A semi-automatic method for peak and valley detection in free-breathing respiratory waveforms

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The existing commercial software often inadequately determines respiratory peaks for patients in respiration correlated computed tomography. A semi-automatic method was developed for peak and valley detection in free-breathing respiratory waveforms. First the waveform is separated into breath cycles by identifying intercepts of a moving average curve with the inspiration and expiration branches of the waveform. Peaks and valleys were then defined, respectively, as the maximum and minimum between pairs of alternating inspiration and expiration intercepts. Finally, automatic corrections and manual user interventions were employed. On average for each of the 20 patients, 99% of 307 peaks and valleys were automatically detected in 2.8 s. This method was robust for bellows waveforms with large variations. © *2006 American Association of Physicists in Medicine*. [DOI: 10.1118/1.2348764]

Key words: peak detection, respiratory waveform, 4D CT, gated radiotherapy

I. INTRODUCTION

Respiration monitoring and the determination of the extreme of inspiration (peak) and expiration (valley) are important for many cardiopulmonary and neurological studies.[1–3] In gated radiotherapy and respiration correlated four dimensional (4D) computed tomography (CT), the same information is needed to identify the respiratory phases for gating the radiation beam and sorting the CT images, respectively.[4–7] It was reported that the commercial real-time processing and post-processing software (RPM, Varian Medical Systems) often inadequately assigned respiratory phases for 4D patients.[8,9] Manual selection of all extremes was proposed to improve the 4D CT reconstruction.[8] Respiratory monitoring sessions may extend from a few minutes to several hours.[1–3,7,9] Since, manual selection is time consuming, a computerized method is desirable.

Peak detection methods using amplitude threshold(s) are limited to signals that have small variations in both the peak amplitude and baseline level.[2,3] To overcome these limitations, methods were proposed by using the first (airflow) and second derivatives of the waveform.[1] However, the accuracy was affected by the often abundant noise spikes in the derivatives. LABVIEW (National Instruments) provides a peak detection procedure by fitting a quadratic curve to successive groups of data points, and then testing whether a point met the criteria for a peak.[10] The results were found to be sensitive to the number of data points used in the curve fitting, and the identified peak locations were not accurate.

II. MATERIALS AND METHODS

A. Acquisition of respiratory waveforms

Respiratory waveforms from 20 patients with thoracic or upper abdominal malignancies were acquired in our 4D CT studies using amplitude sorting.[6,7] The patients were asked to breathe quietly and naturally during the ~10 min sessions. The respiratory waveform was measured by a digital voltage signal from a differential pressure sensor wrapped around the patient's abdomen ("bellows").[7] Software developed on LABVIEW was used to acquire the bellows signal (time and voltage) at a sampling frequency (f_s) of 100 Hz.

B. Peak and valley detection algorithm

1. Estimation of the respiratory period by fast Fourier transform

The respiratory period T was estimated by applying a fast Fourier transform (FFT) (Ref. 11) to the first 15 s of the respiratory waveform. The choice of 15 s was to ensure that at least one breathing cycle was included. To suppress the dc component in the power spectrum, the mean signal amplitude was subtracted from the waveform. The peak location in the power spectrum provided an estimation for the respiratory frequency f, and consequently the period $T(T=1/f)$.

2. Calculation of a moving average curve

A moving average curve (MAC) is calculated at every time point t as

$$\mathrm{MAC}(t)=\begin{cases}\overline{x(\tau)}\big|_0^{2T}, & \text{if } 0\le t\le T\\ \overline{x(\tau)}\big|_{t-T}^{t+T}, & \text{if } T<t\le L-T\\ \overline{x(\tau)}\big|_{L-2T}^{L}, & \text{if } L-T<t\le L,\end{cases} \tag{1}$$

where x is the respiratory signal, L is its length, τ is time, and $\overline{x(\tau)}\big|_{t_1}^{t_2}$ is the average value of x during $[t_1,t_2]$. The MAC appears as a "center" line that intercepts each breath twice; once on the inspiration branch, and another on the expiration branch (Fig. 1). In this way, the effects of baseline variation are removed. The window width for averaging was experimentally chosen to be $2T+1/f_s\approx 2T$.

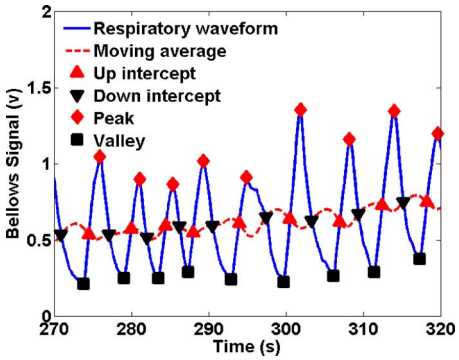

FIG. 1. Detection of peaks and valleys using intercepts of the MAC with the respiratory waveform. The MAC closely follows the trend in the respiratory waveform. Peaks and valleys are respectively determined by the maximum and minimum between pairs of alternating up intercept and down intercept.

3. Determination of peaks and valleys using intercept pairs

The intercepts of the MAC with the respiratory waveform were identified if

$$\{x(t-1) \le \mathrm{MAC}(t-1)\} \cap \{x(t) \ge \mathrm{MAC}(t)\}, \quad (2)$$

$$\{x(t-1) \ge \mathrm{MAC}(t-1)\} \cap \{x(t) \le \mathrm{MAC}(t)\}. \quad (3)$$

Equation (2) represents "up intercepts" on the inspiration branch, while Eq. (3) represents "down intercepts" on the expiration branch (Fig. 1). Ideally, there should be exactly one up intercept and one down intercept for each breath cycle, yielding alternating up/down intercepts. A peak was then defined as the maximum between an up intercept and the following down intercept, while a valley as the minimum between a down intercept and the following up intercept. To make the process robust against noise and short variations or short breath hold within a real breath, when an intercept was within $T/20$ of its previous one, it was disregarded.

Figure 2 shows two issues of the above procedures for irregular breaths. First, there are two consecutive intercepts with the same labels [Fig. 2(a)]. This was automatically detected, and the first intercept was eliminated [Fig. 2(b)]. The choice of deleting the first rather than the second intercept was arbitrary and might need correction as described below. If there were more than two consecutive intercepts with the same labels, only the last one would be kept. Secondly, an inspiration was too small [Fig. 2(b)] and therefore the associated pair of peak and valley was deleted [Fig. 2(c)]. We considered an inspiration or expiration invalid if its amplitude was smaller than 20% of the mean peak to valley amplitude (V_pv). This caused some local errors but avoided the much greater between-breath inconsistency. Consequently, the 4D CT reconstruction did not (and should not) model those small abnormal breaths. If there were consecutive invalid inspirations and expirations, and the total number of associated extremes was even, all of these extremes were deleted. Otherwise, only the maximum (minimum) among these extremes was kept as a peak (valley) if the number of peaks (valleys) was larger (by 1).

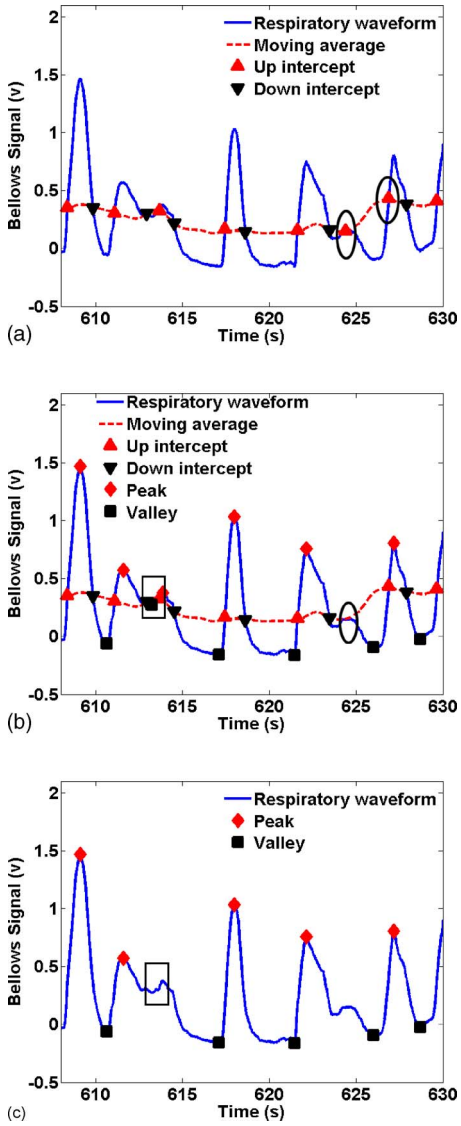

FIG. 2. Illustration of two issues for irregular breaths: (a) There are two consecutive up intercepts (circled); (b) The first intercept is automatically eliminated (circled). There is a small inspiration (rectangle). (c) The peak and valley of the small inspiration are automatically deleted (rectangle).

4. User intervention

Because of the tremendous variations in biological data, such as in respiratory waveforms, user reviewing/intervention is usually necessary to verify/adjust the results of a computerized method. A user interface programmed in MATLAB (The Mathworks) was provided for interactively reviewing the results, and adding or deleting extremes, one at each time.

5. Self-check

Finally, a self-check procedure was employed to verify that each peak (valley) was correctly identified as the maximum (minimum) point between its two neighboring valleys (peaks). Otherwise, the error location was reported to the user to prompt another intervention.

III. RESULTS

The respiratory waveforms demonstrated moderate to large variations. The mean V_pv of the bellows signal was 0.8 V, with a standard deviation (s.d.) of 0.2 V. The mean respiratory period was 4.1 s, s.d. = 0.9 s. The baseline level (valleys amplitude) had a mean of 0.1 V, s.d. = 0.1 V. The respiratory periods estimated by the FFT were close to the actual values with an average absolute difference of 6.1%.

The method was implemented in MATLAB on a personal computer (PC). On average, a respiratory waveform was 575.3 s long, and there were 307 peaks and valleys. For each patient, on average, 99% of all extremes were correctly located (per user's judgment) by the automatic algorithm in 2.8 s. Only three (1%) extremes, on average, required manual user adjustment. A user spent 66.8 s on average for reviewing, and manually adding or deleting extremes. For 9 of the 20 patients, all extremes were automatically detected. We estimated that a user would spend about 2 s for each extreme by using the manual selection method. The high efficiency of the semi-automatic algorithm is clear.

IV. DISCUSSION AND CONCLUSION

The moving average curve closely followed the baseline variation (Figs. 1 and 2), so that the algorithm was not affected by possible signal drift. The algorithm did not use derivatives, and was therefore much less affected by noise. Nonetheless, the MAC would break down if the breathing amplitude changes dramatically. The automatic corrections improved the efficiency and reliability of the method. In summary, we have presented a general, straightforward, and reliable method for peak and valley detection in free-breathing respiratory waveforms.

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