

EVALUATION OF ACOUSTIC SIGNALS FOR THE DETECTION OF SLEEP APNEA EVENTS

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Abstract: The present paper concerns acoustic measurements in connection with the sleep apnea syndrome. The latter induces pathophysiologic changes in the respiratory and circulatory systems and is diagnosed by polysomnography (PSG). PSG comprises the measurement of many parameters and thus tends to be expensive and time consuming. In contrast to available portable devices which use many distributed sensors, we propose a single acoustic sensor in the heart region from which multiple information is derived through excessive signal processing. Measurement is performed by means of a microphone complemented by a stethoscope head. Analysis of the detected signal by means of histogram, cross correlation and FFT shows that it includes information on all three the cardiac activity, the respiratory activity and the snoring sounds. In order to obtain automatic monitoring of obstructive apnea events, a series of signal characteristics in both the time and the spectral domain (e.g., power density, histogram width, PCA components) are subject of processing through adaptive algorithms which are also used for classification. In addition, the method yields breathing patterns and diagnostically relevant physiological data such as changes of the heart rate and the rate of respiration. For the detection of central apnea event, the acoustic sensor is complemented by a magnetostrictive one.

Keywords: Acoustic measurements; Sensors; Physiological monitoring; Signal processing

1 INTRODUCTION

The present study concerns acoustic measurements in connection with the so-called sleep apnea syndrome (SAS). SAS is a complex medical problem which is characterised by cessation of effective respiration during sleep. Two types of the syndrome are of great interest because of their high prevalence in the population:

- (i) The obstructive sleep apnea (OSA) syndrome, characterised by occlusions of the trachea with maintenance of respiratory efforts. It concerns mainly elder and male persons, the prevalence being of several percent according to (diverging) estimations.
- (ii) The central sleep apnea (CSA) syndrome, caused by ceased respiratory efforts due to lacking neural stimulation of the lungs. It affects predominantly children and is a possible cause of the sudden infant death syndrome.

Besides multiple pathologic changes in the respiratory and circulatory systems that lead to pulmonary hypertension and heart failure, SAS may cause severe reduction of life quality, excessive daytime somnolence, strong decrease of life expectancy as well as disruption to other family members [1].

Routine assessment of SAS and the physiological state of patients is performed in sleep laboratories by so-called polysomnography (PSG). Unfortunately, it is expensive and time consuming because of the high amount of sleep parameters to be recorded and attended during an overnight sleep session. The availability of PSG is strongly restricted even in economically leading EU countries. There is a great interest in portable, cheap and patient friendly screening tools.

A wide pallet of portable recording devices is available which register up to about 8 parameters (e.g., MESAM [2] or NighWatch). These devices are characterised by sensors being regionally distributed which yields distinct drawbacks such as high expenditure for ambulant performed hook-up, instruction of the patient, inconvenience for the latter and various sources of data loss, e.g., due to

imperfect sensor attachment. Moreover restrictions to single parameters have been evaluated but failed to deliver reliable diagnosis (e.g., tracheal sounds [3] or oximetry).

In order to connect the advantage of a single detector with the establishment of a high degree of physiological information as needed for effective assessment of SAS and the physiological state of patients, we propose a novel method based on exclusive use of acoustic sounds in the heart region. We apply a simple-to-handle single-spot detector, dispensing with precise detection of many well defined physiological parameters and compensating the lack of exact information by extensive use of statistic approaches and excessive signal processing in both time and spectral domain.

2 EXPERIMENTAL

Sound signals from 30 patients (25 male, 10 with SAS) were recorded via microphone (prepolarized condenser from AKG) attached using a plastic tube to a chestpiece (Littmann) placed on the thorax near the heart (Fig. 1). Firm contact between the skin and the chestpiece was guaranteed by special glue tape. Sound recordings (sample frequency 4kHz, quantisation 16bit) were performed in parallel with PSG as a golden standard.

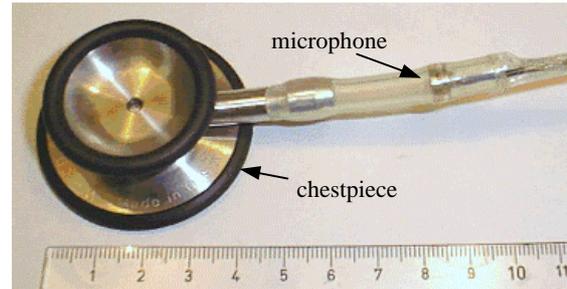


Fig. 1. The sensor – a microphone attached to a chestpiece.

Transmission of body sounds - heart, lung and snoring sounds - is influenced by the acoustic transfer function of both the chestpiece and the plastic tube (Fig. 2). Linear conversion of the sound pressure into a voltage is performed by the microphone with flat frequency response from 20Hz to 2000Hz.

Various *time-based characteristics* were calculated from the raw acoustic sensor signal s after application of partly overlapping windows of 0.256s duration (Fig. 3):

- (i) The signal power in different frequency bands (P for 0.1...2 kHz, P_{VL} for 0...0.1 kHz, P_L for 0...0.3 kHz, P_M for 0.3...0.8 kHz, P_H for 0.8...2 kHz).
- (ii) The root mean square S_{RMS} and the histogram width S_{HIS} in the height of 10%.

The frequency band choice in (i) considers both widely used frequency cut-offs (e.g. 0.8kHz in [2]) and our experience from pre-studies. All characteristics were smoothed by a FIR filter in order to reduce artifacts. Additionally, 21 *event-based characteristics* were defined, for instance the relations between P_L , P_M and P_H during breathing or snoring events.

For improved registration of CSAs - see below - a magnetostrictive bending sensor was used as an additional component in the single spot detector [4]. The latter sensor acts as a transducer from chest wall deformation - due to heart/lung activity - to an electrical signal.

3 EVALUATION AND RESULTS

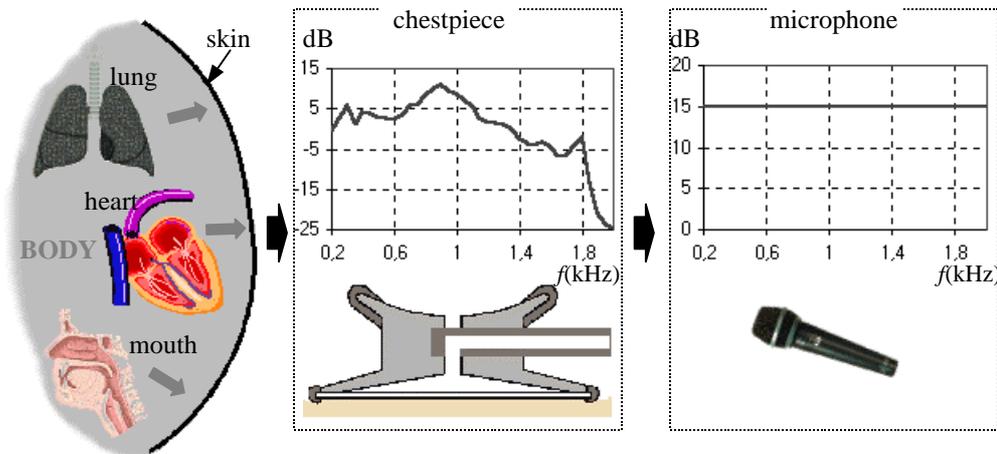
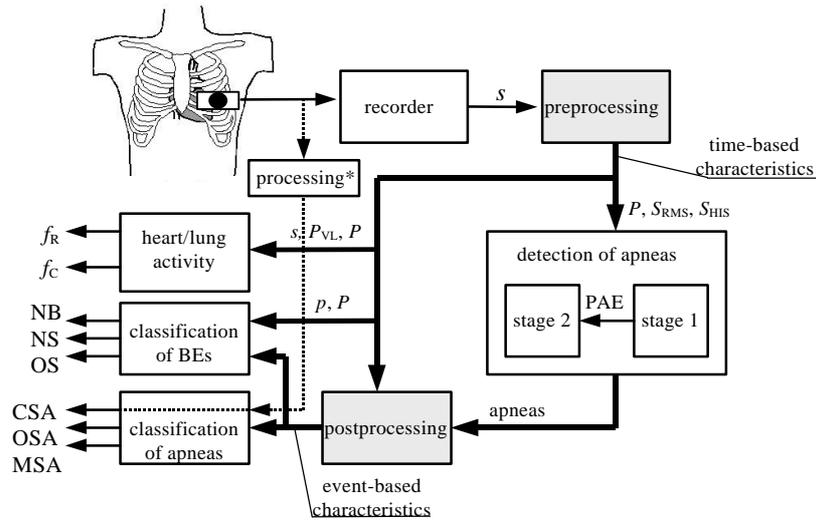


Fig. 2. Generation and recording of sounds. The acoustic transfer function of the chestpiece and the transfer function of the microphone are depicted above.



* Processing of the signal from the magnetostrictive sensor

Fig. 3. Workflow of the algorithm for the registration of f_C and f_R as well as for the detection and classification of apneas (as closer described in [4]) and the classification of BEs.

The sound signal evaluation showed that it consists mainly of three additive components: heart, lung and snoring sounds. However, also some weak interactions among the latter components - resulting in mutual modulation of components - were observed, due to physiological interactions between their respective sources: heart, lung and mouth (e.g., respiratory sinus arrhythmia).

Heart sounds showed spectral components predominantly in the frequency band 0...100Hz, lung sounds in 0.1...1kHz and snoring sounds in 0.1...2kHz. For signal processing, the situation of these bandwidths is very favourable, because direct auscultation of the physiological information (e.g., heart frequency f_C or respiratory frequency f_R) would presume critical recording of the body sounds in the frequency band below 5Hz where the microphone sensibility tends to be extremely low.

3.1. Heart rate and respiratory rate

The registration of f_C was based on the oscillations of P_{VL} due to the periodic appearance of the heart sounds and was performed by FFT of P_{VL} . As a second method, extraction of n quasi-stationary maxima in power density of the acoustic signal $p(f) \in 0..5\text{Hz}$ and inspection of the possible multiple relationship of the frequencies at these maxima led to f_C , according to

$$f_C \approx \max_{f_C' \in 0..5\text{Hz}} \left[\int_0^{5\text{Hz}} p(f) \cdot [\delta(f - f_C') + \delta(f - 2 \cdot f_C') + \dots + \delta(f - n \cdot f_C')] \cdot df \right] \quad (1)$$

(δ ...dirac impulse, $\max(\xi(f))$ calculates $f \in f_C$ at which the argument function $\xi(f)$ has its peak value).

Fig. 4 shows the changes of f_C during an OSA - surrounded by significantly high s and P - with ABEs in the middle. Two consecutive heart sounds are zoomed out in Fig. 4a. f_C shows a decrease (increase) after the beginning (end) of the apnea - a typical f_C behaviour (see "Pulse" in Fig. 7a). A temporal increase of f_C can be observed during the ABEs as a consequence of normalised physiological burden.

Periodical oscillations of P were observed during steady respiratory activity since lung/snoring sounds exhibit high pitched spectral components which contribute to the signal power P . f_R was evaluated by means of FFT from the oscillation period of P (Fig. 5b) which corresponds to the period of breathing events (Fig. 5a).

3.2. Breathing events

The classification of breathing events (BEs), like normal breathing (NB), normal snoring (NS) and obstructive snoring (OS), was carried out evaluating the width and form of the auto-correlation function of p , ΔP and a few other *event-based characteristics*.

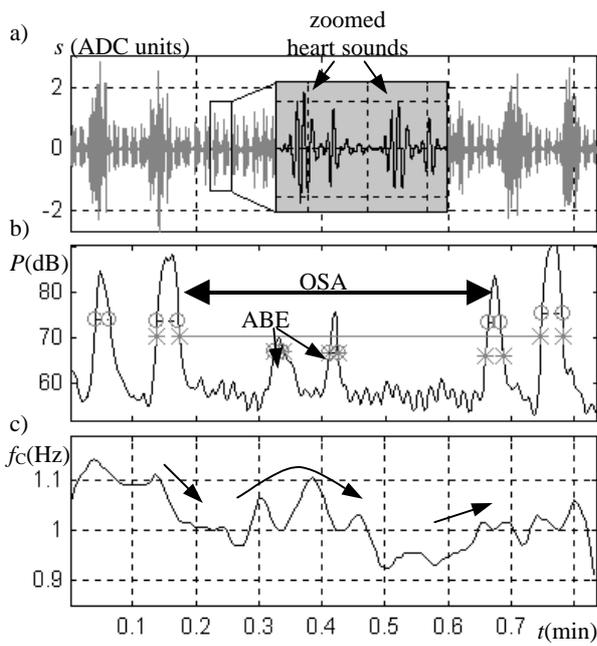


Fig. 4. Detection of f_c during an OSA episode. a) Body sounds in time domain. b) Variable signal power of the sounds. c) Typical variation of f_c during the OSA.

A rough classification of BEs - exact classification is not possible due to the lack of a strict definition - could be attained in a simple way. As shown by Fig. 6, the amplitude range of P increases as the airways get more obstructed, i.e. from breath hold (BH) to OS. Fig. 6a and Fig. 6b demonstrate that the variation width of the signal power P of BH and NB do not differ significantly. Thus in some cases it is difficult to determine whether the patient is breathing or not. However, quite effective classifications of BEs could be attained from the following algorithm:

- (i) BH - $\Delta P=0..10\text{dB}$, $P_L > P_M > P_H$, signal bandwidth 0..0.1kHz,
- (ii) NB - $\Delta P=0..10\text{dB}$, $P_L > P_M > P_H$, signal bandwidth 0.1..0.5kHz,
- (iii) NS - $\Delta P=10..20\text{dB}$, $P_L \geq P_M > P_H$, signal bandwidth 0.1..1kHz,
- (iv) OS - $\Delta P=15..50\text{dB}$, $P_L \geq P_H \geq P_M$, signal bandwidth 0.1..2kHz.

3.3. Apneas

The developed algorithm for the detection of apneas was based on the fact that they are surrounded by significantly high snoring (or breathing) events. The detection was implemented in two stages: (i) detection of potential apnea events (PAE in Fig. 3) by means of adaptive thresholds and histogram analysis of P , S_{RMS} , S_{HIS} , and (ii) exact localisation of apneas - as described in detail in [5]. Furthermore, special methods were applied to extract

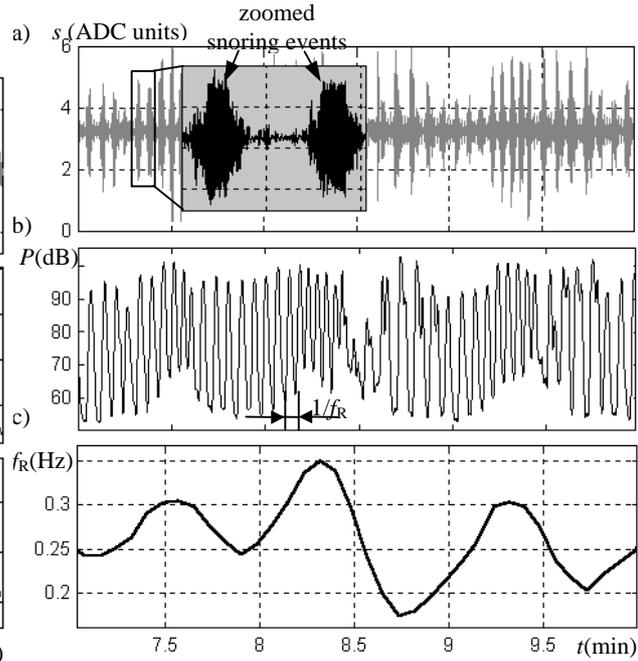


Fig. 5. Detection of f_r during obstructive snoring. a) Snoring in time domain. b) Periodic oscillations of P due to breathing. c) Extracted f_r .

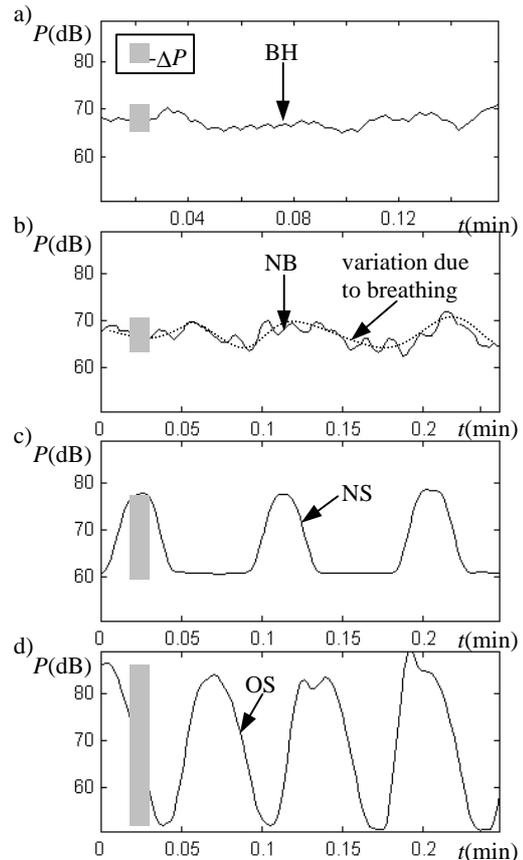


Fig. 6. Classification of BEs by means of ΔP . a) Breath hold (BH), i.e. mere heart sounds. b) Normal breathing (NB). c) Normal snoring (NS). d) Obstructive snoring (OS).

apnea breathing efforts (ABEs) during apneas for the sake of apnea classification.

The classification of apneas was performed by evaluating P during apneas. As an other method, the classification was attempted by reducing the dimensionality of *event-based characteristics* with the help of principal component analysis.

A typical example for the detection of apneas is given in Fig. 7. A mixed sleep apnea (MSA), i.e. an apnea with successive CSA and OSA character, was detected by PSG (Fig. 7a). As well it is clearly indicated by all three *time-based characteristics* S_{RMS} , S_{HIS} and P (Fig. 7b-d). In opposite to the first half of the MSA with CSA character, the second half shows some ABEs as the result of obstructed airway. Small amplitude oscillations in all characteristics are due to the heart activity, its rate corresponding to f_c .

A rough classification between OSAs and hypopneas with OSA character was possible by means of the mean signal power \bar{P} during airway occlusion (Fig. 8). The latter was by OSAs lower because of reduced breathing efforts during occlusion. Moreover a slight patient-specificity of the acoustic sounds was observed.

The detection of CSAs proved to be critical since most of them are not surrounded by NS (or OS) but by mere NB of low signal power (Fig. 6a,b). Significant improvements were reached using a magnetostrictive amorphous bending ribbon as a supplementary sensor component. Fig. 9 shows a typical output signal with marked breathing/snoring events. Amplitude variation due to respiration is at least 10dB higher than that due to heart activity and allows reliable detection of the onset of BH and thus a reliable registration of CSAs.

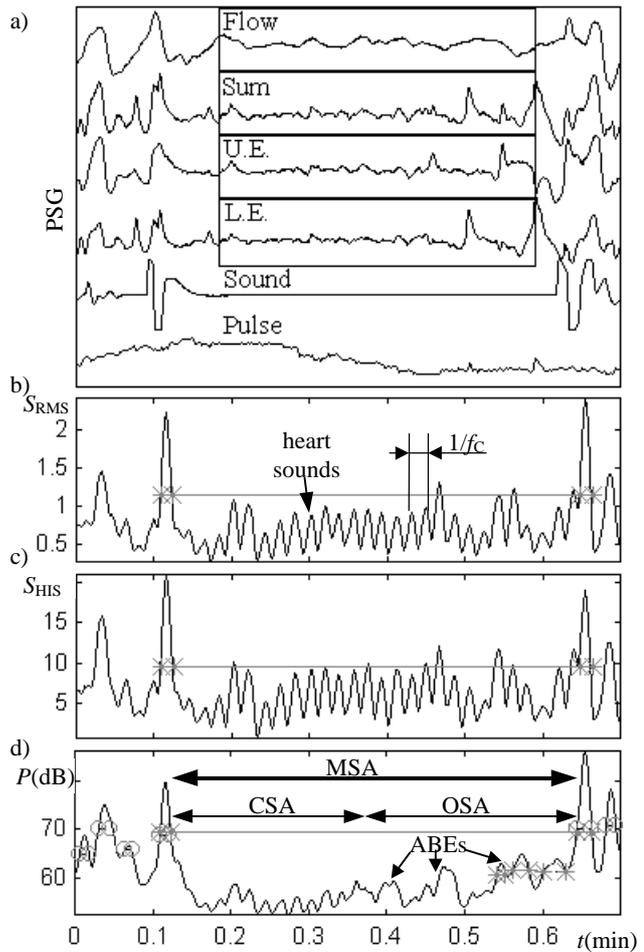


Fig. 7. Detection of a mixed sleep apnea (MSA) by means of acoustic sounds. a) PSG data with enframned MSA. b-d) Corresponding S_{RMS} , S_{HIS} and P with automatically detected MSA.

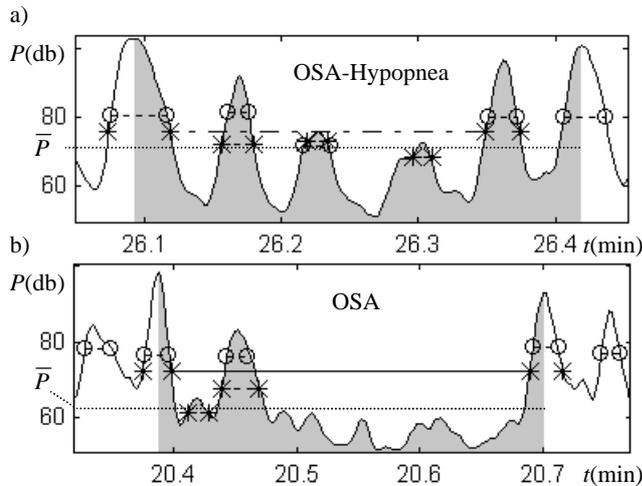


Fig. 8. Classification of apneas by means of mean signal power \bar{P} during airway occlusion. a) OSA-Hypopnea and its \bar{P} (ca.70dB), proportional to the filled area. b) OSA, \bar{P} is ca.62dB.

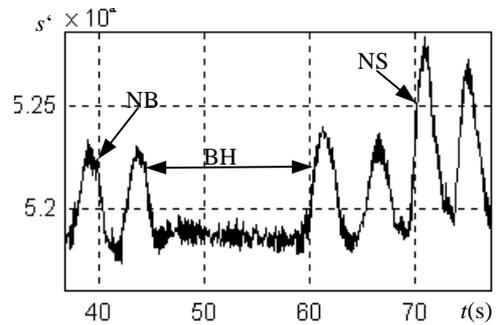


Fig. 9. Output of the magnetostrictive sensor component during normal breathing (NB), snoring (NS) and breath holding (BH) as a simulation of a central sleep apnea.

4 CONCLUSIONS

Single spot detection - a microphone with stethoscope head, arranged close to the heart - yields both obstructive apneas and diagnostically relevant physiological information, if being complemented by extensive processing of acoustic sound signals. The described sensor offers a cheap, compact and wearless tool for long time sleep monitoring. For a reliable detection of central apneas, the single spot detector is being complemented by a magnetostrictive bending sensor.

The global evaluation of results yielded a correspondence rate with PSG of 95% for patients with heavy SAS (RDI>30 of apneas and hypopneas of all types). The number of apneas was highly overestimated in healthy patients as already reported in [6]. The correlation between RDI from the algorithm and from PSG was 0.89, i.e. considerably high.

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REFERENCES

- [1] M. Sergi, M. Rizzi, Sleep apnea in moderate-severe obese patients, *Sleep and Breathing* 3(2) (1999) 47-52.
- [2] T. Penzel, G. Amend, MESAM: a heart rate and snoring recorder for detection of obstructive sleep apnea, *Sleep* 13(2) (1990) 175-182.
- [3] D. Brunt, K. Lichstein, Intensity pattern of snoring sounds as a predictor for sleep-disordered breathing, *Sleep* 20(12), (1997) 1151-1156.
- [4] E. Kaniusas, L. Mehnen, H. Pfützner, A magnetostrictive amorphous acceleration sensor for registration of chest wall displacements, *Proc. of SMM14 (Balatonfüred, 8-10. September 1999, Hungary)*, (1999) 176.
- [5] E. Kaniusas, H. Pfützner, B. Saletu, Detection of apneas by means of snoring sounds (in German), *Proc. of ÖGBT and ÖGMP (Vienna, 21-23. Mai 1998)*, 43 (2) (1998) 85-87.
- [6] J. Cumiskey, C. Williams, The detection and quantification of sleep apnea by tracheal sound recordings, *Am. Rev. Respir. Dis.* 126 (1982) 221-224.

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