Computed tomography protocols optimization using non-prewhitening model observer

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Abstract. The non-prewhitening model observer with eye filter (NPWE) allows assessing of performances of a whole imaging system, taking into account specific aspects of a considered clinical task and also the aspects of human visual system. The aim of this work is to determine the exposure parameters for computed tomography (CT) scan protocols so that they produce the satisfied performances of a human observer (HO). For each of the chosen scan protocols, the assessment of image quality parameters, based on the Fourier metric, as well as the volume CT dose index (CTDI\textsubscript{vol}) measurements, have been performed. The scan protocols using the smooth reconstruction kernel, for an object diameter of 3 mm, 7 mm and 10 mm, and a contrast level of 20 Hounsfield units (HU) and 40 HU, gave the values of the area under curve (AUC) greater than 0.75 for CTDI\textsubscript{vol} ≥ 7.16 mGy. For a sharp kernel, all AUC values were smaller than 0.75 for all investigated scan protocols, object diameters and contrast levels. Through this work it was shown that the NPWE model-observer can be used as an effective and comprehensive optimization tool suitable for application in the various clinical tasks, considering also CTDI\textsubscript{vol} values.

1. Introduction
Optimization is, in addition to justification, one of the basic principles of radiation protection in medical exposure to ionizing radiation [1]. The basic assumption for obtaining the reliable diagnostic information, and thus to achieve the expected benefit, is suitable quality of produced images. Thereby, the radiation dose reduction shouldn’t be such that it affects the image quality to the extent that would compromise the reliability of diagnostic information (ALARA principle).

One of the possible strategies to perform the optimization process is to select the appropriate, optimized scan protocols based on an assessment of image quality, which are obtained under different values and combinations of exposure parameters [2]. This is particularly important for computed tomography (CT) examinations in which the patient doses are at such a level that there exists a risk of a...
cancer induced by ionizing radiation [3]. Standard image quality control tests are not an exhaustive source of information needed for the purpose of optimization. There is a necessity to perform a more robust assessment with the complete imaging chain as well as an observer (radiologist) and a specific clinical task under consideration.

The gold standard in the observer performance assessment is the receiver operating characteristics (ROC) analysis based on the signal detection theory [4-6]. Basically, the ROC method poses a binary problem for the observer who needs to determine the presence or absence of an object (signal) in the specific region of interest (ROI) chosen from the image. The area under the ROC curve (AUC), the detectability index $d'$, or the percentage of correct answers are used as the principal figure of merit (FOM) to describe observer performances in CT protocols optimization [7].

A mathematical or a statistical model-observer (MO) can be used to make predictions of the human observer (HO) performances, for a specific clinical task (so-called a task-based MO), providing indirectly an assessment of image quality [8-10]. As such, MO is suitable as a tool for performing CT protocols optimization [7]. Mathematical MO takes into account physical parameters of image quality (e.g. modulation transfer function (MTF), noise power spectrum (NPS)), obtained on the basis of the signal detection theory, as well as a specific clinical task under consideration. Although by themselves, physical parameter values of image quality do not provide information about possible human observer performances, there still exists the correlation between these values and performances of the human observer [11-14].

The aim of this work is to present the way in which the optimized CT scan protocols can be selected, using the non-presh whitening matched filter with eye (NPWE) model-observer method, on the basis of requested human observer performances (i.e. AUC values) and obtained volume CT dose index (CTDIfd) values.

2. Materials and methods

2.1. Theoretical basis

2.1.1. Modulation transfer function (MTF): The change in the amplitude of the output signal provides the information about the contrast transfer capability of an imaging system for an input signal $f(x)$ at a given spatial frequency ($u$). The MTF quantitatively describes the ability of an imaging system to transfer the contrast depending on the signal spatial frequency. It can be defined using a characteristic function of the system [15]:

$$MTF(u) = \frac{|\mathcal{F}(u)|}{|\mathcal{F}(0)|} = \frac{|\mathcal{F}(\text{PSF}(x))|}{\int_{-\infty}^{\infty} \text{PSF}(x) dx}$$

(1)

where $u$ is the spatial frequency, by definition $MTF(0) = 1$, $T(u)$ is the system’s characteristic function [15] calculated as the Fourier transform of $\text{PSF}(x)$, i.e. $\mathcal{F}(\text{PSF}(x))$, PSF is the point spread function generated as the result of the system's response to a point source input signal.

If the input signal is in the form of a line, the system response function is given as the line spread function (LSF) which is related to the PSF according to the equation:

$$\text{LSF}(x) = \int_{-\infty}^{\infty} \text{PSF}(x,y) dy.$$  

(2)

When a sharp edge object is used to produce an input signal to the imaging system, the corresponding impulse response function is called the edge spread function (ESF) which can be expressed as:

$$\text{ESF}(x) = \int_{x'=-\infty}^{x} \text{LSF}(x') dx'.$$

(3)

According to the equation (3), if the ESF of the system is known the LSF can be determined as [7]:

\begin{align*}
\text{LSF}(x) & = \int_{-\infty}^{x} \text{ESF}(x') dx' \\
& = \int_{-\infty}^{x} \int_{-\infty}^{\infty} \text{PSF}(x',y) dy dx'.
\end{align*}
\[
\text{LSF}(x) = \frac{\partial \text{ESF}(x)}{\partial x}.
\]

When the LSF of an imaging system is known the MTF can be expressed as:

\[
\text{MTF}(u) = \frac{\int_{-\infty}^{\infty} \text{LSF}(x) e^{-2\pi iux} dx}{\int_{-\infty}^{\infty} |\text{LSF}(x)| dx} = \frac{|\text{FT} (\text{LSF}(x))|}{\int_{-\infty}^{\infty} |\text{LSF}(x)| dx}.
\]

In the case of the imaging systems that use iterative reconstruction (IR) algorithms, contrary to the ones that use filtered back projection (FBP) reconstruction algorithms, the assumption that the system is linear and shift invariant (LSI) is not satisfied and the system resolution shows dependence on an image object’s contrast [16]. In this case the different approach is used to determine MTF of an imaging system. The task transfer function (TTF), obtained from the ESF of a disk-shaped image object produced by a rod of a certain diameter inserted into the phantom, is used to describe the contrast dependent resolution performance of an imaging system [17-19]. In the case of a FBP reconstruction algorithm, the TTF value is identical to the MTF value obtained using the traditional wire method [17,20].

2.1.2. Noise power spectrum (NPS): The presence of noise, in addition to the blurring effect, is another factor of image quality degradation. The Fourier metric in the form of NPS is used to fully describe the noise characteristics of an imaging system taking into account the noise texture and correlations, e.g., the spatial frequency noise dependence. For CT systems NPS can be determined from the images of a homogenous test object – usually a water filled cylindrical shaped object [19]. The two dimensional NPS can be obtained as:

\[
\text{NPS}(u, v) = \frac{\Delta_x \Delta_y}{L_x L_y N_{ROI}} \sum_{i=1}^{N_{ROI}} |\text{FT}_{2D}(\text{ROI}_i(x, y) - \bar{\text{ROI}}_i)|^2,
\]

where \(\Delta_x\) and \(\Delta_y\) are pixel sizes in \(x\) and \(y\) direction; \(L_x\) and \(L_y\) are side lengths of a ROI in \(x\) and \(y\) directions, \(N_{ROI}\) is the number of the chosen ROIs, \(\text{FT}_{2D}\) is Fourier transform in 2D space and \(\bar{\text{ROI}}_i\) represents the mean pixel value of the \(i\)th ROI.

2.1.3. Non – prewhitening model observer with eye filter: Using the objective image quality metrics in Fourier domain, e.g. TTF and NPS, the NPWE model/observer allows the performances of a whole imaging chain to be assessed taking into account some aspects of the considered clinical task as well as the aspects of the human visual system [7,19]. The detectability index, \(d^*\), as the corresponding FOM of NPWE model observer, can be obtained as:

\[
\text{d}^*_\text{NPWE} = \frac{\int|W(u,v)|^2\text{TF}^2(u,v)E^2(u,v)du dv}{\sqrt{\int|W(u,v)|^2\text{TF}^2(u,v)\text{NPS}(u,v)E^4(u,v)du dv}},
\]

\(W(u,v)\) is a corresponding task function obtained as the Fourier transform of a signal with a Gaussian contrast profile defined as:

\[
c(r) = \frac{c}{2} \left(1 - \text{erf}\left(\frac{r-d}{n}\right)\right),
\]

where \(c(r)\) is a value of the signal at a radial distance \(r\), \(C\) is the peak contrast of the signal against the background expressed in HU, \(d\) is the signal (object) diameter, \(n\) is a blur factor of the Gaussian contrast-profile (e.g. standard deviation with a value equal to 1 in this work) and \(\text{erf}\) is the Gaussian error function. The function \(E(u,v)\) represents the eye filter which describes the spatial frequency dependence response of the human visual system to stimuli [21,22].

The AUC value can be calculated from \(d^*\) according to the equation:
AUC = \frac{1}{\sqrt{2}} \Phi(d'), \hspace{1cm} (9)

where \( \Phi(x) = \int_{-\infty}^{x} \phi(y)dy \) is a Gaussian cumulative probability distribution function and \( \phi(x) = \frac{1}{\sqrt{2 \pi}} e^{-\frac{x^2}{2}} \) is a Gaussian probability density function.

2.1.4. CT dose index: As the basic standard of patient dose assessment in CT scanners, a CT dose index [23], CTDI\textsubscript{PMMA,100} for the cylindrical PMMA phantom is defined as [24]:

\[
\text{CTDI}_{\text{PMMA,100}} = \frac{1}{nT} \int_{-50 \text{ mm}}^{+50 \text{ mm}} D(z)dz. \hspace{1cm} (10)
\]

The integration limit of ±50 mm corresponds to the length of 100 mm of a pencil beam ionization chamber. In practice the measurements are performed for one axial scan with an ionization chamber placed in the center hole of the phantom, measuring a CTDI\textsubscript{PMMA,100,C} value, and for the four holes at the periphery of the phantom, measuring the CTDI\textsubscript{PMMA,100,P} values. The weighted CT dose index, \( C_{W} \), is calculated as [24,25]:

\[
C_{W} = \frac{1}{3} (\text{CTDI}_{\text{PMMA,100,C}} + 2\text{CTDI}_{\text{PMMA,100,P}}). \hspace{1cm} (11)
\]

Due to the fact that in the case of helical scan the radiation dose is inversely proportional to the value of pitch factor (dose \( \propto \frac{1}{\text{pitch}} \)), the CTDI\textsubscript{o} is converted to the so-called volume CT dose index (CTDI\textsubscript{vol}), e.g.:

\[
\text{CTDI}_{\text{vol}} = \frac{\text{CTDI}_{W}}{\text{pitch}}. \hspace{1cm} (12)
\]

2.2. Materials

The Philips medical systems brilliance 16 series performance phantom is used to perform image quality measurements. The images of the phantom's module intended to determine the accuracy of CT numbers for water, acryl, teflon and polyethylene material, are used to obtain TTF on a teflon disk for three consecutive slices. The TTF was determined using imQuest v7.1 software [26]. The spatial frequencies obtained at 50% of TTF values have been reported. The images of the phantom's module intended to determine CT numbers uniformity are used to obtain NPS using the same software. The phantom images have been obtained for nine different scan protocols with the Siemens Somatom Sensation 16 CT scanner. The imaging parameters of these protocols are presented in Table 1.

| Table 1. CT scan protocols |
|----------------------------|
| Protocol's number | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 |
| Tube voltage (kV) | 80 | 80 | 80 | 100 | 100 | 120 | 120 | 120 | 120 |
| Quality Reference (mA) | 80 | 100 | 120 | 120 | 100 | 120 | 80 | 100 | 120 |
| Rotation time (s) | 0.5 | | | | | | | | |
| Beam width (mm) | 16 x 0.75 | | | | | | | | |
| FOV (mm) | 321 | | | | | | | | |
| Pitch | 1.15 | | | | | | | | |
| Reconstruction slice thickness (mm) | 5 | | | | | | | | |
The images are reconstructed using FBP algorithm and two kernels: medium smooth+, B31f, and sharp, B60f.

For obtained results for TTF and 2D NPS, using imQuest v7.1 software [26], the detectability indices \(d'\), and AUC values of NPWE model observer have been determined for the objects of Gaussian contrast-profile of the diameters of 3 mm, 5 mm, 7 mm and 10 mm and with the peak contrasts against background of 10 HU, 20 HU and 40 HU.

The relevant values of CTDI\(_{vol}\) have been calculated for the measurement results obtained using the Adult Body Phantom (32 cm in diameter and 15 cm of length PMMA phantom with five ionization chamber positioning holes; one in the center and four at the periphery), CT Chambers 30009 and DIADOSE E diagnostic dosimeter.

3. Results and discussion

Figure 1 represents the spatial frequency values at 50% MTF value dependence on mAs values, for different voltages and reconstruction kernels, as well as the dependence of CTDI\(_{vol}\) values on mAs values.

From figure 1(a) it can be concluded that the values of spatial frequencies, \(f_{50}\), are not affected by the changes in mAs values but the reconstruction kernels affect significantly the values of \(f_{50}\) in a way that the values of \(f_{50}\) are 2.08 – 2.34 times greater for the kernel B60f than for B31f.

From figure 1(b) the linear relationship between CTDI\(_{vol}\) and mAs is evident, and the slope of the CTDI\(_{vol}\) curve increases with the tube voltage (the slope is 0.036 for 80 kV, 0.052 for 100 kV i 0.068 for 120 kV).

The NPS dependencies on spatial frequency are shown on figure 2 and figure 3 for kernel B31f and B60f, respectively. The peak spatial frequency values of NPS for the kernel B60f are shifted towards higher spatial frequencies in comparison to the kernel B31f (the average peak spatial frequencies are 0.55 mm\(^{-1}\) and 0.29 mm\(^{-1}\) for kernels B60f and B31f, respectively). This suggests the finer texture of images reconstructed with the B60f kernel. Also, these two kernels produce different noise magnitude, figure 4. The noise magnitude values for the kernel B31f are in the range of 21.20 HU to 9.54 HU for the CTDI\(_{vol}\) range values from 3.04 mGy to 10.14 mGy. For the same CTDI\(_{vol}\) range values, in the case of kernel B60f, the noise magnitude values cover the range from 87.26 HU to 39.29 HU.

From figures 2 and 3, it can be concluded that the shape of NPS curve does not change with the tube voltage or mAs values. A significant decrease of the noise magnitude value, figure 4, shown in figures 2 and 3 as a decrease in NPS intensity value, occurs when the tube voltage value was changed from 80 kV to 100 kV. For this change of the tube voltage the noise magnitude was decreased for 6.41 HU, for the kernel B31f. For the same kernel the decrease in the noise magnitude value was equal to 3.12 HU when the tube voltage was changed from 100 kV to 120 kV. Similar has been found for the the kernel B60f, where the noise magnitude value decreased 26.8 HU when the tube voltage changed from 80 kV to 100 kV, and 12.98 HU, when the tube voltage changed from 100 kV to 120 kV. Therefore, regardless of the kernel used, the noise magnitude value was significantly decreased by increasing the tube voltage, with the decrease being more pronounced at lower CTDI\(_{vol}\) values, which is in accordance with the previous study results [27].

\(^1\) In further text, the manufacturer’s reconstruction kernel labels are kept.
Figure 1. Dependence of $f_{50}$ and $\text{CTDI}_{\text{vol}}$ values on mAs.
The dependencies of AUC values on the values of CTDIvol, for selected scan protocols, are shown on figures 5, 6 and 7. From figure 5(a) and 5(b) it can be concluded that the AUC values, for peak contrast value of 10 HU, linearly increase with CTDIvol for each object's diameters and each kernel. For kernel B31f the AUC has the values of 0.560 and 0.670 for object's diameters of 5 mm and 10 mm, respectively. For kernel B60f the AUC has the values of 0.522 and 0.572 for object's diameters of 3 mm and 10 mm, respectively. It can be concluded that, in the case of peak contrast value of 10 HU and the objects of the diameters in the range of 3 mm to 10 mm, the B31f kernel provides higher AUC values.
than the kernel B60f for the same CTDIvol values. Despite that, neither one of this two kernels, for the given CTDIvol values, peak contrast and object's diameters, provide the minimum required AUC value of 0.75 for the object to be considered visible.

For identical scan protocols and object diameters, with the 20 HU peak contrast value, in the same way it can be concluded that the AUC values for the B31f kernel are larger than for the B60f kernel. These values are, for the kernel B31f, equal to 0.610 and 0.810 for the objects of 5 mm and 10 mm diameter, respectively, and for the kernel B60f are equal to 0.5 and 0.640 for the objects of 3 mm and 10 mm diameter, respectively. For the B31f kernel and the peak contrast value of 20 HU, there exist the scan protocols for which the AUC values are larger than 0.75 for the CTDIvol values of 7.16 mGy, 8.97 and 10.14 for the objects of the diameter of 10 mm, 3 mm and 7 mm, respectively.

In the case of the peak contrast value of 40 HU, for the chosen scan protocols, the AUC values for the kernel B31f are in range of 0.710 to 0.96 for the object of the diameters of 5 mm and 10 mm, respectively. For the kernel B60f the AUC has the values of 0.590 and 0.770 for object's diameters of 3 mm and 10 mm, respectively. For the kernel B31f all the scan protocols with CTDIvol value larger than 5.08 mGy provide AUC value greater than 0.75 for all object's diameters and peak contrast value of 40 HU. For kernel B60f and object's diameter of 10 mm, there is only one scan protocol of CTDIvol value of 10.14 mGy which provide an AUC value greater than 0.75.

What can be noticed with both kernels, all the object’s diameters and peak contrast values, is the significant increase in AUC value that occurs when the tube voltage value is changed from 80 kV to 100 kV. As mentioned earlier, a significant decrease in the noise magnitude level with the tube voltage increase can be characterized as the main reason for this increase in AUC value, which is especially pronounced at lower values of CTDIvol.

The additional characteristic related to the kernel B31f is that the AUC values for an object with the diameter of 3 mm are larger than for the objects with the diameter of 7 mm and 10 mm, figures 5 and 6. At the first glance it could be expected that the blurring effect, related to the smooth B31f kernel, significantly reduces visibility of the objects with smaller diameters. However, the effect of the NPS on reducing the visibility of the objects with larger diameters is more significant in this case. Namely, from the shape of the NPS spectrum (figure 2) obtained with the B31f kernel, it is visible that the noise peak

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**Figure 4.** The noise magnitude.
value is shifted towards the lower spatial frequencies in Fourier domain, which corresponds with the larger dimension objects in spatial domain. In other words, NPS significantly reduces the visibility of the objects of larger dimensions. This is not the case for the kernel B60f where the peak values of NPS are displaced towards higher frequencies, reducing the visibility of smaller objects, and in the same time the visibility of larger dimension objects increases due to decreasing NPS values at lower spatial frequency.

![Graphs showing AUC vs CTDIvol for different kernels and object sizes](image)

**Figure 5.** AUC vs CTDIvol.

The reason why the NPWE model – observer, for the case of the kernel B60f, mainly gives AUC values $\leq 0.75$, predicting the poorer observer performances, can be explained by the fact that B60f is a sharp type of the kernel suitable for use in cases where the clinical task assumes the search for the high resolution and contrast objects. This is also confirmed by the spatial frequency values for this kernel at 50% MTF which ranges from $0.7 \text{ mm}^{-1}$ to $0.75 \text{ mm}^{-1}$, while the values for B31f kernel range from $0.33 \text{ mm}^{-1}$ to $0.34 \text{ mm}^{-1}$. Also, the noise magnitude in the range of 39.29 HU to 87.26 HU, for the kernel B60f, as well as the shift of a peak frequency in NPS towards higher frequency values and a small peak contrast of the objects are additional reasons for NPWE model – observer to predict poor observer performances for this kernel. As in this work a clinical task, which assumes the detection of a low contrast object, has been chosen, the smooth kernel B31f produces better results, providing higher observer performances for most selected scan protocols. Even with this kernel and 10 mm object diameter, none of the selected protocols provide an AUC value $\geq 0.75$ of NPWE model – observer for an object with the peak contrast of 10 HU. It means that, for the purpose of solving this clinical task, it is necessary to choose a scan protocols with CTDIvol values greater than 10.14 mGy to additionally reduce noise magnitude and in such a way, increase the AUC value of NPWE model – observer.
Figure 6. AUC vs CTDI\textsubscript{vol}.

4. Conclusions

The presented results have shown that the AUC values, calculated by using the NPWE model observer, can serve as an optimization tool in the process of selecting CT protocols, considering also CTDI\textsubscript{vol} values. For a chosen AUC detectability limit (e.g. AUC ≥ 0.75 or higher) it can be determine the protocol parameters which give the optimal CTDI\textsubscript{vol} value to ensures successful resolution of a set clinical task.

In the case of kernel B31f, for the peak contrast value of 20 HU, CTDI\textsubscript{vol} values, for which the objects of the diameters of 3 mm, 7 mm and 10 mm will be considered visible, are equal to 8.79 mGy, 10.14 mGy and 7.43 mGy, respectively. For a 5 mm diameter object, it is necessary to ensure the CTDI\textsubscript{vol} value greater than 10.14 mGy to achieve the AUC value ≥ 0.75. For the peak contrast value of 40 HU, the CTDI\textsubscript{vol} values of 3.04 mGy, 5.08 mGy, 3.18 mGy and CTDI\textsubscript{vol} < 3.04 mGy will ensure an AUC value ≥ 0.75, for the objects with the diameters of 3 mm, 5 mm, 7 mm and 10 mm, respectively. For the objects of all diameters at the peak contrast value of 10 HU, it is necessary to achieve much higher CTDI\textsubscript{vol} values to make the objects visible (e.g. to achieve AUC value ≥ 0.75).
In the case of the kernel B60f, there is only one protocol with CTDI$_{vol}$ value of 10.14 mGy that ensure AUC value $\geq$ 0.75, and only in the case of the peak contrast value of 40 HU and object’s diameter of 10 mm. Because of this, the B60f kernel is only suitable to be used for those clinical situations in which a task is to detect higher contrast structures.

Using the NPWE model observer approach, the possibility of selection of CT scan protocols is realized that are adjusted to the clinical tasks under consideration. Even more, NPWE model observer allows the exposure parameters, such as the tube voltage and mAs values, to be determine on the basis of projected CTDI$_{vol}$ values to meet any higher AUC value requirements for the given peak value contrasts and object’s diameters. In such a way, an NPWE model observer via AUC values establishes a connection between the physical image quality parameters (MTF and NPS) and the exposure parameters (kV and mAs).

The NPWE model observer represents a comprehensive approach in the optimization of CT scan protocols that takes into account the physical image quality parameters, determined by the exposure parameters and thus the radiation dose, applied kernels, a specific clinical task, defined by the task function, and the impact of the human visual system, and all in order to ultimately predict the performances of a human observer and thus evaluate the entire imaging chain.

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