

Real-Time Detection of Sleep Breathing Disorders

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Abstract

Diagnosis of sleep-related breathing disorders (SBD) usually relies on manual retrospective analysis of the signals recorded during a whole night. We evaluated a novel detector, implemented in a cardio-respiratory Holter device, and capable of automatic event-based detection of SBDs in real time.

Events (hypopneas and apneas) detected in real time were compared to those scored on the simultaneously recorded gold standard polysomnography (PSG).

4240 events were recorded by the PSG in 30 severe obstructive SBD patients. The sensitivity and positive predictive value of the detector were 86.2% (C.I. 85.2-87.2%) and 60.7% (C.I. 59.5-61.9%) respectively.

The performance of this novel detector suggests that it could be used to trigger and/or adjust an event-based SBD treatment.

1. Introduction

Sleep-related breathing disorders (SBD) are an under-diagnosed health problem associated with increased cardiovascular morbidity and mortality [1]. The gold-standard method for diagnosis of SBD is in-lab polysomnography (PSG), a comprehensive multi-channels monitoring of sleep and cardio-respiratory signals. The complete recording is scored manually, offline, by a trained specialist, according to international rules that define the main respiratory events as apneas - cessations of flow for at least 10 seconds - and hypopneas - reductions in flow amplitude ended by either an oxygen desaturation $\geq 3\%$ or an arousal and lasting more than 10 seconds [2]. The number of apneas and hypopneas per hour is the Apnea + Hypopnea Index (AHI), quantifying the severity of the SBD and being a crucial information for patient therapeutic decision.

Currently, the first line therapy is CPAP (continuous positive airway pressure), which re-opens the upper

airway and is proved efficient, even though associated with a rate of 15% initial refusal and low adherence in the long term for 20-30% of the sleep apnea patients [3]. There are perspectives to develop tailored personalized therapies in response to the occurrence of individual apneas or hypopneas, instead of a continuously applying PAP during the whole night. This approach would require detection of breathing disorders in real-time, as opposed to the current scoring that critically relies on the consequent oxygen desaturation or micro-arousal that occur well after the end of hypopneas.

The present study reports the performance of a novel detector that processes instantaneous nasal airflow data acquired with a cardio-respiratory Holter (Spider SAS, Sorin CRM, France).

2. Real-time detector

The detector functions in real-time. Hence, a decision is made for each incoming nasal pressure sample, whether breathing is normal or disturbed. The output of the detector is therefore a scoring for each data-sample: either 0 (normal breathing), or 1 (apnea) or 2 (hypopnea). The scoring is the result of a series of processing and decision steps depicted in Figure 1 and described in details in the following sections.

2.1. Pre-processing

Pre-processing of the nasal pressure signal consists in two sequential steps: causal low-pass filtering (typical cut-off frequency: 2.5 Hz) and offset subtraction (offset determined by an exponential moving average). The former filters out snoring, movement artifacts and other high-frequency noise. The latter centers the signal around zero so that inhalation is positive and exhalation is negative.

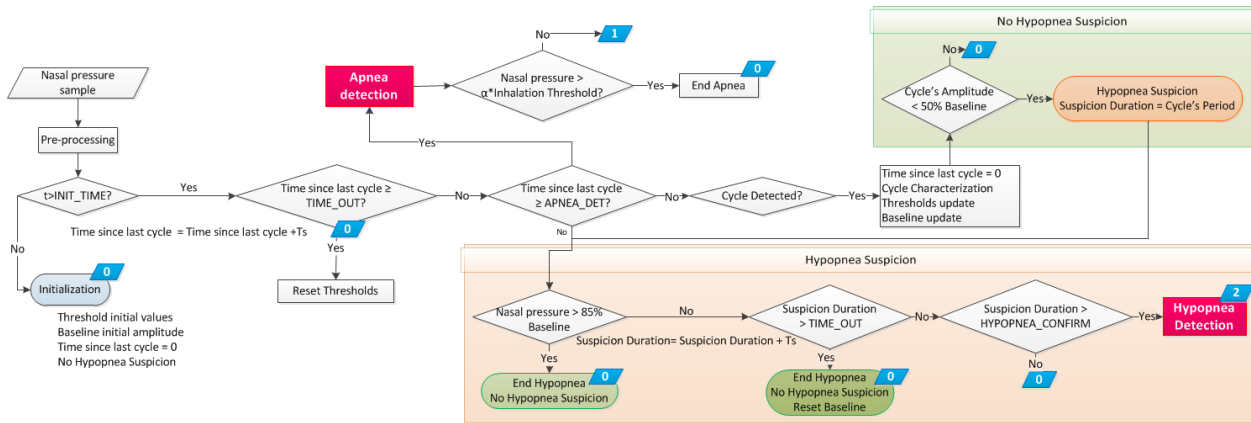


Figure 1: Real-time SBD detector. The input is nasal pressure signal sampled at a frequency of $1/T_s$ (typically 8-200 Hz). The detector output is a signal at the same sampling frequency with values 0, 1 or 2 for normal breathing, apnea and hypopnea respectively (values in the blue trapeziums). Squared box, diamond box, circled box are a process, a decision or a stop step respectively (details in section 2). Parameters' values are given in Table 1.

2.2. Respiratory cycle detection

Respiratory cycle detection is based on two independent auto-adapting thresholds: a positive threshold for inhalation and a negative threshold for exhalation (Figure 2B). More precisely, a cycle is detected when the signal increases and exceeds the positive threshold. It must then be followed by exhalation, which is confirmed when the signal decreases and falls below the negative threshold. Then, the next cycle can be detected once the signal increases again and exceeds the positive threshold and so on (Figure 2B).

2.3. Threshold and baseline amplitude update

Once a cycle is detected, it is characterized for its amplitude (peak inhalation value) and period (Figure 2B). The cycle's amplitude and period must abide by strict criteria to be classified as *valid* to update the thresholds and breathing baseline amplitude. If this is the case, then the positive (negative) threshold is updated to $1/10^{\text{th}}$ of the average peak inhalation (exhalation) of the last 10 *valid* cycles. Similarly, the baseline amplitude is updated to the median value of the last 10 *valid* cycles' amplitudes. Hyperventilated (hypoventilated) cycles that follow (precede) apneas/hypopneas are discarded from the *valid* cycles (see stable baseline amplitude during hypopnea detection in Figure 2D). Thresholds and baseline amplitude are initialized using the maximum peak inhalation and exhalation of the signal during the first 120 seconds (TIME_INIT in Figure 1 and Table 1).

2.4. Apnea detection

An apnea is detected when no breathing cycle is detected for a prolonged period of time. Here, detection of a respiratory cycle starts a timer (Time since last cycle,

Figure 1). If this timer exceeds a programmable value (APNEA_DET), typically 8 seconds, an apnea is detected. This timer is reset for each new detected cycle.

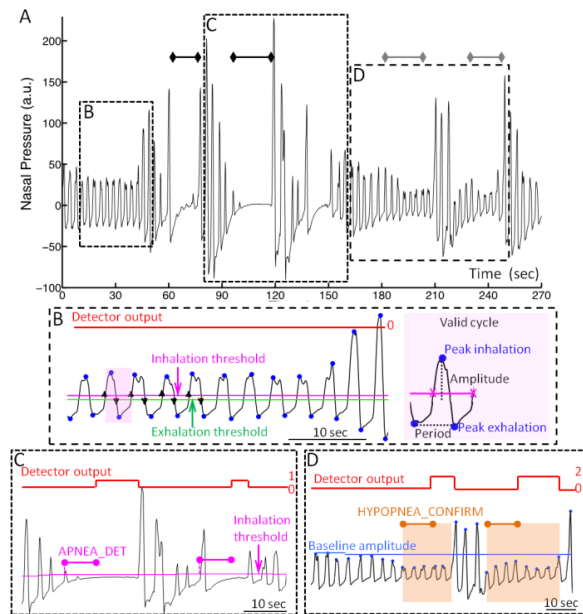


Figure 2: A: Nasal pressure after pre-processing. Arrows indicate PSG scoring: 2 apneas (black) and 2 hypopneas (grey). B: cycle detection on the first 30 seconds of signal shown in A. The detector outputs zeroes for each data points as respiration is normal. C: Apnea detection when the timer triggered by a cycle detection reaches 8 seconds (programmable APNEA_DET). The detector yields ones for apneas. D: Hypopnea detection when suspicion (orange block) duration exceeds 10 seconds (programmable HYPOPNEA_CONFIRM). The detectors outputs twos. Suspicion starts once a cycle's amplitude falls below 50% of the baseline amplitude.

Normal breathing resumes when the inhalation is

sufficient again, i.e. largely above the inhalation threshold (factor α in Figure 1 and Table 1). Note the hysteresis on the magnitude of inhalation required to start and to end an apnea respectively. This ensures sufficient sensitivity and precocity of the detection, while small baseline fluctuations due to respiratory efforts or signal noise during the apnea do not mislead the detector.

Furthermore, an apnea cannot exceed 120 seconds (TIME_OUT). Such apneas are erroneous detections due to wrong threshold values or absence of signal due to cannula removal or in-excess mouth breathing.

2.5. Hypopnea detection

Hypopneas are detected in two steps: a suspicion followed by a confirmation phase (Figures 1 and 2D). The suspicion phase starts once a respiratory cycle's amplitude falls below 50% of the baseline amplitude. A timer (Suspicion Duration) starts with this cycle and is incremented with every new sample. If the timer exceeds 10 seconds (programmable HYPOPNEA_CONFIRM), then a hypopnea is confirmed and detected. This timer is reset when the magnitude of breathing returns to 85% of the baseline amplitude.

Note that during hypopnea suspicion, an apnea can be detected. Similarly to the case of apnea, a hypopnea suspicion is limited to 120 seconds (TIME_OUT).

Table 1: Detector parameters as per Figure 1. Values applied on the validation set are in the third column.

Parameter	Range	Applied
INIT_TIME (sec)	40 - 300	120
TIME_OUT (sec)	90 - 180	120
APNEA_DET (sec)	4 - 12	8
HYPOPNEA_CONFIRM (sec)	5 - 12	10
α	>1 - 8	3

3. Evaluation of the real-time detector

3.1. Clinical study

The real-time detector was tested prospectively in 43 severe obstructive sleep apnea (OSA) patients (mean age: 58.5, BMI: 30.1 kg/m², AHI: 47.7 /h, 75% male). All patients underwent a full PSG. Simultaneously, nasal pressure (NP) was acquired at 200 Hz with the SpiderSAS (Sorin CRM, France). Using a T-deviation system, the NP signals were the same for both SpiderSAS and PSG. The Holter NP signal was transmitted every 100 ms via Bluetooth to a laptop that performed, in real-time, the SBD detection for each incoming signal sample as described above. Central reading of PSG was done by an expert, blinded core-lab (CHU Grenoble, France) who provided the overall AHI together with an accurate scoring of individual events (start, duration and type of

event).

In one patient, the Holter did not start properly. A first set of 12 patients (training set) was used for the definition of the detector parameters given in Table 1 (applied values). The performances of the real-time detector were assessed on the remaining 30 OSA patients (validation set).

3.2. Data analysis

The detector was assessed using three approaches.

First, a Respiratory Disturbance Index (RDI) was calculated as $RDI (/h) = (\sum \text{apneas} + \sum \text{hypopneas}) / \text{total detection time (in hrs)}$ and compared to the AHI found by PSG. For calculation of the RDI, both hypopneas and apneas were detected retrospectively at 10 seconds to abide by conventional definitions [2]. The relationship between RDI and AHI was investigated by correlation analysis (Pearson's correlation product). Bland and Altman plots were used to compare the measurement techniques. Bias and variability in the index were evaluated by calculating the mean difference together with its 95% confidence interval.

Second, PSG-annotated events were matched with the real-time detector output, after time-alignment of the NP signals from both modalities. A true positive (TP) event was defined as an event detected less than 10 seconds before or within the duration of a PSG-annotated event. Apneas detected as hypopneas (and reciprocally) were regarded as true positives. The detector sensitivity (SE) and positive predictive value (PPV) were calculated as:

- $SE (\%) = TP / (TP + FN) \times 100$
- $PPV (\%) = TP / (TP + FP) \times 100$

The 95% Confidence Intervals of the sensitivity and the PPV were calculated using the Wilson score.

Third, PSG-annotated apneas and hypopneas were converted to a digital scoring at the same sampling frequency as the incoming signal. A TP was defined as a sample scored as SBD by PSG and detected as hypopnea or apnea by the real-time detector. SE and PPV were calculated as above and specificity (SP) and negative predictive value (NPV) were additionally calculated, as:

- $SP (\%) = TN / (TN + FP) \times 100$
- $NPV (\%) = TN / (TN + FN) \times 100$

4. Performances of the real-time detector

The relationship between RDI and AHI was highly significant ($r = 0.96$, $p < 0.001$) (Figure 3A). The mean error (systematic bias) was small and equal to 0.6 events/h of recording (Figure 3B). These performances are better than those observed with a single-threshold detector previously used in Holters [4]. They are in fact very similar to commercial portable polygraphs that use automatic analysis of nasal airflow and/or oxygen

saturation, although here all patients had an AHI>15 [5].

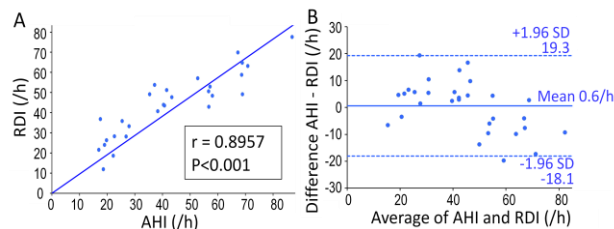


Figure 3: Index-based performances. Correlation (A) and Bland-Altman plot (B) between RDI and AHI.

A total of 4240 events (2561 apneas and 1679 hypopneas) were scored on the PSG. The detector is highly sensitive 86.2% (Table 2). This is significantly higher than reported event-based detections of PAP devices [6]. Note that apneas (hypopneas) were detected in real-time with a median delay of 6.0 (8.7) seconds after the start of the annotated event. This detection delay mostly explains the comparatively reduced sensitivity, when using a sample-based approach (58.7%). This might nevertheless be sufficient to trigger efficient treatment.

The event-based PPV is 60.7%, partly because some detected events did not last 10 seconds (see second apnea in Figure 2C), and mostly because of many FPs occurring while the patient was awake. When subtracting the FPs during non-sleep periods, the PPV rises to 78.3% (C.I. 77.1%-79.4%) for apneas and hypopneas altogether. Adding an actigraph to the Holter could prevent some of the real-time FP detections.

Table 2: Performances of the detector (validation set): values given with 95% confidence limits (Wilson score) for apneas only (A), hypopneas only (H) and both (A+H).

	SE (%)	PPV (%)	Sp (%)	NPV (%)
Events	86.2%	60.7%	-	-
A+H	[85.2-87.2]	[59.5-61.9]	-	-
Events	90.4	67.6	-	-
A	[89.2-91.5]	[66.0-69.1]	-	-
Events	79.9	51.7	-	-
H	[77.9-81.7]	[49.7-53.6]	-	-
Samples	58.7	46.2	82.3	88.5
A+H	[58.1-59.2]	[45.7-46.7]	[82.1-82.5]	[88.3-88.7]
Samples	66.9	80.7	97.1	94.3
A	[66.2-67.6]	[80.0-81.3]	[97.0-97.2]	[94.1-94.4]
Samples	47.1	25.0	84.3	93.5
H	[46.2-48.0]	[24.4-25.5]	[84.1-84.5]	[93.3-93.7]

Accurate real-time detection of hypopneas remains a challenge, as highlighted by the low sample-based PPV for hypopneas (25.0%). In particular, the update of the baseline amplitude so as to follow the patient's changes in ventilation, notably during different sleep stages and postures, or during episodes of airflow limitation, is difficult. To anticipate the detection of severe hypopneas, one could reduce HYPOPNEA_CONFIM according to

the amplitude of NP drop, while milder airflow reductions could be detected with the usual 10 seconds.

For apneas, three patients had a particularly high number of FPs, due to very low respiratory rate in some sleep stages (7-9 vs. usual 12-18 breaths/minute). One solution would be to calculate APNEA_DET as a function of the valid respiratory cycles' periods, so that this parameter would be patient- and sleep stage- specific.

5. Conclusion

The real-time detector performances suggest sufficient sensitivity to drive future event-based SBD therapies. Although designed for detection in real-time, it could also serve as the basis of an off-line detector for SBD screening.

Acknowledgements

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