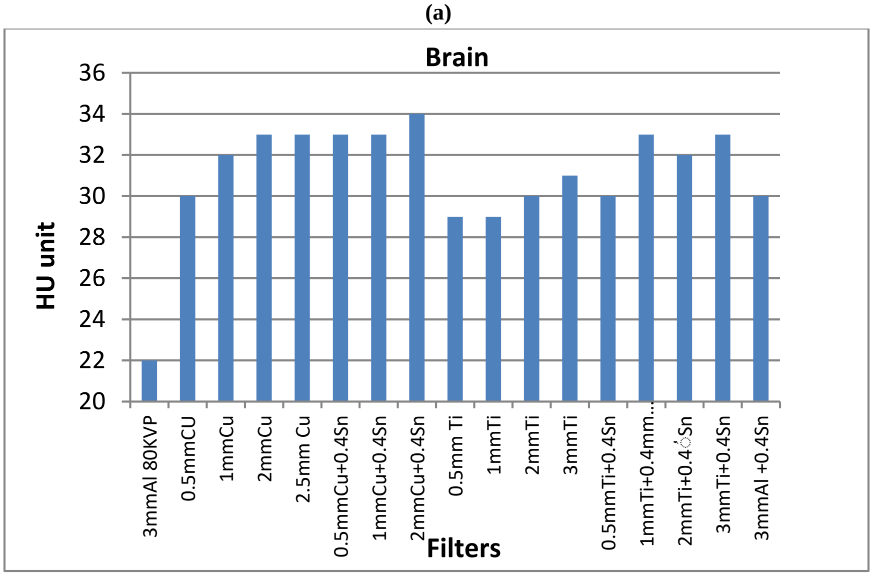
Figures.3,shown that when the source spectrum formed by applying 3mmAl in front of the x-ray source at 80 kVp excitation voltageused to measure the HU(80), the maximum differences revealed with the HU value derived from 140 kVp excitation voltage, with superimposing 2mmCu plus 0.4mmSn as added filters. This significant differences between HU(80) and HU(140) are seen in figures .3, for cortical bone, calcium carbonate 50%, calcium carbonate 30%, B200 (bone mineral), inner bone, adipose tissue, and Breast. However, in adipose tissue small differences in Hounsfield are not tangible in different X-ray spectrum. This difference, between HU(80)/HU(140), is enough to produce high contrast resolution to discriminate the above materials which are inserted in the 467 Phantom by simulation study.

Also, the HU(80) has appreciable differences with HU(140) for the same material, in the inserted samples in the phantom, when 1mmCu plus 04mmSn have been used as added filters in front of the x-ray source with 140 kVp excitation voltage.

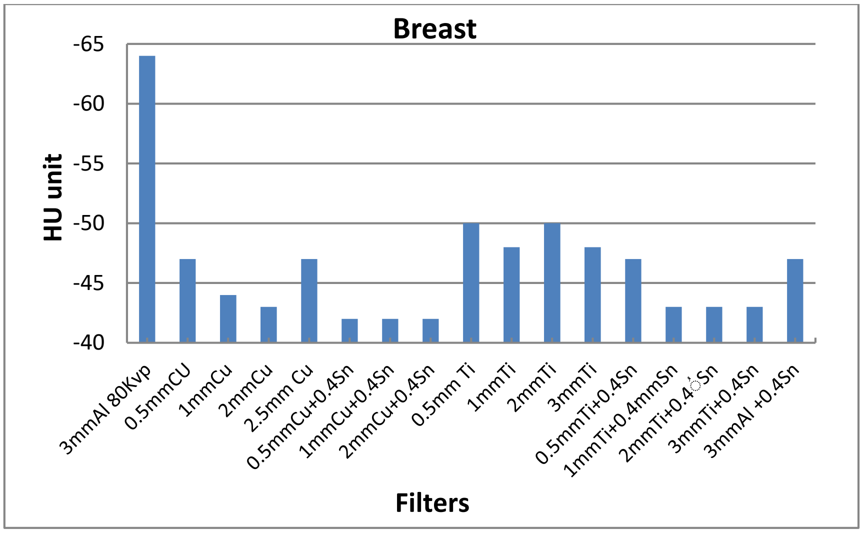
Figures 3.shown that the difference of HU(80)/HU(140) with 2mmCu plus 0.4mmSn added filters, inserted in the path of the x-ray beam with 140kVp excitation voltage, are higher than the same result due to applying 3mmAl plus 0.4mmTi (as added filters in front of 140kVp excitation voltage). It has to be noted that the later filter configuration is used in CT Somatom definition flash system.



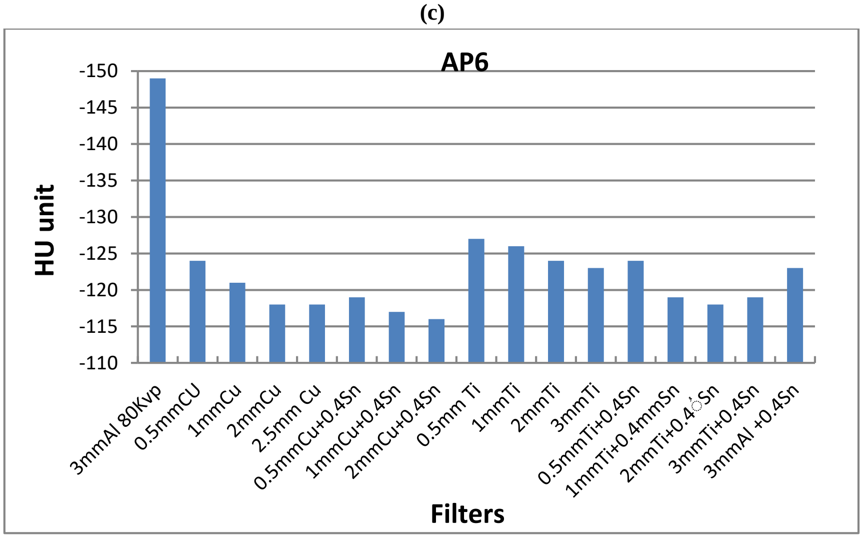
(a)



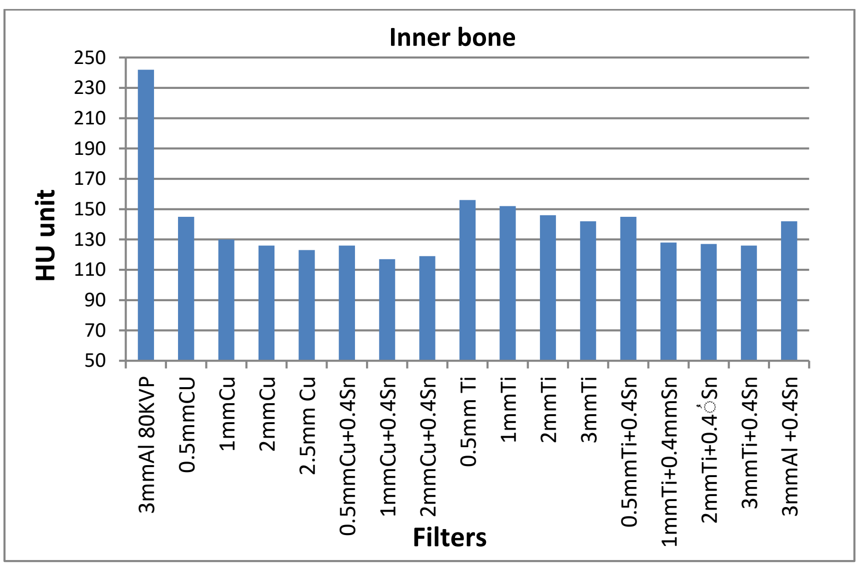
(b)


Breast
HU unit
-65
-60
-55
-50
-45
-40
3mmAl 80Kvp
0.5mmCU
1mmCu
2mmCu
2.5mm Cu
0.5mmCu+0.4Sn
1mmCu+0.4Sn
2mmCu+0.4Sn
0.5mm Ti
1mmTi
2mmTi
3mmTi
0.5mmTi+0.4Sn
1mmTi+0.4mmSn
2mmTi+0.4Sn
3mmTi+0.4Sn
3mmAl +0.4Sn
Filters

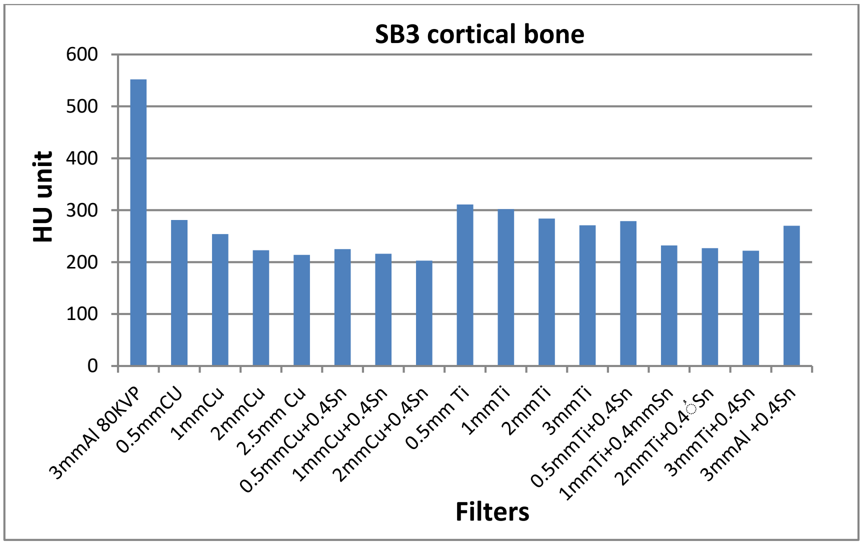
(c)


AP6
HU unit
-150
-145
-140
-135
-130
-125
-120
-115
-110
3mmAl 80Kvp
0.5mmCU
1mmCu
2mmCu
2.5mm Cu
0.5mmCu+0.4Sn
1mmCu+0.4Sn
2mmCu+0.4Sn
0.5mm Ti
1mmTi
2mmTi
3mmTi
0.5mmTi+0.4Sn
1mmTi+0.4mmSn
2mmTi+0.4Sn
3mmTi+0.4Sn
3mmAl +0.4Sn
Filters

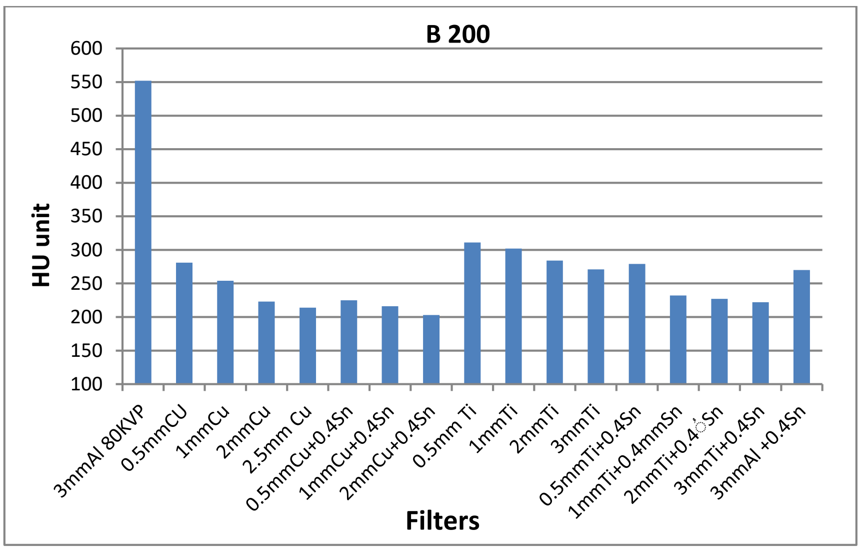
(d)


Inner bone
HU unit
250
230
210
190
170
150
130
110
90
70
50
3mmAl 80KVP
0.5mmCU
1mmCu
2mmCu
2.5mm Cu
0.5mmCu+0.4Sn
1mmCu+0.4Sn
2mmCu+0.4Sn
0.5mm Ti
1mmTi
2mmTi
3mmTi
0.5mmTi+0.4Sn
1mmTi+0.4mmSn
2mmTi+0.4Sn
3mmTi+0.4Sn
3mmAl +0.4Sn
Filters

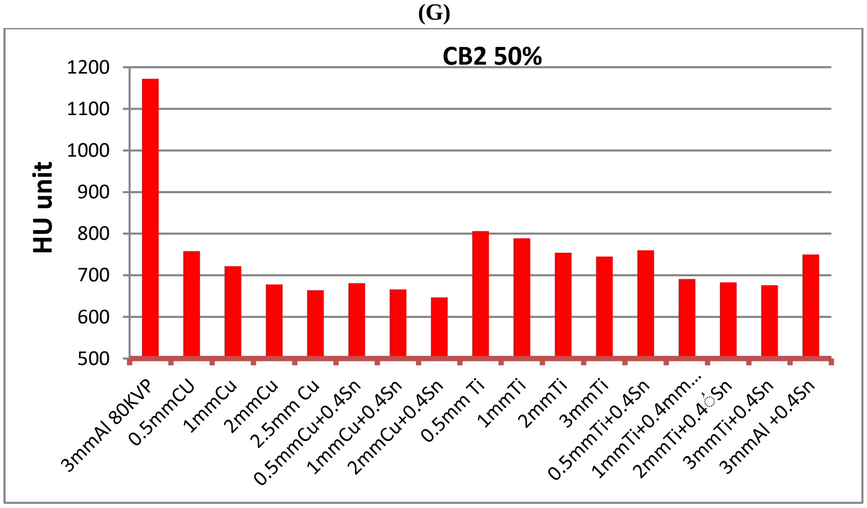
(e)



(f)



(G)



(H)

Fig. 3. Hounsfield unit versus different filters of liver, brain, breast, AP6, inner bone, cortical bone, B200and CB2-50% which was simulated by IMASIM

Table 1 shows that while the effective energy has the maximum value of 99.2 keV, at 140 kVp x-ray source with 2mm Cu plus 0.4mmSn, the related output has the minimum value of 8.2µGy/mAs. According to Imasim simulation applying 1mmCu plus 0.4mmSn added filter is able to produce considerable amount of x-ray output (13.42µGy/mAs) compare to the above added filter (2mmCu plus 0.4mmSn). The effective energy of the two cases, 2mmCu+0.4mmSn and 1mmCu+0.4mmSn, are comparable.

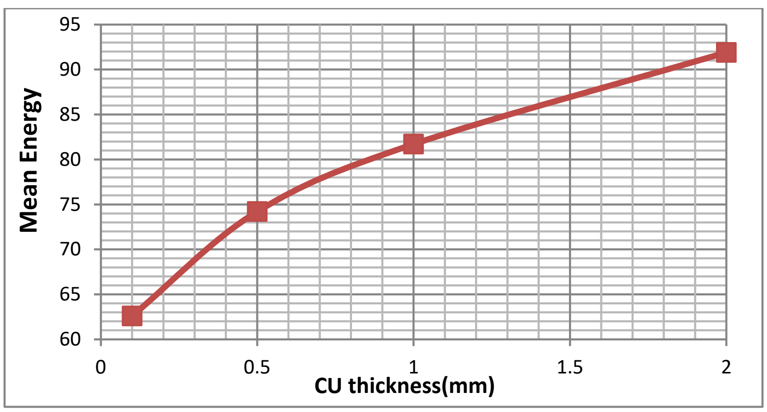
Also, Table .1 shows that the x-ray output in the form of Bremsstrahlung decreases as the thickness of copper filter increases from 0.1 to 3mm with fixed Tin filter (0.4mm thickness).

It can be seen in Figs.3 that the effective energy shifts to higher side of energy as the thickness of copper filter increases (when the thickness of Tin filter is fixed at 0.4mm) in the path of the 140kVp x-ray beam.

As seen adding the tin filter to either copper or titanium causes an increment in the effective energy of the x-ray spectrum but it decreases the bremsstrahlung and characteristic x ray air kerma thus the mAs should be increased for compensation has to shift to discussion

Table 1.filter thickness, effective energy , doses from Bremsstrahlung ,characteristic x-ray and mean energy from copper and copper + 0.4mm Sn

Filter Cu(mm)	Effe Energy(keV)	Brems X ray(uGy/mAS)	CharX ray mGy/mAs	Mean Energy(keV)
0.1	47.4	110.7	14.4	62.6
0.5	66.3	46.67	8.18	74.2
1	76.2	28.4	4.075	81.7
2	88	14.88	1.048	91.9
Filter(Cu+0.4tin)	Effe Energy(keV)	Brems X ray (uGy/mAS)	CharX ray mGy/mAs	Mean Energy (keV)
0.1	79.2	22.13	2.19	86.8
0.5	87.2	17.11	1.29	91.1
1	91.8	13.4	0.66	95
2	99.2	8.28	0.18	101



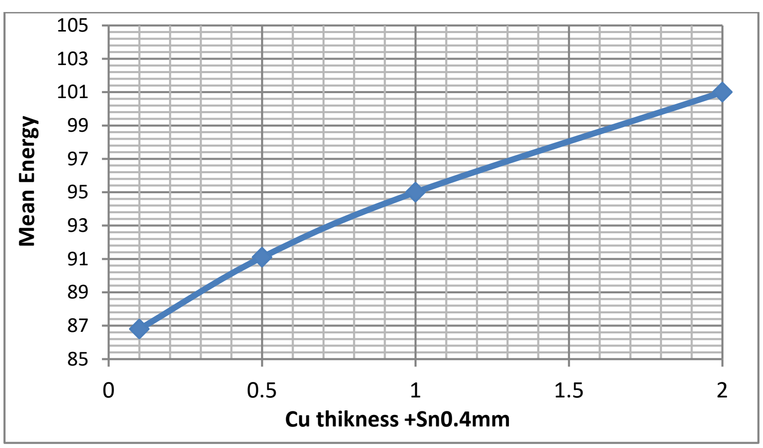


Fig 3. Mean Energy versus Copper thickness and Copper +0.4mm Sn filtration

Table 2 states that the effective energy increases as the thickness of Titanium filter and titanium filter plus tin increases. As it is expected the x-ray output decreases by increasing the thickness of the titanium filter alone and also titanium in combination with tin filters.

Table 2.filter thickness, effective energy, doses from Bremsstrahlung ,characteristic x-ray and mean energy from titanium, titanium + 0.4mm Sn and 3mm Al +0.4 Sn x-ray spectrum

Filter(Ti)mm	Effe Energy(keV)	Brems X- ray(uGy/mAS)	CharX ray (mGy/mAs)	Mean Energy(keV)
0.5	48.7	103.1	13.97	63.3
1	56.4	72.44	11.77	67.6
2	64.8	47.34	8.36	73.1
3	70	35.17	5.95	76.9
0.5+0.4Sn	80	21.74	2.15	87
1+0.4Sn	83.3	19.63	1.82	88.4
2+0.4Sn	86.6	16.61	1.31	90.6
3+.4Sn	89	14.24	0.95	92.5

It can be seen in Table 3 that the effective energy and x-ray output of the 140kVp x-ray beam due to 3mmAl plus 0.4mmSn are comparable to the related values with applying 0.5mmTi+0.4Sn and 0.1mmCu+0.4mmSn.

Table 3. Mean energy versus Copper thickness, Copper +0.4mm tin filtration and 3mm Al with 0.4mmSn

Filter	Effe Energy(eV)	Brems X ray (uGy/mAS)	CharX ray (mGy/mAs)	Mean Energy(keV)
80kvp 3mm Al	43.6	16.64	0.4013	50.9
140KVp 3mmAl +0.4Sn	78.2	20.47	2.042	86.5

Discussion

In this study x-ray source spectrum were simulated using various filters with different thicknesses and material combinations at 140kvp. The simulated x-ray source spectrum were evaluated by scanning the same material with different filter combinations. The HU value of the scanned material at 140 kVp was used to assess the effect of filter combination in the path of the x-ray beam.

The results show that adding tin filter to the either copper, aluminum or titanium increases the Hounsfield differences at 80 and 140 kVp of bone , brain, breast, Liver, B200, cortical bone,SB3 and CB2 50. However in comparison between different filter combinations 2mm Cu+ 0.4mm Sn produce the maximum differences between HU(80) and HU(140) or has the highest DE ratio. Then, this can be used as a good choice for DECT tube filtration in front of 140kVp source spectrum. The main disadvantages of this filter combination (2mmCu+0.4mmSn in front of 140kVp) is increasing noise which can be compensated by mAs increment.

This type of filtration can reduce more low energy part of the spectrum. It has shown in table 2 that kerma or dose per mAs decreased significantly by simulating this type of filter combination (2mmCu)+0.4mmSn) at 140kvp, ….mGy/mAs, compare to to 3mm Al +0.4mm Sn, …mGy/mAs. Therefore radiation dose to the patient is less than the current type of filter (3mmAl+0.4mmSn) which is used in Somatom Definition Flash CT Scanner.

The effective energy of the source spectrum shifted to 99.2 keV with 2mmCu plus 0.4mmSn filter combination at 140kVp. While the effective energy is 78.2 keV with 3mmAl plus 0.4mmTi in Dual-Source Dual Energy CTlike Somatom Definition Flash system. It is clear that the energy separation between 80 kVp (with E_{eff}=43.6keV) and 140 kVp (E_{eff}=99.2keV) is increasing due to applying suggested filter combination (in front of 140 kVp x-ray source). The effect of this energy separation is manifested in the HU(80) and HU(140) of the same material as is shown in Figs …..Then it can be explained the cause of the significant differences between HU(80) and HU(140) for high and low dense tissues such as bone and breast respectively.

Primak simulated source spectrum by using aluminum filter in his study (Primak et al., 2009). While Masatoshi Saito used different materials with different thicknesses to simulate different source spectrum with different filter combinations. He measure beam hardening effect and noise level in his research. He also measured air kerma in different simulated source spectrum to calculate electron density. Calculated electron density was used as an input factor in treatment planning system to calculate dose distribution for therapy purposes. This study also showed that the tube output and air Kerma decrease with increasing the thickness of the filter. He found that the beam hardening error and noise have the lowest values with Bi/Mo filter combination (Saito, 2011; Saito, 2009).

Conclusion

Using 2mmCu plus 0.4mmSn filter at 140 kVp with conventional 3mmAl at 80kVp may produce better energy separation in Dual Source and Dual-Energy CT systems. The source spectrum produced by this type of filter combinations are able to discriminate materials with similar chemical characteristics which cannot discriminate by conventional CT and hardly can distinguished by the current Siemnse Dual-Source Dual-Energy Flash system. The limitation of this type of filtration is quantum noise which can be compensated by higher mAs.

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