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# Pulse Photopletismography Amplitude Decrease Detector for Sleep Apnea Evaluation in Children

Eduardo Gil*, Violeta Monasterio, Pablo Laguna
Communication Technologies Group, I3A,
Zaragoza University
María de Luna 1, 50018 Zaragoza Spain
{edugilh,laguna}@unizar.es

José María Vergara
Sleep Department
Miguel Servet Children Hospital, Zaragoza
vergeur@comz.org

***Abstract*— A method for automatic detection of sleep apnea using pulse photopletismography signal (PPG) is proposed. This method is based on a detection of decreases on PPG amplitude fluctuations. The proposed detector is composed of three stages: pre-processing, envelope detection, based on root mean square series or Hilbert transform, and decision algorithm based on an adaptive threshold. The detector has been evaluated using simulated and real signals. Sensibility and positive predictive value of the detector where 76% and 73% for real signals. A clinical study to sleep apnea diagnosis in children based on this detector has been carried out. PPG attenuation events per hour ratio $E_h$ has statistical significance ($p < 0.05$) to classify children as normal $13.5 \pm 6.35$ $E_h$ ($mean \pm SD$) or pathologic $21.1 \pm 8.93$ $E_h$.**

## I. Introduction

Obstructive sleep apnea syndrome (OSAS) is characterised by repetitive episodes of upper airway obstruction during sleep, involving periods of breathing cessation [1], [2]. The prevalence of OSAS is estimated at 4% in adult men, 2% in adult women and 2% to 3% in children, most of whom are undiagnosed and untreated [3]. The resulting sleep fragmentation not only causes daytime sleepiness but also may in turn lead to systemic hypertension [4], an increase of cardiovascular disease, arrhythmias and problem behaviors in children [5], [6]. The preferred treatment nowadays is the application of a continuous nasal positive airway pressure (CPAP) via a nasal mask in adults [7] whereas adenotonsillectomy is the first line of treatment for most children [8].

The gold standard diagnostic test for OSAS is overnight polysomnography (PSG). PSG generally includes monitoring of the patient's airflow through the nose and mouth as well as the measurement of blood pressure, electrocardiographic activity, blood oxygen levels, brain wave patterns, eye movement, and the movement of respiratory muscles and limbs. Because of the cost and requirement for technical expertise, a number of alternatives to PSG have been proposed [9].

Pulse photopletismography signal (PPG) is one of these alternatives. This signal, obtained through non invasively pulse oximetry systems, is based on light absorption [10]. PPG shows continuous arterial blood volume and as a consequence of that a marked cyclical pattern appears synchronized with heart rate. Arterial blood volume depends on both blood pressure within the cardiac cycle and on artery contractibility degree, the latest depending on sympathetic and parasympathetic system activity.

The autonomic nervous system (ANS) regulates the bodily activities which are beyond conscious control. The ANS actually consists of two subsystems which operate in reverse of each other, thus, the sympathetic nervous system is the dominant system when physical activity is called for, and the parasympathetic nervous system dominates during relaxation. Both these subsystems innervate the same organs and act to maintain a proper balance of the internal organ environment. Wall arteries are covered by muscles which are able to contract or relax, producing arteriolar constriction or dilatation. This is regulated by several mechanisms such as the vegetative system which determines the vascular muscle tone. According to the dominant system (i.e. sympathetic or parasympathetic) the blood vessel contracts (vasoconstriction) or dilates (vasodilatation).

Several studies suggest that when an apnea occurs, an increase in sympathetic activity is produced. Hypoxia plays a key role in this relationship. This increase is associated with a vasoconstriction and is also possibly related to transient arousal [11]–[14]. Vasoconstriction is reflected at PPG signal by a decrease on the fluctuation of the signal amplitude. So an automatic detection of PPG attenuation periods could be useful to indirectly quantify apneas during sleep. There are some studies about OSAS diagnosis based on vasoconstriction detection that use peripheral arterial tonometry [15]–[17], which is a similar physiologic signal.

This work presents an automatic strategy for PPG attenuation detection and quantification.

## II. Materials and Methods

### A. Signals

*1) Simulated signals:* The detector was first evaluated using simulated signals. The simulated PPG is generated by a repetition pattern which corresponds to a real PPG cardiac cycle, so the simulated signal keeps the real signal's morphology. The sampling rate used is 50 hertz. To generate the simulated signal several parameters can be adjusted: baseline level, oscillation amplitude, signal to noise ratio (SNR), attenuation events frequency and deep.

An apneic event is defined as a attenuation greater than 66% in PPG oscillation amplitude of at least 3 seconds.

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Then, non apneic events correspond to periods of attenuation lower than 66%. Location in time, duration and oscillation amplitude of events are randomly generated using uniform variables so that durations are comprised in the interval of 3 to 30 cardiac cycles. Noise contamination is simulated by white noise.

*2) Real signals:* The study includes 25 children (16 Boys, 9 girls) suspected of having OSAS. Children have a mean age $4.56 \pm 1.79$ ($mean \pm S.D.$). Sleep studies were carried out in Miguel Servet Hospital, Zaragoza, using a digital polygraph (EGP800, Bitmed) recording six EEG channels, two electro-oculogram channels, a chin electromyogram channel, an ECG channel, air flow (oronasal thermocoupler) and respiratory plethysmography with transducers placed around the chest and abdomen. PPG and arterial oxygen saturation were continuously recorded by pulse oximetry (COSMO ETCO2/SpO2 Monitor Novametrix, Medical Systems). All signals were stored using a sampling rate of 100 hertz.

The polysomnogram was scored manually according to standard criteria [18] discriminating children suffering from OSAS from those who does not.

In addition, decreases in the amplitude of PPG fluctuation greater than 60% were manually scored, in 9 signals, by a person while being blinded to PSG and to detector design.

### B. Detector

The proposed detector includes a PPG pre-processing, an envelope detector and a decision algorithm to determine vasoconstriction episodes, see figure 1.

*1) Pre-processing:* The mean PPG cardiac cycle length, $T$, is automatically estimated at the first stage from the PPG signal, $x_p(n)$, and is used as time reference by the detector. The cardiac cycle is computed by means of a zero crossing detector applied to $x_p(n)$ after been corrected for the mean. After that a running mean subtraction is applied using a moving average filter, as indicated.

$$s(n) = x_p(n) - \frac{1}{M} \sum_{k=n-(M-1)}^{n} x_p(k) \qquad (1)$$

Where $M$ is the filter length. A value of $M = 25 \times f_s \times T$, where $f_s$ is the sampling rate, has been probe to be a good trade off between smoothing and tracking of the different mean values at the attenuation episodes.

*2) Envelope detection:* The second stage detector target is to get an adequate signal to be compared with a threshold in the third stage, see below. Two alternatives have been implemented, one based on the root mean square (RMS) series, $x_e^{RMS}(n)$, and another based on Hilbert transform, $x_e^h(n)$.

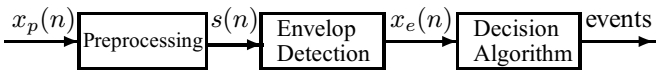


Fig. 1. Detector diagram

$x_e^{RMS}(n)$ was estimated by the root mean squared in an N length window.

$$x_e^{RMS}(n) = \sqrt{\frac{1}{N} \sum_{k=n-(N-1)}^{n} s^2(k)} \qquad (2)$$

Actually $x_e^{RMS}(n)$ is only required to be computed at lower sampling rate (two times the heart rate) reducing the computational load.

According to [15] vasoconstriction periods last between 3 to 30 seconds. The bigger the N value the higher the low-pass filtering and then slower transitions at $x_e^{RMS}(n)$, so a small value of N (equivalent to two cardiac cycles) is used. This election has the risk of possible increase of short length false positives which are due to the high signal variability, and also several apneic events instead of one can be considered as different when the signal amplitude is near the threshold at the decision rule. In order to solve the problems mentioned before two parameters are included: the minimum events' duration and the minimum distance between events. In this way sort length false positives are suppressed and near detections are grouped together.

The second alternative is based on Hilbert transform, see [19]. The envelope $x_e^h(n)$ is obtained as a low pass filtered signal from $s_e(n)$.

$$s_e(n) = \sqrt{s^2(n) + \hat{s}^2(n)} \qquad (3)$$

Where $\hat{s}(n)$ is the Hilbert transform of $s(n)$, obtained by DFT calculation, Hilbert Transform at the frequency domain and inverse DFT calculation again.

Finally $s_e(n)$ is low pass filtered using a cut frequency of 0.3 hertz to get the envelope $x_e^h(n)$.

*3) Decision algorithm:* The detector's last stage is a decision rule based on an adaptive threshold. An event is considered as apneic by the decision algorithm when $x_e(n)$ is lower than the established threshold, $x_e(n) < \zeta(n)$.

$$\zeta(n) = \begin{cases} \frac{U}{100L} \sum\limits_{\substack{k=n-(L-1)-T_{L,n} \\ k \in \{n_a\}}}^{n} x_e(k) & n \in \{n_a\} \\ \zeta(n-1) & n \in \{n_c\} \end{cases} \qquad (4)$$

Where $\{n_a\}$ is the sample set which fulfils the criterion of being eligible for threshold updating and $\{n_c\}$ is the sample set which does not fulfil this criterion, $T_{L,n}$ is the number of samples $\in \{n_c\}$ inside the interval $[n-(L-1)-T_{L,n}, n]$ so that L is always the number of samples in $\{n_a\}$ set.

The threshold is calculated as the U percent of the mean of the L pass samples in $\{n_a\}$.

The set of samples not eligible for threshold update can result from the following different criteria:

- When an apneic event is detected. If the sample n accomplishes that $x_e(n)$ is lower than the threshold $\zeta(n)$.
- An artefacts' detector based on Hjorth parameters [19] has been implemented. A sample $n$ is included in $\{n_c\}$

when it is considered as an artefact by this detector. The Hjorth parameters, $\mathcal{H}_1$ represents an estimate of the dominant frequency and $\mathcal{H}_2$ is representative of half the bandwidth, are calculated using a shifting overlapped window whose length $P = 5$ seconds $\times f_s = 500$ samples.

$$x^{(1)}(n) = x(n) - x(n-1) \tag{5}$$

$$x^{(2)}(n) = x(n+1) - 2x(n) + x(n-1) \tag{6}$$

$$\hat{\overline{w}}_i(n) \approx \frac{2\pi}{P} \sum_{k=n-(P-1)}^{n} \left(x^{(i/2)}(k)\right)^2, i = 0, 2, 4 \tag{7}$$

$$\mathcal{H}_1(n) = \sqrt{\frac{\overline{w}_2(n)}{\overline{w}_0(n)}} \tag{8}$$

$$\mathcal{H}_2(n) = \sqrt{\frac{\overline{w}_4(n)}{\overline{w}_2(n)} - \frac{\overline{w}_2(n)}{\overline{w}_0(n)}} \tag{9}$$

When either $\mathcal{H}_1(n) \leq \tilde{\mathcal{H}}_1 - 1$ or $\mathcal{H}_1(n) \geq \tilde{\mathcal{H}}_1 + 1.4$ or $\mathcal{H}_2(n) \geq \tilde{\mathcal{H}}_2 + 3$ are accomplished the sample $n \in \{n_c\}$. Where $\tilde{\mathcal{H}}_1$ and $\tilde{\mathcal{H}}_2$ are the $\mathcal{H}_1(n)$ and $\mathcal{H}_2(n)$ median value respectively in the total record.

- When an abrupt change in $x_e(n)$ occurs, so when $x_e(n)$ amplitude starts to fall, due to an apneic event, the threshold keeps constant. These abrupt changes are controlled by means of the derivation of $x_e(n)$ according to (10).

$$|x_e(n) - x_e(n-1)| \leq 0.05A_0 \tag{10}$$

Where $A_0$ is half the oscillation range amplitude of $x_p(n)$ at the beginning of the recording.

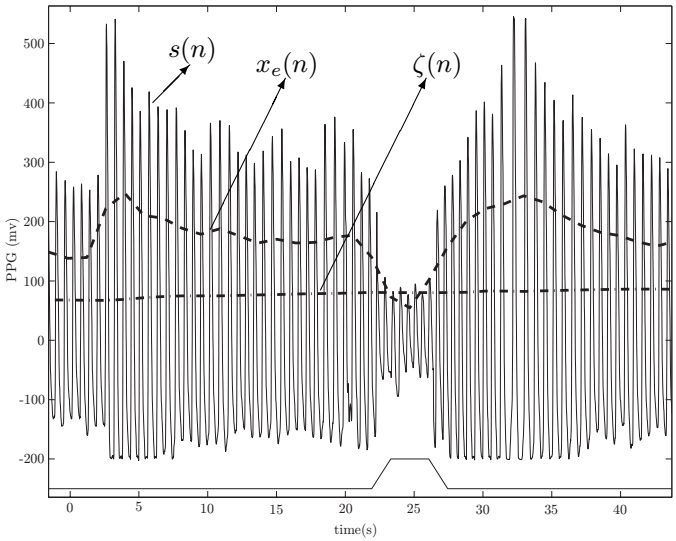


Fig. 2. Example of an event detection. $s(n)$ (solid line), $x_e(n)$ (dashed line) and $\zeta(n)$ (dash-dotted line)

When some of this three criteria are accomplished, it is considered that $n \in \{n_c\}$.

Fig. 2 shows how the detector works and how a vasoconstriction event is detected. The detector parameters at this example are $U = 45$, $L = 30 \times f_s \times T$ and RMS envelope detection.

### C. Data analysis

In order to evaluate the detector two studies have been developed. The first using simulated signals and the second using real signals. The beginning and end of detected events is the detector output. The sensibility (S) and positive predictive value (+PV) are calculated by comparing this output with the reference annotations. The number of true positives, false positives and false negatives is estimated through the comparison of the events' onset.

The events per hour ratio $E_h$ has been calculated for 11 children with OSAS and 11 children without. Results are computed as $mean \pm SD$ of $E_h$. Student's t test was used for comparison between groups. We considered $p < 0.05$ to be statistically significant.

## III. Results

500 simulated signals of one hour length were used. These simulations were done in 5 groups of 100 signals depending on parameters configuration. All simulated signals have a baseline of 250 millivolts, an amplitude oscillation of 500 millivolts, 20 apneic events, corresponding to a moderate OSAS in adults and 10 non apneic attenuation events. SNR and detector threshold percent U were the variable parameters. The results obtained are shown on table I. The bigger the noise the smaller the S and +PV, as expected. A threshold increase supposes also an augmentation on sensibility, due to more attenuation events considered as apneic, but also a +PV decrease so that false positives increase too.

In the real signals experiment, the manually annotated events become the reference. The detector has been evaluated using several values of L and U (see table II). The S and +PV have been obtained using gross averages. $S = 0.76$ and $+PV = 0.73$ are the best results obtained for $U = 50$, $L = 20 \times f_s \times T$ and envelope detector based on Hilbert transform.

In addition, a study to evaluate the accuracy of this method for OSAS diagnosis in children has been carried out. The detector parameters used are those that got the best results in the evaluation with manually annotated real signals. For patients with OSAS, mean $E_h$ was $21.13 \pm 8.93$ and for patients without OSAS, mean $E_h$ was $13.49 \pm 6.35$ ($p < 0.05$).

TABLE I
SIMULATION STUDY RESULTS

| | $SNR = \infty$ $U = 35\%$ | $SNR = 30dB$ $U = 35\%$ | $SNR = 25dB$ $U = 35\%$ | $SNR = \infty$ $U = 40\%$ |
|---|---|---|---|---|
| $S$ | 0.9819 | 0.9756 | 0.9336 | 0.9832 |
| $+PV$ | 0.9723 | 0.9770 | 0.9684 | 0.9317 |

TABLE II
REAL SIGNAL RESULTS

| Detector | RMS | | Envelope | |
|---|---|---|---|---|
| L / U | 45 | 50 | 45 | 50 |
| 20 | $S = 0.70$ | $S = 0.83$ | $S = 0.61$ | $S = 0.76$ |
| | $+PV = 0.77$ | $+PV = 0.60$ | $+PV = 0.86$ | $+PV = 0.73$ |
| 25 | $S = 0.71$ | $S = 0.83$ | $S = 0.61$ | $S = 0.76$ |
| | $+PV = 0.76$ | $+PV = 0.59$ | $+PV = 0.85$ | $+PV = 0.71$ |
| 30 | $S = 0.73$ | $S = 0.84$ | $S = 0.63$ | $S = 0.77$ |
| | $+PV = 0.75$ | $+PV = 0.59$ | $+PV = 0.84$ | $+PV = 0.71$ |

## IV. DISCUSSION

The present work aims to be a first step in evaluating the PPG usefulness to OSAS diagnosis. In order to do this, two strategies of PPG amplitude decrease detectors have been developed. These events are due to vasoconstrictions produced by an ANS activity increase and reflect an apneic event probably related to an arousal. The detector has been evaluated using simulated and manually notated real signals. It was intended to get the optimum detector parameters in terms of maximum $S$ and $+PV$. The best parameters values are a compromise since a considerable inter signals variability exists. Both strategies have obtained similar results with only marginal differences on performance. Finally a clinical study to SAOS diagnosis in children has been carried out. Although a wider study using more cases would be needed, the discriminant index $E_h$ has statistical significance to classify children as control or pathologic.

A deeper study analysing the relationship between each clinically annotated apnea an each PPG amplitude decrease event is necessary. Obviously not all PPG amplitude decrease events correspond to an apnea so that a high value of $E_h$ has been obtained even for control children. Others criteria to discriminate between the PPG amplitude decrease associated with apnea and those without are necessary. According to [20], heart rate variability could be an interesting alternative to explore as Schnall et al. [15] proposed.

PPG seems to offer interesting information to OSAS diagnosis, with the great advantage of being less complicated and better suited for ambulatory monitoring.

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