Software for Modeling Estimated Respiratory Waveform

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Abstract— In the imaging of chest or abdomen, motion artifact is an unavoidable problem. In the radiation treatment, organ movement caused by respiratory motion is a problem unavoidable also to realizing safe and effective cancer treatment preserving healthy tissue. In this article, we compare two modalities 3D CT and 4D CT and present the main difference between them, which is compensating the breathing motion. However, we record a real breathing signal using ANZAI belt, analyze the resulting signal and simulate the estimated respiratory waveform based on three different models.

Keywords— CT parameters, respiratory signal, 4DCT, motion artifact.

I. Introduction

SIMULATORS and models of the respiratory system range from simple mechanical devices to complex systems. These systems have the considerable utility in the clinical physician education, the leading treatment methods, evaluating new devices and methods, and in improving of our understanding of cardiorespiratory system. Simulators and models include 3 types: physiologic models, anatomic models, signs-and-symptoms simulators [1].

However, all the simulators and models types base on highly accurate mathematical models. In this article we describe developed software for modeling respiratory waveform which computes waveform by 3 different methods.

In radiotherapy, this is a very important part of the treatment because we may simulate clinical scenarios, from small deviations to disastrous emergency situations can be simulated. Increasing the importance of reducing the delivered dose to patients leads to increase the complexity of CT imaging technology and increase the importance of CT scanning parameters to create lower dose. In our report, we present and discuss the influence of CT parameters such as voltage, current, and reconstruction kernel on the images.

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II. Method and Material

A. Experiment Setup

We have a phantom of pelvic region, which is scanned several times with different scanning parameters (voltage, current- time product, field of view (manual), and reconstruction kernel). We performed three tasks, which are varying voltage and current-time-product is fixed, varying current-time-product and voltage is fixed, and varying the reconstruction kernels with same value of voltage and current-time-product. In the first task, scanning is performed using four voltage values 80 kV, 100 kV, 120 kV and 140kV, and the current-time-product fixed to 150 mAs. In the second task, scanning is performed using three current-time-product values as follows 50 mAs, 100 mAs and 250 mAs, and the voltage is fixed to 120 kV. In the third task, we change the reconstruction kernels, which are provided by the reconstruction software and determine the image sharpness.

B. Influents of CT Parameters

We perform several CT scans of pelvic phantom. During the experiment, we plan to change the parameters of the CT to see the influence of the parameters on the resulting images. The main parameters are the tube voltage, the tube current, and the reconstruction kernel.

X-ray tube potential indicates the peak energy of the x-ray photons (in kilovolts) in a spectrum of x-ray energies. We change the voltage of the X-ray tube in the range of 80 to140 kV. Radiographers can change voltage settings on the X-ray machine in order to manipulate the properties of the X-ray beam produced. We need various intensity and energy levels of X-ray to scan different part of body. The increase in X-ray tube voltage increases the average photon energy (i.e., increased penetration).The goal is to investigate the effect tube voltage on image quality, radiation dose, and contrast.

Tube current-time-product (mAs) is the product of the X-ray tube current (in mAs) and the CT scanner exposure time per rotation (in seconds). In general, increasing tube current or tube current-time-product results in a proportional increase in radiation dose as tube current-time product is proportional to X-ray intensity. For example, if other parameters are held constant, increasing the tube current-time product from 100 to 200 mAs will double the photon output and hence double the exposure to the patient.

Image reconstruction in CT is a mathematical process that generates tomographic images from X-ray projection data acquired at many different angles around the patient. Image reconstruction has fundamental impacts on image quality and therefore on radiation dose. For a given radiation dose it is desirable to reconstruct images with the lowest possible noise without sacrificing image accuracy and spatial resolution. Reconstructions that improve image quality can be translated into a reduction of radiation dose because images of the same quality can be reconstructed at lower dose. Reconstruction kernel should be based on specific clinical applications. For example, smooth kernels reduce image noise and enhance low contrast, whereas sharper kernels are used to change the sharpness of edges within the image.

Field of view determines how much anatomy is scanned. We change field of view to decrease the diameter of the area being scanned. The smaller the diameter is, the smaller the delivered radiation to the phantom and the smaller the dimensions of the reconstructed images.

C. 3DCT of Stationary and Moving Phantom

We use respiratory phantom, which can incorporate stimulated motion, with low rate (10rpm) or high rate (15rpm) as we can see in Fig. 1.

Fig. 1 Respiratory phantom used in experiment

Two tasks are arranged, stationary phantom and moving phantom. We use identical scan parameters for the two experiments (120 kV, 100 mAs). We use 3DCT to scan stationary phantom, low rate respiratory phantom and high rate respiratory phantom, to get a direct presentation of motion artifacts.

D. 4DCT for Compensating Motion Artifact

We use AZ-733V respiratory system to support the 3DCT system and this combination called 4DCT. This combination will compensate motion artifact.

We choose 0%, 15%, 50%, 85%, 100% of inhale phase and 15%, 50%, 85% of exhale phase to classify the images based on the respective phase and value.

E. ANZAI Belt's Measurement

ANZAI system is breathing monitoring systems. It is used to detect breathing signal and this signal is used as reference to compensate organ motion, and result in precise radiation treatment. The basic idea of combining respiratory signal is called respiratory gating, which is using a pressure-sensor with motion monitoring system. The goal of this signal in radiation treatment is minimizing the area of treated target tumor which moves due to patient respiration. Respiratory signal is acquired for 10 minutes from ANZAI belt around the abdomen. The recorded signal suffers from several types of artifacts such as noise, baseline-drift, saturation, etc. Fig.2 shows a respiratory signal of 7 seconds. In this figure we can notice the parameters of the signal, which they are respiratory rate, inhalation phase and exhalation phase. Fig.3 shows the signal with baseline-drift artifact. We remove it using a polynomial function that we fit to the signal to overcome this effect.

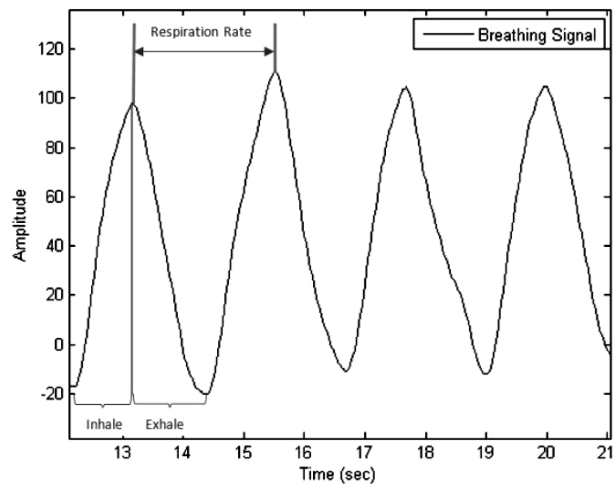


Fig. 2 Respiratory signal recorded from ANZAI belt

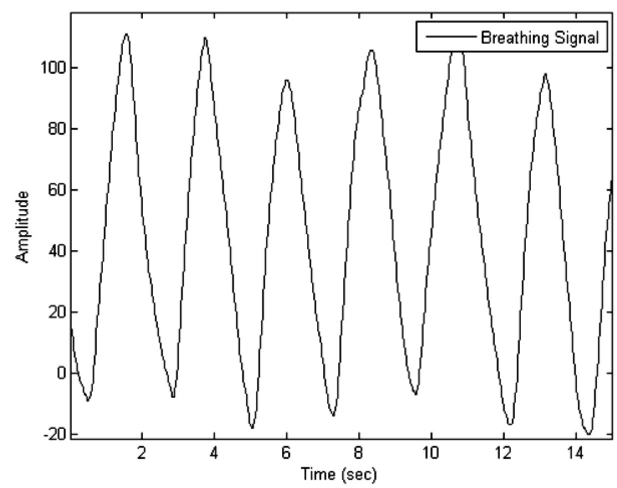


Fig. 3 Respiratory signal of 15 sec of normal breathing with baseline-drift

III. Result and Discussion

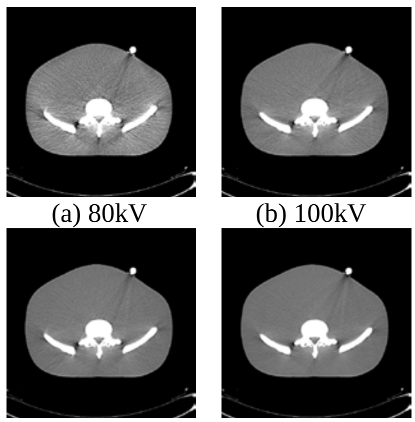
A. Influence of CT Parameters

The intensity of the radiation dose would facilitate accurate comparisons of radiation doses used for different tube voltages, for example, a 14% decrease in tube voltage from

140 to 120 kV will reduce patient exposure and decrease radiation dose by up to 35%. Results of our report shows that it is possible to reduce radiation exposure substantially by decreasing the tube voltage from 140 kV to 80 kV but the noise level is the lowest at 140 kV and the highest at 80 kV. The disadvantage of lowering tube voltage, however, is increased image noise, which can degrade image quality.

The image is taken at 80 kV voltage in Fig.5a is more noisy than the image is taken at 140 kV voltage Fig.5d. We have to use higher energy of radiation to diagnose patients with high weights. When we increase current-time the noisiness of the image decreases. The image is taken at 50 mAs current in Fig.5a is noisier than the image taken at 250 mAs current in Fig.5c. For soft tissue, it use small value of the current- time (which means reduce the dose), but for high density tissue, they use large value of the current. The TABLE 1 presents the results of calculating the PSNR of different voltage values (the maximum PSNR value is the value of the image taken at 140 kV). The TABLE 2 presents the results of calculating the PSNR of different current-time values (the maximum PSNR is the value of image taken at 250 mAs). For calculating the PSNR values, we took the image at 120 kV and 150 mAs as reference image for calculating the PSNR of different voltage and current values.

Reconstruction kernel (filter or algorithm) has a significant impact on spatial frequency and noise characteristics of an image. Smooth kernels reduce high spatial-frequency information and image noise. Sharp kernels increase high spatial-frequency information and image noise. In Fig.6, If the value of kernel has relatively low number the image is"smoother" (Fig.6a-6c), but if value of kernel has relatively high number the image is"sharper" (Fig.6d-6f). Therefore, for the visualization of soft tissues using a lower kernel number (20-40) is recommended. To visualize tissues (bones, lung tissue), a higher kernel number (40-70) provides high resolving power.



(a) 80kV (b) 100kV

(c) 120kV (d) 140kV

Fig. 4 Influence of different voltage values on noise level in CT images

(a)50mAs (b) 100mAs (c) 250mAs

Fig. 5 Influence of different voltage values on noise level in CT images

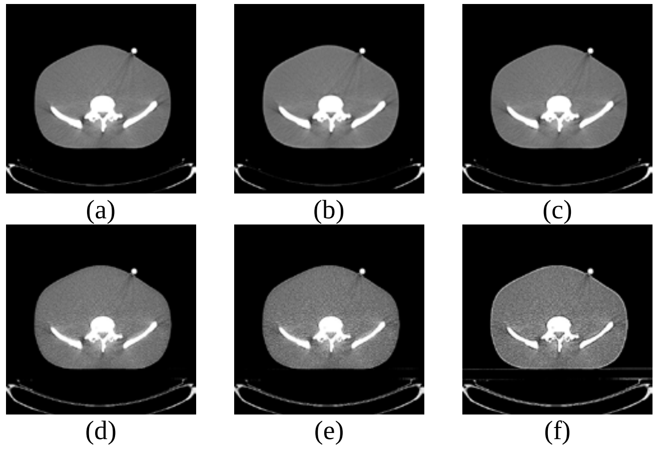


Fig. 6 Influence of different kernel number (from very smooth to very sharp)

Table 1. PSNR for Different Voltage Values

Voltage / kV	80	100	140
PSNR, dB	19.5719	23.4080	24.3381

Table 2. PSNR for Different Current-Time Values

Current / mAs	50	100	250
PSNR, dB	15.3829	15.6406	15.7558

A. Analysis of ANZAI Belt's Measurements

Baseline-drift is the short time variation of the baseline from a straight line caused by electric signal fluctuations. There are several ways to remove baseline-drift such as a linear approximation, a cubic spline interpolated approximation, and a recurrent neural network approach mimicking an adaptive filter, and the final method involved calculating the first and second derivatives of the signal in order to attenuate the baseline drift.

To remove the baseline-drift, we fit a polynomial to the data. The algorithm is composed of three steps. It is calculating the coefficient for a polynomial $p(x)$ of degree n that is a best fit (in a least-squares sense) for the data. The coefficients for a polynomial $p(x)$ of degree 5 that is a best fit for the data. Equation (1) represents polynomial of five degree fitted to the respiratory signal to remove the baseline-drift.

$$p(x) = p_1x^5 + p_2x^4 + p_3x^3 + p_4x^2 + p_5x + p_6 \quad (1)$$

Table 3 shows the polynomial's parameters, which perform best fit to the respiratory signal. Fig. 3 shows 15 seconds of the respiratory signal, which suffers from baseline-drift. Fig. 7 shows the signal after removing baseline-drift using polynomial fitting to the data. Baseline-drift is common effect for all biological recorded signals.

Fig. 8 shows the difference between the respiratory signal with baseline-drift and without it.

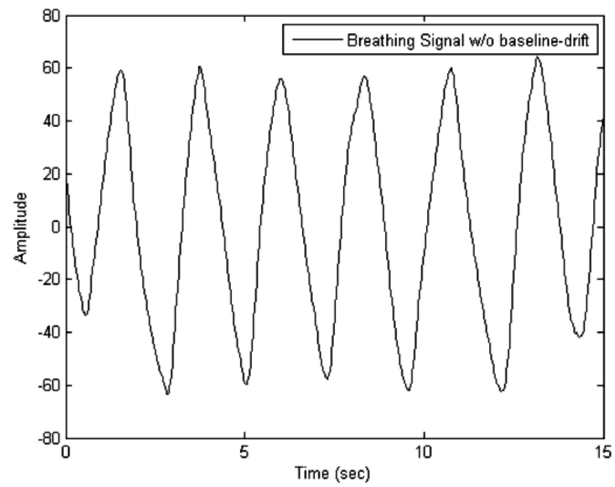


Fig. 7 Respiratory signal after removing the baseline-drift artifact

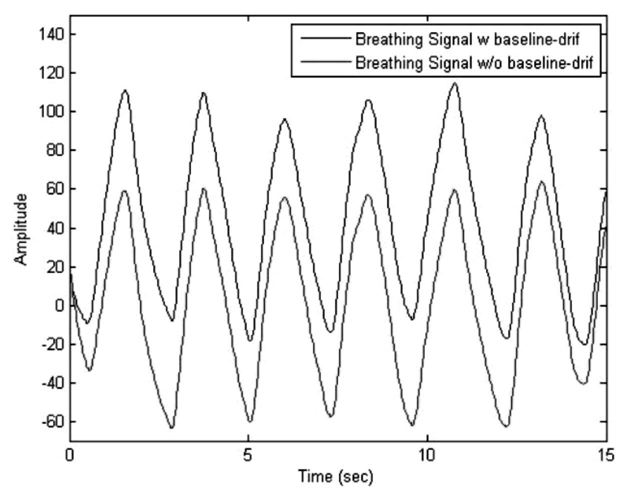


Fig.8 The difference between the signal with and the signal without baseline-drift artifact

Table 3. Parameters of the Polynomial Function

p_1	p_2	p_3	p_4	p_5	p_6
10.239	-9.385	-37.872	15.816	23.121	44.593

Fitting sine function to the respiratory signal. The fitting algorithm is working based on fitting sine wave to the breathing signal by taking the maximum amplitude value and the minimum value of the signal. The difference between the min and max values will be used as peak to peak amplitude. Next step is computing zero-crossing and estimating the period and the offset. The fitting function is calculated from sine function as we can see in equation (2). The last step is to fit the sine signal to the respiratory signal by calculating the least-square cost function and minimizing it as in Fig. 9. The parameters of the fitted function are listed in Table 4. We apply the fitting function after removing the baseline-drift artifact.

$$b(1)\sin\left(\frac{2\pi x}{b(2)}+\frac{2\pi}{b(3)}\right)+b(4) \tag{2}$$

Table 4. Parameters of the Fitting Function

$b(1)$	$b(2)$	$b(3)$	$b(4)$
50.876	2.327	-0.734	0.000

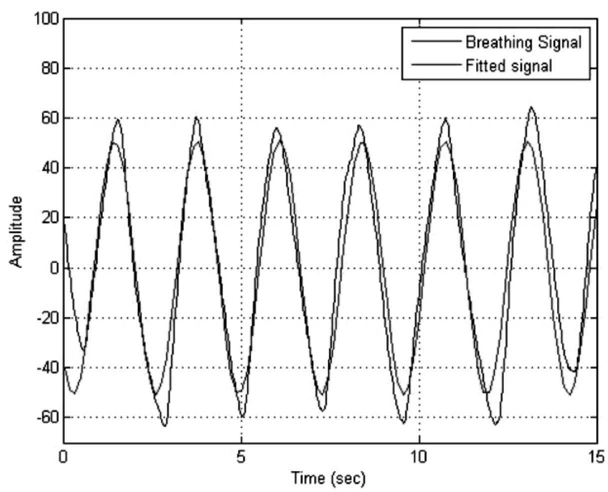


Fig.9 Respiratory Signal (Black) and Sine Wave Fitted to the Respiratory Signal (Grey)

Respiratory Cycle Modeling (Lujan model). It is generally assumed that all the points in the volume reach their final position at the same time and that the temporal behavior along the trajectory is determined by a 1-D breathing signal. Several models of breathing cycles have been proposed in the literature. Lujan et al. model models the dynamic breathing volume curve [3, 4]. It is based on a periodic but asymmetric function (more time spent at exhalation versus inhalation).

In (3), V_0 is the volume at exhalation, corresponds to the tidal volume (TV) which is the amount of air breathed in or out during normal respiration, $V_0 + b$ is the volume at inhalation, τ is the period of the breathing cycle, n is a parameter that determines the general shape (steepness or flatness) of the model, and is the starting phase of the breathing cycle in Fig. 10. This model represents a priori knowledge of a conventional breathing cycle.

$$V(t) = V_0 + b\cos^{2n}\left(\frac{\pi}{\tau}t+\phi\right) \tag{3}$$

Fig. 11 shows respiratory signal modeled by Lujan et. al. model as in equation (3). n is a parameter that determines the general shape (steepness or flatness) of the model and after the fitting is equal to 0.721. V_0 is the volume at exhalation and after the fitting is equal to 49.556. φ is the starting phase of the breathing cycle and we get -1.207. τ is the period of the breathing cycle and we get 2.327 after the fitting.

In Table 5 the result after fitting Lujan model to our respiratory signal. After fitting the signals to two different functions, sine function and Lujan model, we find that both signals are suitable to represent the respiratory signal. As we can notice from Table 2 and Table 3 that the results of the both fitting function are almost the same. The most important step for the both fitting function is setting the start point, which plays a big role of initialization the fitting function.

Respiratory rate. Breaths per minute is called the ventilation rate. It is 25 breathing cycle in one minute in our experiment. The normal value is based on the normal range and it is for adult 30-60 Breaths per minute.

Table 5. Lujan Model Parameters

b	τ	ϕ	n	V_0
49.556	2.327	-1.207	0.721	0.000

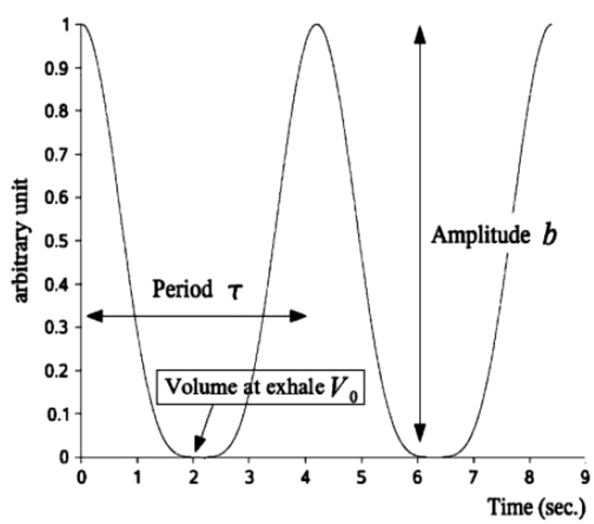


Fig.10 Breathing Cycle Modeling Proposed by Lujan et. al. (n = 2) [3]

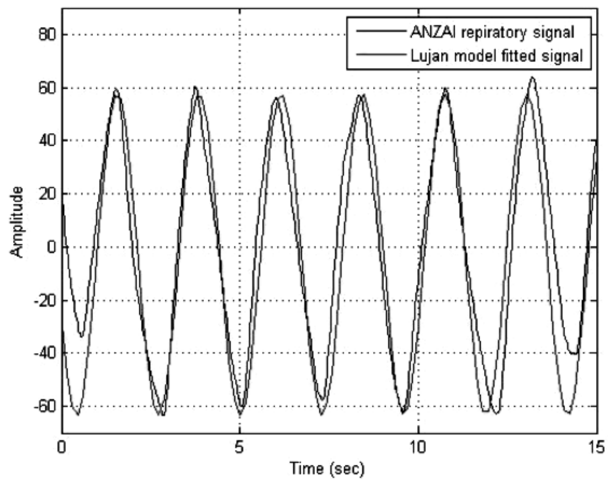


Fig.11 ANZAI Respiratory Signal (Black), Lujan et. al. Model (Fitted Function to the Respiratory Signal) (Grey) (n = 0.7215)

IV. CONCLUSION

We've concluded our results as the following: with a reduction of the tube voltage from 140 kV to 80 kV at abdominal CT, the radiation dose can be reduced but the noise will increase. Although decreasing tube current is the most means of reducing CT radiation dose. This also reduces the contrast- to-noise ratio, which may affect the diagnostic outcome of the examination. Detailed understanding of the basic CT scan parameters is essential, and knowledge of how to manipulate these parameters to produce diagnostic images at lower doses is critical for safe imaging CT scan parameters that can be altered or optimized to reduce patient radiation dose. Although there is always a trade-off between image quality or noise and patient radiation dose, in many cases, a reasoned manipulation of these parameters can allow the safer imaging of patients (with lower dose) while preserving diagnostic image quality.

4D CT provides solution for breathing motion effect by combining the breathing signal of a sensor to 3D CT to compensate the motion effect. In this article, we performed the software for analyzing a respiratory signal and calculation the estimated respiratory waveform by mathematical methods which was shown before. All the methods describe respiratory signal with relatively same accuracy.

To acquiring a respiratory signal our program use ANZAI belt, respiratory signal data upload to program in txt format. The program output display as estimated respiratory signal in txt file and plots calculated by three methods. Thus, performed program is the first step to developing device which consists software and hardware.

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