Technical standards for using type III devices (limited channel studies) in the diagnosis of sleep disordered breathing in adults and children

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Obstructive sleep apnoea, limited sleep studies, respiratory polygraphy, polysomnography Type scoring, utility

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Abbreviations

AASM	American Academy of Sleep Medicine
AHI	Apnoea/hypopnoea index as defined using polysomnography
BMI	Body Mass Index
COPD	Chronic obstructive pulmonary disease
COVID19	Coronavirus Disease 2019
ECG	Electrocardiography
EEG	Electroencephalography
EMG	Electromyography
EOG	Electrooculography
HSAT	Home Sleep Apnoea Testing
Hz	Hertz
ICSD	International classification of sleep disorders
OAHI	Obstructive apnoea-hypopnoea index
ODI	Oxygen desaturation index
OSA	Obstructive sleep apnoea
OSAS	Obstructive sleep apnoea syndrome
OSAHS	Sleep apnoea/ hypopnoea syndrome
PAT	Peripheral arterial tonometry
PG	Polygraphy
PM	Portable monitoring
PSG	Polysomnography
PVDF	Polyvinylidene fluoride film
PWA	Pulse-wave amplitude
REI	Respiratory event index
REM	Rapid eye movement
RIP	Respiratory inductive/inductance plethysmography
RP	Respiratory polygraphy
SDB	Sleep disordered breathing
TF	Task Force
TRT	Total recording time

Abstract

For over three decades, Type III devices have been used in the diagnosis of sleep disordered breathing in supervised as well as unsupervised settings. They have satisfactory positive and negative predictive values for detecting obstructive and central sleep apnoea in populations with moderately-high pre-test probability of symptoms associated with these events. However, standardisation of commercially available Type III devices has never been undertaken and the technical specifications can vary widely. None have been subjected to the same rigorous processes as most other diagnostic modalities in the medical field. Although Type III devices do not include acquisition of electroencephalographic signals overnight, the minimum number of physical sensors required to allow for respiratory event scoring using standards outlined by the American Academy of Sleep Medicine remains debatable. This Technical Standard summarises data on Type III studies published since 2007 from multiple perspectives in both adult and paediatric sleep practice. Most importantly, it aims to provide a framework for considering current Type III device limitations in the diagnosis of sleep disordered breathing whilst raising research and practice-related questions aimed at improving our use of these devices in the present and future.

Introduction

In adults, the obstructive sleep apnoea/hypopnoea syndrome (OSAHS) is a highly prevalent disorder, which increases the risk of hypertension, cardiovascular mortality and is associated with impaired quality of life and traffic accidents [1-8]. The optimal diagnosis of OSAHS and the determination of its severity in individual patients is currently under debate [9] and includes discussion on how best to integrate nocturnal breathing disturbances and the degree to which they directly impact on symptoms and co-morbidities [10, 11]. Nevertheless, the apnoea/ hypopnoea index (AHI) remains essential to the diagnosis of individuals with OSAHS. The current classification system defines OSAHS based on an AHI ≥5 events/hour of sleep accompanied by symptoms of excessive daytime sleepiness, or with AHI ≥15/ hour sleep (ICSD-3, 2014). The AHI is calculated according to the number of apnoea and hypopnoea events per hour of sleep, with an apnoea defined as a pause in respiration ≥10 seconds and a hypopnoea defined as a ventilation reduction ≥30% resulting in an arterial oxygen desaturation of ≥3% or 4% or an arousal (AASM, 2021). OSAHS severity is classified as mild, moderate or severe according to AHI score cut-offs and can determine the type of treatment offered to the patient.

OSAHS in adults is diagnosed using either in-lab or unattended polysomnography or increasingly frequently by using Type III devices (see Table 1). Type III devices, also referred to a home sleep apnea testing if unattended (HSAT), respiratory polygraphy (RP) or limited channel studies, use 3-7 sensors/channels to acquire electrophysiological signals during the sleep period without incorporating any electro-encephalographic data [12]. Although in use for over 3 decades and initially designed as screening tools for sleep disordered breathing, there has never been any attempt to standardize or set agreed technical specifications for the sensors or algorithms utilized in acquiring data nor in the nomenclature, scoring criteria or cut-off values for diagnosing different types of sleep disordered breathing [13, 14]. Type III studies are primarily used as a ‘cheaper’ and ‘more convenient’ alternative to PSG in countless sleep centers world-wide, again based on limited evidence.

In children, OSAHS is also common, particularly during early childhood in association with lymphoid tissue overgrowth and increasingly in the context of obesity [1]. Scoring

criteria differ to those of adults, but the principles of investigation remain the same with Type III studies increasingly used in the diagnosis of sleep-disordered breathing in children, both in-hospital and at home. Apart from a document defining standards on using PSG and other devices for use in France [15], and the AASM guidelines on scoring paediatric sleep, no attempts have been made to standardise Type III device use in this group. The adolescent group is subject to most variation in assessment and no separate standards exist that are applied consistently. An AASM position paper published in 2017 did not support the use of home sleep studies for the diagnosis of OSAHS in children due to insufficient validation and monitoring available for most devices (i.e. absence of CO_2 partial pressure, arousal monitoring and calculation of total sleep time) [16] . Since the publication of this position paper, a number of studies comparing Type III devices to in-laboratory PSG in children have become available making this evaluation necessary [16].

The aims of this TF were to examine and establish standards and specifications in the acquisition and scoring of respiratory events using limited studies in both adults and children and to call to attention the fact that very few technical standards exist at all with respect to terminology, quality and technical specifications of equipment used for acquiring the physiological signals, respiratory event scoring criteria and patient information provided.

Methods

The TF was comprised of experts in managing and scoring adult and paediatric PSG and respiratory polygraphy (RP) (table 1). Two patient representatives were also included. Members were assigned to working groups within the TF.The following areas were covered: technical specifications of Type III devices, utility of Type III devices in comparison to PSG for investigating sleep-disordered breathing, scoring criteria for sleep related breathing disturbances using Type III devices in adults and scoring criteria for sleep related breathing disturbances using Type III devices in children.

The work was co-ordinated by email and through teleconference interactions, and no physical meetings of the full TF were held on account of the COVID19 pandemic

(2019-2021). Each working group completed their section, which was integrated into a final report by the TF chairs (RLR, WR).

A systematic literature search (PubMed) was performed by a research assistant (KS) together with the members of each working group from January 2007 to November 2021, and the respective publications were retrieved. Reference lists were systematically examined for relevant articles and included. Keywords were selected that were appropriate to the relevant working group, then, appropriate search words were added. Details of search criteria, keywords and comparisons can be found in the online supplement. PRISMA flowcharts were used to document the search results [17].

The year 2007 was used as the initial year for searches on account of the publication of the American Academy of Sleep Medicine's new technical specifications, rules and terminology for polysomnography testing and scoring [18], subsequently adopted internationally. For the technical specifications, the year 2011 was used on account of the publication of the SCOPER paper which discussed signal derivation from Type III studies [19]. The criteria and flow charts for the literature searches for each working group are provided in the supplementary material (see online supplement). Inclusion criteria were articles in any language and data on human subjects; exclusion criteria were reviews, guidelines or case reports. Each working group extracted and analysed data as considered relevant to the Technical Standard.

The creation of this Technical Standard combined an evidence-based approach as far as feasible with the expertise of the TF members. Discussion was undertaken within each working group initially, followed by review of the entire Technical Standards by the whole group. All members of the TF approved this document and reached consensus on the Technical Standards. Accordingly, this Technical Standard provides an overview of current knowledge and practice in the area in addition to clarifying limitations that require further attention and research to allow for future recommendations.

Results

Technical specifications for Type III devices

The literature search retrieved 250 references (see online supplement figure e1). After abstract and text screening, 82 references remained. Based on further evaluation of the reference lists of these 82 references, 61 references were finally included.

Evidence overview of sensors used to acquire physiological signals during sleep

The various sensors used to analyse breathing disorders during sleep have been reviewed and are summarised in the online supplementary materials (see online supplement table e3 – e23). Studies published since the SCOPER paper (2011) were included [19].

Measurement of respiratory flow signals

The pneumotachograph is the gold standard for accurate assessment of breathing flow [20-22]. Nasal cannulas are excellent surrogates, are used most frequently and have been validated extensively; all have their limitations (see table e4). Thermistors are less sensitive in detecting hypopnoeas but perform well in obligate mouth breathers and when there is reduced nasal patency. A new thermal-based sensor system has been developed for low air flow detection, with low-power dissipation, high linearity and of small dimensions [23]. Innovative sensors such as polyvinylidene fluoride film (PVDF) nasal flow sensors have been introduced, which are much more sensitive than thermistors while encompassing their advantages [24]. The tracheal sound sensor (PneaVoX) is a threefold sensor and holds promise in optimising assessment while decreasing the number of sensors applied to the body; it allows for wireless recording, with response characteristics that are linear over a wide range of frequencies [25-28].

Characterisation of breathing during sleep

Full PSG with oesophageal pressure measurement is considered the ‘gold standard’ for characterising sleep breathing events. In RP, different surrogates are used,

including thoraco-abdominal movements, pulse transit time, peripheral arterial tonometry (PAT)/photo plethysmography (PPG), jaw movement, and suprasternal pressure. The most common surrogates used are the thoraco-abdominal bands, especially respiratory inductance plethysmography (RIP), while effort belts with PVDF may be used, just as RIP often is, as a 'back-up' signal for detecting respiratory events when nasal pressure signals become artefactual or are lost [29]. RIP belts have replaced piezoelectric belts in more recent studies and can be used in a calibrated or uncalibrated manner. Algorithmic approaches can enhance the performance of piezoelectric belts [30]. PPG has been extensively used in recent years and can extract features from different frequencies of the RR interval signals to detect OSA as well as sleep stages [31-46]. PTT which reflects changes in pleural pressure and detects autonomic arousals is a useful tool for distinguishing central from obstructive events. Different parameters and machine learning algorithms can improve its systemic accuracy [47]. Chest-worn accelerometry can be a robust and accurate method for the measurement of respiratory features, based on a single point of mechanical contact with the chest. Wrist worn accelerometry can provide a degree of surrogate measurement of respiratory movement as well.

Quantification and measurement of snoring

The nasal cannula shows poor reliability and accuracy for measuring snoring, since it only detects frequencies up to 100 Hz, compared to the 4 kHz that a microphone can capture [48]. Microphone based technologies can be optimised to perform automatic analysis of snoring, including determination of synchronisation with inspiration below a maximal frequency level (500 Hz) and exclusion of any noise resulting from movement [49]. Piezoelectric vibration sensors can provide data on snoring, as well as movement and heartbeat during sleep, profiting from new algorithms for automatic snoring detection [50].

Position sensing during sleep

Accelerometers make use of 3-D signals, identify the orientation of the device relative to the line of gravity, thus quantifying position shift [51-58] and indicating arousals.

Pulse Oximetry

Pulse oximeters make use of photoplethysmography; they behave differently, depending on the sampling rate, the technology utilised as well as the measurement site (e.g. finger versus ear lobe).

For a detailed overview of measurement techniques, their advantages and disadvantages, the reader is referred to the online supplement (tables e3 – e23).

Limitations and Remarks regarding sensors used to record physiological signals during sleep

There have been considerable advances since 2011 in the development and refinement of non-invasive sensors and techniques for measuring respiratory and sleep variables. However, few have been standardised against each other or against an 'ideal' acquisition signal. Variations in acquisition, sampling rates and sensitivity can affect signal quality and integrity, hence scoring and diagnostic outcomes.

Technical Standards re: Type III device specifications

The nasal cannula is the best-validated surrogate for hypopnoea detection owing to its good frequency response, whilst the thermistor is the recommended sensor for apnoea detection. PVDF sensors and tracheal sound sensors deserve a more prominent role, given their high sensitivity.

RIP bands should be the standard technique used to discriminate between the types of respiratory events in a routine setting. Jaw movement, suprasternal pressure, accelerometers and use of indirect signals like peripheral arterial tonometry/photoplethysmography are alternatives that are less obtrusive but require further validation.

The lack of consistency between snoring sensors affects future research on the clinical significance of snoring. Standardisation of objective snore measurements is necessary.

The acquisition parameters of pulse oximeters should be disclosed whenever oximetric data are reported, and efforts should be made to standardise them.

Scoring criteria for sleep related breathing disturbances when using Type III devices in adults

The literature search retrieved 991 references. After abstract and text screening, 286 references remained. Further evaluation of the references resulted in 48 references being included (see online supplement figure e2).

Methods for estimating total sleep time

Evidence overview

Since Type III studies do not include EEG measurement, the number of apnoeas and hypopnoeas cannot be expressed as being *per hour of sleep*. Thus, total recording time (TRT) is often used as the denominator to calculate respiratory event frequency or the oxygen desaturation index (ODI) [59-77]. The difference between the mean TRT and mean total sleep time (TST) ranges between 1 and 3 hours based on the literature. Different techniques have been used to optimise TST, by increasing the accuracy of start and stop times and/or by removing estimated wake periods. These include event markers [78-80], actigraphy or position sensors [61, 63, 81-85], sleep diaries [61] or combined use of actigraphy, position and questionnaires [86-89]. TST has also been obtained by eliminating episodes with poor signal quality [67, 81, 85, 88, 90-92]. Studies assessing the diagnostic accuracy of sleep time estimation have shown improved agreement by removing periods of probable wakefulness based on heart rate, breathing pattern, movement, oximetry and activity [70], or by using an algorithm to identify sleep and wake periods based on single-lead EEG, airflow, actimetry, snoring, suprasternal pressure and thoracoabdominal belts [71] and a combination of movement and respiratory signals [93].

PAT devices have an algorithm that uses movements (actigraphy) for sleep/wake detection [32, 94-101]. Moderate agreement for sleep/wake classification has been shown [32, 34]. Manual editing can improve estimation of REM sleep duration [102].

For a detailed overview on methods, used to estimate sleep time, the reader is referred to the online supplement (table e24).

Limitations and remarks on estimating total sleep time

There is no standardised definition to evaluate TST. TRT and thereby, TST, may vary as the device can be manually turned on and off when getting into bed or automatically at pre-specified times. The AASM recommends using the term *"monitoring time"*, defined as TRT minus artefact periods and awake time determined by actigraphy, body position, respiratory pattern or patient diary, as the denominator to calculate respiratory indices [103]. There is little evidence to advocate one method over another. These methods will have the highest impact in cases of low sleep efficiency, in the presence of artefacts or short sleep time and should be validated in different patient populations.

Technical Standards for estimating total sleep time

Evidence suggests that *monitoring time*, incorporating removal of artefact and estimated wake periods from TRT, is appropriate for use as the denominator to calculate event indices using Type III devices. Methods used for estimating sleep time will vary by device and thus it is important to clearly state these methods in clinical reports and research studies and to highlight that the reported TST is estimated (events per hour of monitoring time or events per hour of estimated total sleep time).

Scoring criteria for respiratory events

Evidence overview

Apnoea

Most authors have used a 90% reduction in flow using nasal pressure [62, 68, 69, 72-75, 86-88, 93] or thermistor [82, 84, 85] for ≥10 seconds. Cessation of airflow, for ≥10 seconds [80, 104, 105] or without time specification [63, 66, 76, 79, 81, 83, 92], was also used for obstructive apnoea; 80% reduction has been used for automatic scoring [69].

Hypopnoea

Hypopnoea scoring criteria have varied over time for PSG and Type III studies, (see online supplement table e2 a+b). The major limitation in Type III studies is the inability to undertake arousal scoring.

Vat et al. [60] evaluated four different Type III hypopnoea criteria (3% or 4% desaturation with or without-pulse-wave amplitude (PWA) drops as a surrogate arousal). The best diagnostic accuracy for mild and moderate OSA was shown using the hypopnoea criterion requiring ≥3% desaturation without PWA drops. Incorporation of PWA drops only added accuracy in detecting severe OSA. The lack of diagnostic accuracy improvement for mild-to-moderate OSA was attributed to the poor correlation between PWA drops and EEG arousals. Xu et al. [86] compared two different Type III hypopnoea criteria (3% or 4% desaturation) to two different PSG hypopnoea criteria (3% desaturation + arousal or 4% desaturation + arousal). The mean difference in AHI between the Type III and the PSG equivalent was -1.2 and -1.4/h respectively, with narrow limits of agreement. At higher values the Type III scoring resulted in larger underestimates of PSG AHI. Ayappa et al. [81] explored two different hypopnoea scoring techniques: 50% flow reduction + 4% desaturation and 50% flow reduction + 1% desaturation + surrogate arousal and reported good correlation with equivalent PSG indices.

Respiratory event type

A limited number of studies [74, 79, 83, 84, 86, 89, 104, 106] showed separate results for scoring central, obstructive and/or mixed apnoeas. Standard criteria were used to score these different apnoea types: presence of respiratory effort for obstructive apnoeas, absence of respiratory effort for central apnoeas and absence of respiratory effort at the beginning and appearance of effort during the latter part of the respiratory events for mixed apnoeas. Most of these studies showed similar or good agreement for the scoring of central and/or mixed apnoeas compared to obstructive apnoeas. One study calculated an 'obstructive ratio', defined as the obstructive apnoea index per apnoea index, also reporting good correlation with the ratio from PSG [106]. An even lower number of studies reported a distinction between obstructive and central hypopnoeas [83, 86]. Criteria used for the scoring central hypopnoeas were: absence of snoring, flow limitation and paradoxical movement of the chest and abdomen. Nagubadi et al. reported better accuracy of the Type III device in hospitalized patients that did not have significant CSA.

Other respiratory events

PAT studies do not include any apnoea or hypopnoea scoring criteria [32, 34, 94-97, 99-102]. Respiratory events are derived from attenuation of the peripheral arterial tone (PAT) signal, accompanied by heart rate increase and oxygen desaturation at the end of a 'respiratory event' [32, 101, 102].

For a detailed overview on scoring criteria for respiratory events, the reader is referred to the online supplement (table e24).

Limitations and remarks on scoring respiratory events using Type III devices

For scoring an apnoea, the majority of studies required ≥90% reduction in airflow in line with the current AASM standard [39]. Of note, most studies used nasal pressure to detect apnoea, not a thermistor, which may result in event misclassification [107]. Although hypopnoea rules are variable, the majority of studies used the current AASM recommended hypopnoea definition for Home Sleep Apnoea Testing (HSAT) [103], where hypopnoeas require ≥30% airflow reduction and ≥3% desaturation. The AASM recommended hypopnoea rule for PSG requires ≥3% desaturation or EEG-based arousal. Since EEG-based arousal cannot be scored using Type III devices, there is likely to be a reduction in the number of events per hour of monitoring time or estimated total sleep time vs. PSG AHI.

Comparison studies against PSG suggest the inclusion of a surrogate arousal measure does not substantially improve agreement and diagnostic accuracy beyond that obtained using current AASM recommendations [60, 90].

Technical Standards for scoring respiratory events using Type III devices

At present, the recommended AASM scoring rules for apnoeas and hypopnoeas are appropriate. Although hypopnoeas defined using an arousal during PSG will not be scored during Type III recordings, there is no compelling evidence to use surrogate arousal measures.

Methods used for arousal scoring

Evidence overview

Since Type III devices do not record EEG, it is not possible to score respiratory events based on the presence of EEG-based arousal. Studies comparing Type III devices to PSG have used alternative methods to detect the presence of arousals [60, 64, 81, 90, 93, 105]. Methods include:

- A combination of changes in head position, pulse rate and snoring sounds [81, 105].
- Pulse oximetry derived heart rate increase [64].
- Sudden increase in amplitude or frequency of airflow or respiratory bands [90].
- Pulse wave amplitude drops [60].
- Body movement indicated by an abrupt change in thoraco-abdominal signals [93].

Conclusions based on comparison to PSG are limited in that studies have compared Type III scoring to outdated or custom scoring criteria [64, 81, 105], have used atypical OSA or population based subject groups [60, 64], or have compared Type III studies to PSG on a separate night [64]. Furthermore, comparisons with automatic algorithms [81, 105] are problematic, as algorithms may be updated without notification. The studies of Masa et al. [90] and Vat et al. [60] utilised respiratory event scoring criteria equivalent to current AASM recommended standards [103] with manual scoring of simultaneous Type III and PSG recordings. Both studies reported minimal benefit in incorporating surrogate arousals into event scoring definitions.

For a detailed overview on methods used for arousal scoring, the reader is referred to the online supplement (table e24).

Limitations and remarks on scoring arousals using Type III studies

Limitations noted for surrogate arousal methods include:

- Movement based methods may miss brief arousals without movement [93, 105],
- Heart rate methods may be affected in patients with heart disease, autonomic neuropathy or on a beta blocker [105].
- Kinoshita et al. [95] reported that arterial stiffness due to aging may attenuate the accuracy of PAT measurements.

It is difficult to draw conclusions about the superiority of one method over another, as there are: (i) no direct comparisons of methods in a single study, (ii) limited direct comparisons between surrogate and PSG scored arousals, and (iii) no studies assessing scorer reliability.

Technical Standards for scoring arousals using Type III devices

Inability to score EEG-based arousals is considered a limitation of Type III devices, resulting in inability to score events that result in sleep disturbance without, or with minimal oxygen desaturation. Although several different surrogate arousal detection methods have been described for Type III devices, there is no evidence to determine superiority of one method over another, and very limited evidence to support general use.

Scoring of oximetry

Evidence overview

The presence of ≥3% or ≥4% desaturations has typically been used to score hypopnoeas, according to the different hypopnea scoring rules [60-77, 79-81, 83-93, 104, 106]. To et al. [105] also included 1% desaturation if the event was accompanied by changes in pulse rate, head position or snoring sounds, which implied arousals. The PAT devices use an incorporated algorithm to score respiratory events using 3% and 4% desaturations [32, 34, 82, 94-102].

Chang et al. [87] found lower oxygen saturation values in Type III recordings compared to simultaneous PSG in COPD patients, emphasizing that different pulse

oximeters could influence oxygen saturation findings and clinical decision making. Oxygen desaturation index (ODI) was not significantly different from PSG despite a denominator difference in total sleep time in the order of 90 minutes. Polese et al. [79] showed no difference in oxygen saturation measures in elderly patients between PSG and simultaneous portable monitoring (PM). For both studies, SpO_2 differences between home PM and PSG were explained by different oximeter technology, but also by possible artefacts impacting home oxygen saturation values [79, 87]. Aurora et al. [74] showed a high correlation between automated and manually scored ODI values for two devices. Bridevaux et al. [76] showed almost perfect agreements between ODI scores of different observers and automated scores.

For a detailed overview on scoring of oximetry, the reader is referred to the online supplement (table e24).

Limitations and remarks re: scoring oximetry using Type III devices

Due to relative measurement simplicity, it is likely that there is better agreement for oxygen saturation measures compared to the respiratory event index between PSG and Type III recordings. However, different oximeter technology and artefact can lead to significant differences in oxygen saturation findings, particularly in patients with comorbidities. Additionally, the same issues impacting the respiratory event index regarding the denominator will also influence ODI. For ODI, although the AASM oxygen desaturation definition is not well defined, the AASM recommend using the term *monitoring time*e as the denominator [108] and the term *per hour of estimated total sleep time* could also be used

Technical Standards for scoring oximetry

The use of *monitoring time in hours* or *estimated total sleep time in hours* is appropriate for use as the denominator to calculate ODI, as well as mean values and percentages of time with oxygen saturations less than a particular threshold, using Type III devices. Be aware of differences across devices.

Utility of Type III devices in comparison to Polysomnography (PSG) for diagnosing Sleep-disordered Breathing in Adults

The literature search retrieved 914 references. After abstract and text screening, 184 references remained. Based on further evaluation of the reference lists of these 184 references, 35 references were included (see online supplement figure e3).

Diagnostic accuracy of Type III devices in sleep disordered breathing

Evidence overview

Table e25 summarises the key results from prospective, single-blind studies published from January 2007 to November 2021 comparing commercially available Type III devices with PSG in both attended (simultaneous with PSG) and unattended settings [69, 72-74, 78-82, 86, 87, 95, 96, 98-101, 104, 105, 109-122]. Sensitivity of in-lab PSG studies to detect apnoeas and hypopnoeas at various cut-offs compared to simultaneous attended studies using Type III devices ranged from 100-80%, and specificity from 0-100%. When comparing Type I and Type II studies (PSG) to home/unattended Type III studies, the diagnostic sensitivity ranged from 96 – 74% and specificity 88-25% (dependent on AHI cut-off value). Comparing the number of respiratory events sored using the same rules in studies with Type III devices vs. in-lab PSG demonstrated both under-and-over-reporting of severity of sleep disordered breathing. Type III device to manually scored PSG respiratory event indices also varied according to population examined, type of device, whether autoscoring was used and whether the studies were conducted simultaneously or separately in time.

For a detailed overview of the diagnostic accuracy of Type III devices in sleep disordered breathing, the reader is referred to the online supplement (table e24).

Limitations and Remarks regarding the diagnostic accuracy of Type III devices

There were significant differences across commercial devices in terms of number of sensors utilised as well as the AASM scoring rules over time. Airflow, heart rate, oximetry and respiratory effort were considered integral to acquiring and scoring

sleep-disordered breathing events. Although classified as a Type III device, the PAT device lacks measurement of airflow and one study in over 500 patients suggested that inbuilt autoscoring systems alone would result in 30-50% misclassification of OSA [118]. Since 2007, there have been no published data on severity classification of sleep disordered breathing using Type III devices. Previous studies using older equipment and devices (again, not standardised) have suggested that an A+H per estimated hours asleep (or hours in bed) of >15 was consistent with a diagnosis of moderate to severe sleep disordered breathing [123]. However, Type III devices showed reasonable diagnostic sensitivity and specificity for adults with a high pre-test probability of OSA in attended settings even in the presence of co-morbidities [Table 1]. Manual scoring was recommended by authors who compared manual to automatic scoring [69, 86, 118]. Algorithms for automated scoring were not disclosed. In all cases, Type III device data led to either over- or underestimation of the total number of breathing disturbances but this was not always significant. Unattended/home Type III studies resulted in significantly lower sensitivity and specificity for detecting sleep disordered breathing and higher technical failure rates (data loss ranging from 3.5% to 61%).

Technical Standards for optimising the diagnostic accuracy of Type III devices

The recommended minimum number of signals to score respiratory events accurately using current AASM criteria include heart rate, oximetry, nasal airflow signals, respiratory effort bands [12]. A position sensor should be used to differentiate supine from non-supine respiratory event severity. Peripheral arterial tonometry does not measure airflow and may lead to misclassification of OSA at higher and lower rates of sleep disordered breathing. Its utility is likely greatest in a younger population with high pre-test probability of OSA and no significant comorbidities as a screening tool. Diagnostic accuracy of OSA severity is significantly lower when using Type III devices in an unattended setting and failure rates can be high. Keeping a record of study failures, reasons for failure and information on study quality is recommended. Manual scoring is recommended, and manual editing of automated scoring programs should be possible. Most studies suggest that the sensitivity and specificity for diagnosing OSA in an attended setting is sufficiently high with an AHI >10 irrespective of scoring criteria utilised. The TF agrees that an in-lab PSG/attended PG is required to ensure diagnostic accuracy (determination of sleep efficiency) in

patients with no or mild OSA on a home-based Type III study but high clinical probability of OSAHS.

The term AHI should not, by virtue of the absence of the EEG, be used to describe the summary of breathing events acquired using Type III devices. More suitable terms include one of the following: *apnoeas+hypopnoeas per estimated hours asleep* [124], *respiratory events index* (*per estimated hours asleep*) (REI) [108] or *apnoeas +hypopnoeas per estimated hours of monitoring time.*

Patient and Health care professional experience of using and scoring Type III devices.

Evidence overview

Patient perspective

Most studies reviewed were undertaken in patient populations presenting at a sleep centre with a raised pre-test probability of OSAHS, predominantly male, predominantly middle-aged (30-60 years) and with an average body mass index (BMI) of 30kg/m^2. No studies were naturalistic; all were part of a trial with inherent selection bias. Three studies were done in a group of patients with COPD [78, 87, 120]. Three studies included patients with heart failure [74, 110, 124]. One study examined people with neuromuscular disorders [109] and one study was undertaken in pregnant women [100]. Most of the studies undertaking home Type III studies provided information to patients on how to wear the Type III devices in an unattended setting/home. Six studies requested patient feedback on the experience [78, 87, 109, 114, 116, 121].

Scoring Type III studies

Evidence overview

Level of qualification of the scoring staff ranged from experienced technician to a formal North American qualification of registered polysomnographic technician (RPSGT) [72, 73, 101, 112, 114, 117, 119, 120, 124]. No studies commented specifically on whether the Type III software was user-friendly. Only two studies undertook intra-and inter-scoring concordance [72, 112]. There was no mention in any study on how equipment was cleaned and re-used, and the specific infection control procedures required by type of device used. Manual scoring or manual editing of automated scoring improved diagnostic accuracy compared with automated

scoring alone. Ideally, Type III devices should be capable of displaying the raw data for review by the scorer, in order to allow assessment of the quality of the data. Data from the entire duration of the study should be available to review, rather than just an automated summary of the data.

Economic aspects of using Type III studies

Evidence overview

Masa et al. [91] documented costs and found that it was at least 40% more expensive to do PSG than unattended Type III studies for equal efficacy; patient costs were higher for unattended Type III studies compared to PSG. No other studies examined cost to the sleep service overall, impact of technical failure on diagnostic pathway, time taken to hand out/mail out a Type III device, give patient-specific instructions and support patients undertaking home studies or the time taken to score or repeat a study in either an attended or unattended setting. Formal assessment of the economic impact of using Type III devices using appropriate tools e.g. EQ5D, calculation of QALYs was not undertaken in any study and has not been reported on since 2007.

Limitations and remarks concerning population applicability and practical aspects of performing Type III studies

Published information on the acceptability, sensitivity, and specificity of Type III studies in populations other than obese, middle-aged men with symptoms consistent with OSA is limited. The economic aspects of high failure rates in unattended Type III studies have not been explored in any depth. Information on user-friendliness and scoring ease was not cited in the published literature but should be a criterion for choosing a Type III device for clinical use.

Technical Standards on applicability and practical deployment of Type III devices in a clinical setting

When incorporating Type III devices in the diagnostic pathway of a sleep centre, all aspects of using the device including quality of the sensors and scoring software, disposable and non-disposable consumables, cleaning protocols, patient acceptability and device reliability must be considered. Patients should be asked to document their experience with the device, the quality of their sleep, any disruptions,

or difficulties with using the device on the night of their study. Patients should be advised of the risk of having to repeat the study or undertake PSG to make an accurate diagnosis particularly if the study is unattended. All Type III studies undertaken in subjects out with a published demographic must be assessed strictly in the clinical context in which the study is being undertaken. Competence in scoring Type III studies should be standardised at least nationally through specific, accredited sleep training pathways.

Manual scoring or manual editing of automated scoring of limited studies is recommended in order to improve diagnostic accuracy. Finally, the application, interpretation, and follow-up of Type III studies are best handled by experienced sleep healthcare providers.

Type III Devices for diagnosing Sleep-disordered Breathing in Children

The literature search retrieved 981 references. After abstract and text screening, 45 references remained. No articles were excluded after further evaluation of the references (see online supplement figure e4).

Differences across currently available Type III devices and technology utilised; specifications required to acquire signals in a regulated fashion.

Evidence overview

Technical differences across currently available Type III devices that have been utilised in children are summarised in the online supplementary table e26. Type III devices are usually set up in the child's home by trained staff or by the parents. Repositioning of sensors is not possible during the night if the corresponding signals are lost or they are inadequate for analysis.

A few Type III devices have been compared to full PSG in paediatric patients (see online supplement table e27) [125-131]. Other reports have included results of respiratory PG with Type III devices in paediatric patients without comparison to PSG (see online supplement table e28) [132-167]. Full PSG equipment is used in many paediatric sleep centres across the world for performance of respiratory PG by omitting placement of the EEG, EOG and EMG channels [131, 138, 141, 145, 161,

162]. Devices with only two channels i.e. airflow via nasal pressure transducer and pulse oximetry have also been used in paediatric populations [127, 152, 155].

Michelet et al. demonstrated that over 80% of PGs performed either in the hospital or at home are interpretable and the main reasons of non-interpretability were: poor SpO_2 signal (80%), poor nasal cannula signal (41%), poor abdominal belt signal (29%), and poor thoracic belt signal (18%) [133]. Scalzitti et al showed that in-lab portable monitor set-ups were technically acceptable (term not defined by the authors) in 93.9% patients and 75% had interpretable data on 3 channels for at least 360 minutes. For PGs completed at home, 88.9% were technically acceptable, and 67% had interpretable recordings [130]. In a retrospective investigation by Gudnadottir et al., the requirement of 3 hours of valid data for an *acceptable* study was not fulfilled for nasal airflow in 40% and for SpO_2 in 19% of cases, while in 11% of patients both parameters were missing [168]. Moreover, in 5% of PGs other problems were noted, like the caregivers misunderstanding the instructions or the equipment batteries malfunctioning [168].
Scalzitti et al. studied 33 children with simultaneous laboratory PSG and PG (portable monitor) [130]. Twenty patients also underwent home studies, with 16 having 2 nights of monitoring. AHI by PG performed in the sleep laboratory or at home was significantly different from that obtained by PSG. The sensitivity of the portable monitor for diagnosing OSAS was best for in-lab use.

Lesser et al. used a portable device to screen for OSAHS in obese adolescents in the sleep laboratory. The device had a high negative predictive value for ruling out OSAHS while automatic scoring using the device software was found to be as accurate as manual scoring in this age group [128].

For a detailed overview on differences across currently available Type III devices and technology utilised, the reader is referred to the online supplement (tables e26 – e29).

Limitations and Remarks on the heterogeneity of Type III devices using in paediatric sleep medicine

Approximately 70% of PGs performed at home are interpretable and this frequency is higher when the study is performed in the sleep laboratory. Most common technical problems are poor SpO_2 or nasal airflow tracings.

Technical Standards on the use of Type III devices in the diagnosis of sleep disordered breathing in children

Type III devices can be used at home for the diagnosis of sleep disordered breathing in children, with a high rate of success in obtaining adequate signals. Type III devices and PSG systems without EEG can be used, when more advanced equipment is not available or in an attempt to reduce the time required for setting up and interpreting the sleep study, respectively.

Type III devices should incorporate respiratory inductance plethysmography (RIP) technology for detecting thoracic and abdominal wall movements. This approach has the added benefit of RIP flow tracing as loss of airflow signal is the most frequently encountered problem in children while performing PG at home. The addition of actigraphy to the PG channels might increase the reliability of the obtained tracings and facilitate recognition of wakefulness.

<u>Scoring criteria for sleep related breathing disturbances using Type III devices in children.</u>

Evidence overview

Various rules have been used for automatic or manual scoring of obstructive, central and mixed apnoeas and hypopnoeas in Type III devices, but in most cases the 2012 or 2007 American Academy of Sleep Medicine (AASM) scoring rules have been applied (see online supplement tables e27 and e26) [125-128, 131, 133-136, 140, 143, 145-149, 156, 158, 159, 169-172]. Automated scoring of PG in children was reliable only for central apnoeas in a study by Blanc et al. [135]. In another study by

Orntoft, apnoea-hypopnoea index (AHI) was consistently overestimated by automatic analysis [166]. In contrast, Masoud et al. demonstrated that automatic analysis after exclusion of poor tracings had a very good sensitivity with low specificity for OSAS defined as an AHI ≥1.5 /h (95.5% and 66.7%, respectively) [129]. Lesser et al. also demonstrated that automatic scoring and manual analysis of PG provided similar results [128].

Gudnadottir et al. evaluated the interrater reliability of PG scoring [168]. They also explored whether the calibrated respiratory inductance plethysmography (RIP) flow signal could be used for the scoring of respiratory events when the airflow tracing is unreliable. They reported moderate agreement between the scorers when nasal airflow was present while the scoring of respiratory events alone based on the RIP flow signal was scorer-dependent.

Since total sleep time can only be estimated with respiratory PG, total recording time is usually used in the denominator for calculating the frequency of respiratory events [126]. Total sleep time can be approximated when respiratory PG is completed in the hospital with PSG equipment by using the sleep technologist's notes and the video recording of the sleeping child [141, 173].

For a detailed overview on scoring criteria for obstructive, central and mixed apnoeas and hypopnoeas and terminology with Type III devices in children, the reader is referred to the online supplement (tables e26 – e29).

Limitations and Remarks on scoring sleep related breathing disturbances using Type III devices in children.

When PG is performed, it is unknown whether the child has had adequate sleep time, and in particular, REM sleep during which most obstructive events may occur, because sleep scoring is not possible. The inability to score arousals may lead to underestimation of the number of hypopnoeas and central apnoeas associated with arousals from sleep (without accompanying hypoxemia). Moreover, the lack of arousal scoring results in inability to evaluate the degree of sleep fragmentation. Use of total recording or calculated sleep time instead of the actual total sleep time leads

to underestimation of the various respiratory event indices because time is included in the denominator.

Technical Standards sleep related breathing disturbances using Type III devices in children.

Very limited evidence indicates satisfactory correlation between the AHI obtained from automatic analysis and AHI calculated by manual scoring of the tracing obtained using a type III device. The task force agrees on manual scoring, based on the current AASM scoring rules in order to limit over-or-underestimation of the RP parameters.

Cut-off values for the frequency of apnoeas and hypopnoeas scored using a Type III device along with sensitivity and specificity for diagnosing sleep disordered breathing in children.

Description of the frequency of obstructive and mixed apnoeas and hypopnoeas using a Type III device

The frequency of obstructive and mixed apnoeas and hypopnoeas is calculated by dividing the total number of scored obstructive and mixed apnoeas and hypopnoeas by the total recording time or total calculated sleep time [141]. It should be noted that the OSAHS severity category (mild, moderate or severe) cannot be defined using the traditional AHI 5/h,15/h, and 30/h cut-off values applied in adults. In the ERS Statement on the Diagnosis and Management of Obstructive Sleep-Disordered Breathing in 2- to 18-year old children, the AHI cut-off values 1/h and 5/h have been proposed for defining mild and moderate-to-severe OSAHS, respectively [174].

Evidence overview

A small number of studies have compared various Type III devices used at home or in the sleep laboratory against fully attended PSG (see online Supplement table e27) [125-131]. In a study by Alonso-Alvarez et al., scoring of recordings obtained from a Type III device systematically overestimated the actual AHI [126]. In two other paediatric studies, overestimation of the AHI was attributed to (i) events scored

during wakefulness and (ii) pseudo-events related to either reduced amplitude of the nasal airflow channel resulting from mouth breathing and/or artifactually reduced flow post-arousal [125, 127]. Alonso-Alvarez et al. used the eXim Apnea polygraph in combination with the 2007 AASM scoring rules to evaluate otherwise healthy children with OSAHS symptoms [126]. The investigators showed that an obstructive apnoea-hypopnoea index (OAHI) ≥3 episodes/h using in-home PG had a sensitivity of 72.5% and specificity of 90% for detecting OAHI ≥1 episodes/h in polysomnography. In addition, an OAHI ≥6.7 episodes/h using PG detects OAHI ≥5 episodes/h in PSG with sensitivity 81.8% and specificity 92.9%.

In a study by Ikizoglu et al utilizing PSG as the gold standard, the NoxT3 portable monitor used at home had a high sensitivity (100%) for detecting an AHI ≥1 episode/h in children with Down syndrome but very low specificity, positive and negative predictive values (<40%) [125]. The monitor also overestimated the true AHI in this patient group; an AHI ≥3 episode/h in PG was predictive of an AHI ≥1 episode/h on PSG with a sensitivity 100% and specificity 85%. Masoud et al. reported strong agreement between AHI obtained from PSG and the respective index calculated from PG [129].

Tan et al. compared attended PSG in the sleep laboratory without EEG channels (respiratory PG) to fully attended PSG [131]. They found that the AHI is underestimated mostly due to underscoring of hypopnoeas which are accompanied by arousals without desaturations [131].

Various cut off values for defining OSAHS have been adopted in studies of Type III devices without comparison to PSG (Online Supplementary Table e29). Brockmann et al obtained PG recordings in 37 healthy full-term infants at the ages of 1 month and 3 months that were analysed using the 2012 AASM scoring rules [170]. The 95th percentile for the frequency of obstructive and mixed apnoeas and hypopnoeas per hour of estimated sleep time was 5.8 episodes/h at 1 month and 3.4 episodes/h at 3 months of age. The respective values for the oxyhemoglobin desaturation (≥3%) index were 24.9 episodes/h and 24 episodes/h. In a Canadian cohort including healthy infants who underwent PG at the age of 1 year, the 90th percentile was 0.5 episodes/h for the obstructive apnoea index, 7.1 episode/h for the central apnoea

index, 15.8 episode/h for the AHI (obstructive, central and mixed apnoeas and hypopnoeas per hour of estimated sleep time), 10.7 episodes/h for the oxygen desaturation (≥3%) index [134].

For a detailed overview on cut-off values for the frequency of apnoeas and hypopnoeas scored using a Type III study along with sensitivity and specificity for diagnosing sleep disordered breathing in children, the reader is referred to the online supplement (tables e26 – e29).

Limitations and Remarks regarding cut-off values for diagnosing severity of sleep disordered breathing scored using a Type III study in children.

The appropriate cut-off value of the frequency of apnoeas and hypopneas for diagnosing OSAS with a Type III device is affected by its technical specifications and the setting in which the study is performed (attended in-laboratory vs. unattended at home). Thus, the measured AHI may overestimate or underestimate the real AHI. As a result, the AHI cut-off value to define OSAS in studies involving type III devices varied from 1/h to 5/h (Table e29).

Technical Standards for diagnosing severity of sleep disordered breathing scored using a Type III study in children.

The term AHI should not, by virtue of the absence of the EEG, be used to describe the summary of breathing events acquired using Type III devices. More suitable terms include one of the following: apnoeas+hypopnoeas per estimated hours asleep [124] respiratory events index (per estimated hours asleep) (REI) [108] or apnoeas +hypopnoeas per estimated hours of monitoring time.

PG in children performed using a Type III device at home provides a frequency of apnoeas and hypopnoeas per estimated hours of monitoring time which is greater than the true AHI i.e. AHI obtained from full in lab video PSG. This discrepancy has been attributed to (i) events scored during wakefulness and (ii) pseudo-events related to either reduced amplitude of the nasal airflow channel during mouth breathing and/or artifactually decreased flow post-arousal. In contrast, when PG is performed in

the sleep laboratory using a PSG system without recording the EEG, EOG and EMG channels, the calculated frequency of apnoeas and hypopnoeas is lower than the true AHI obtained from full PSG for two main reasons: (i) underscoring of hypopnoeas that are accompanied by arousals but not desaturations; and (ii) use of total recording or calculated sleep time instead of the actual total sleep duration leads to underestimation of the various respiratory event indices because time is included in the denominator.

The frequency of apnoeas and hypopnoeas per estimated hours of monitoring time ≥3 /h in a Type III device-based study is a reasonable predictor of AHI ≥1 episode/h in PSG.

Conclusion

Evaluation of the available evidence has shown that there are no universally defined technical standards in place for one of the most frequently used technologies in sleep practice in adult and particularly in paediatric populations. Application of the equipment, acquisition of signals and scoring of the signals and terminology for reporting is also not standardized, leading to huge variation in outcomes and treatment choices across centres which may carry significant financial implications. This is of importance not only to the individual patient but also for research studies, epidemiological studies, and the health economy overall. As diagnostic tools used exclusively for capturing sleep disordered breathing during sleep, Type III devices are at their most specific and sensitive when there is high a pre-test probability clinically of such a disorder being present. For patients with lower pre-test probability of sleep disordered breathing as the source of their symptoms, an unclear differential diagnosis or suspected additional sleep disorders, PSG remains the diagnostical test of choice. Type III monitors are a diagnostic tool that must be tailored to a specific diagnostic problem. As such, they are also subject to additional considerations determining their use, including access to PSG, waiting list times, concerns regarding operator error in unattended settings, the preferences and practices of the medical institution and reimbursement and insurance issues.

With regard to the type of device deployed clinically (in the context of the almost universal adoption of AASM criteria for scoring respiratory events [103]), there are additional questions regarding the number of sensors required to record data. On the basis of this statement, and a recent review of the literature in a similar vein, it is currently suggested that a minimum of 3 sensors that attach directly to the body are necessary to obtain the minimal physiological signal dataset required to accurately score respiratory events [175].

At present, generalising the cut-offs for classifying mild, moderate or severe OSA using unattended Type III studies remain unclear in both adult and paediatric populations and may also be specific to each device and the setting in which the study is undertaken. In populations with moderate to high pre-test probability of OSAHS, no unstable co-morbid conditions as well as reasonable sleep efficiency in an attended setting, diagnostic capability is reasonably reliable. The nomenclature for reporting the number of breathing pauses per estimated hours asleep needs to be differentiated from the AHI which should strictly remain in use for PSG studies only. We recommend either *apnoeas+hypopnoeas per estimated hours asleep*, or *respiratory events index per estimated hours asleep* (REI) as suggested by the AASM (2020)[108] or *apnoeas +hypopnoeas per estimated hours of monitoring time.*

Manual scoring of events by qualified and registered sleep technologists is recommended as well as the facility to override/correct automated algorithms that are incorporated into most commercially available devices. First, the criteria can change considerably over time depending on the standards adopted by the AASM and secondly, many devices may have data signal acquisition limitations in respect of the reliability of their sensors that make their inbuilt automatic scoring algorithms unreliable.

Innovation is unstoppable. New technologies, incorporating artificial intelligence are in constant development; their adoption could contribute to improved algorithms for extracting sleep stages from ECG, pulse wave detection, respiratory dynamics and movement sensors. They could thus overcome the current weaknesses of Type III systems. An increase in the processing and integration capacity of electronic devices, as well as advances in low-power wireless communications, has also enabled the

development of unwired intelligent sensors with a wide set of applications [25-28, 176] (see e-supplement). However, consideration should also be given to the 'black box' of their unique scoring algorithms that cannot be manually examined or altered with time. Reflecting on the very disparate results recorded in the studies reviewed in this paper, standardising clinical testing protocols is to be encouraged [177]. Thorough validation of such devices and extensive testing in both adult and paediatric subjects is essential and should in the very least include power and effect size calculations, failure rates and their reasons, side-effects of wearing the devices, patient feedback as well as trials in a variety of clinical settings and populations.

The Covid-19 pandemic has had a major impact on sleep medicine [178]. Problems with hygiene (cleaning and disinfection of devices) as well as limited access to sleep laboratories have raised questions about the technical equipment utilised. For example, single-use devices (disposables) could play a role in the future; however disadvantages for the environment could be considerable. Device manufacturing companies should be encouraged to develop contactless monitoring and evaluate its efficacy in both attended and unattended settings. Although this could include 'nearables', at present they are devised largely for the consumer market and cannot be classified strictly as Type III devices. There is an urgent need for standardized telehealth options for screening, diagnosis and follow-up of patients suspected of having sleep-disordered breathing. Significant hurdles to such progress comprise legal and ethical dilemmas regarding data ownership and curation, scientific robustness in trialling new equipment, the lack of universally defined standards for physiological signal acquisition and processing and the future implications for financial resources as well as reimbursement in increasingly stretched healthcare systems. At the time of writing this Technical Standard, data on any developments concerning the above were either unavailable or outside the defined search period.

Table 1: The AASM, American College of Chest Physicians and the American Thoracic Society have divided portable monitoring into 4 types:

Type I: full attended polysomnography (≥ 7 channels) in a laboratory setting
Type II: full unattended polysomnography (≥ 7 channels)
Type III: limited channel devices (4-7 channels)
Type IV: 1 or 2 channels usually using oximetry as 1 of the parameters

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Technical Standards for using Type III devices (limited channel studies) in the diagnosis of sleep disordered breathing in adults and children

TF-2019-16

Supplementary Material

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Abbreviations

AAI	Autonomic Activation Index
AASM	American Academy of Sleep Medicine
ABD	Abdominal
ACAT	Auto-correlated Wave Detection with Adaptive Threshold
AHI	Apnoea/Hypopnoea Index
AUC	Area under the Receiver Operating Characteristic Curve
BLE	Bluetooth Low Energy
BMI	Body Mass Index
BPM	Beats per minute
CAI	Central Apnoea Index
CC	Correlation Coefficient
CI	Confidence Interval
CMRR	Common-Mode Rejection Ratio
COPD	Chronic Obstructive Pulmonary Disease
CPAP	Continuous Positive Airway Pressure
CPE	Circuit Playground Express
DAP	Decreases in the Amplitude Fluctuation of PPG
DBA	DTW Barycenter Averaging
DC	Direct Current
DTW	Dynamic Time-warping
ECG	Electrocardiogram
EEG	Electroencephalography
EIT	Electrical Impedance Tomography
EMG	Electromyography
EOG	Electrooculography
ESS	Epworth Sleepiness Scale
FENet	Frequency Extraction Network
GH	Growth
HFF	High-Frequency Filter
HI	Hypopnoea Index
HR	Heart Rate
HRP	Home Respiratory Polygraphy
HSAT	Home Sleep Apnoea Testing

Hz	Hertz
ICC	Intraclass Correlation
IQR	Interquartile Range
KHz	Kilohertz
LFF	Low-Frequency Filter
LR	Likelihood Ratio
MAE	Mean Absolute Errors
MEMS	Microelectromechanical System
MOAHI	Mixed Obstructive Apnoea-Hypopnoea Index
MPS	Mucopolysaccharidosis
mRMR	Minimal-Redundancy-Maximal-Relevance
MT	Monitoring Time
n	Number
NA	Not Applicable
NPP	Nasal Prong Pressure
OAHI	Obstructive Apnoea-Hypopnoea Index
OAI	Obstructive Apnoea Index
ODI	Oxygen Desaturation Index
OSA	Obstructive Sleep Apnoea
OSAHS	Obstructive Sleep Apnoea Hypopnoea Syndrome
OSAS	Obstructive Sleep Apnoea Syndrome
PAT	Peripheral arterial tonometry
PE	Piezoelectric
PG	Polygraphy
PM	Portable monitoring
PPG	Photoplethysmogram
PSG	Polysomnography
PtcCO2	Transcutaneous Partial Carbon Dioxide Pressure
PTSD	Post-Traumatic Stress Disorder
PTT	Pulse Transit Time
PVDF	Polyvinylidenefluoride
PWA	Pulse Wave Amplitude
PWV	Pulse Wave Velocity
QI	Quality Index

R	Ratio
RCT	Randomised Controlled Trial
RDI	Respiratory Disturbance Index
REI	Respiratory Event Index
REM	Rapid Eye Movement
RERA	Respiratory Effort-Related Arousals
RIP	Respiratory Induction Plethysmography
ROC	Receiver-Operating Characteristic
RP	Respiratory Polygraphy
RR	Respiratory Rate
SAHS	Sleep Apnoea-Hypopnoea Syndrome
SBP	Snoring and body position
SD	Standard Deviation
SSP	Suprasternal Pressure
SVM	Support Vector Machine
TH	Thermistor
TRT	Total Recording Time
TST	Total Sleep Time
USD	US Dollar
Vs	Versus
Yrs	Years

Search Criteria for all groups

[Sleep Apnea/Cardiorespiratory] search terms:

(apnoea or apnea or hypopnoea or hypopnoea or OSA or CSA or sleep disordered breathing or hypoventilation or periodic breathing or cardiorespiratory or Cheyne-Stokes or heart failure)

AND

[Type 3 devices] search terms:
(attended or unattended or Type 3 or Type III or T3 or portable or polygraphy or home or HSAT or arterial tone or pulse rate or watchpat or sleep scout)

Additional search criteria for "Technical specifications of using type III devices"

[Device specific] search terms:

(device or sensor or technology or guideline or parameter or equipment or monitor or SOFTWARE or AASM or airflow or respiratory effort or blood oxygenation or nasal pressure transductor or inductance plethysmography)

Additional search criteria for "Scoring criteria for sleep related breathing disturbances in type III devices in adults"

[Scoring related] search terms:

(scor* or measure* or criteria or guide* or evaluation or parameter or valid or Time in bed or time asleep or sleep time or recording time or arousal or desaturations or centrals or central apnoeas or central apnea or central hypopnoeas or central hypopnoea or obstructive apnoea or obstructive apnea or obstructive hypopnoea or obstructive hypopnoea or automatic or manual or AASM or rules or Oximetry)

Additional search criteria for "Type III devices in comparison to PSG for investigating sleep-disordered breathing"

[Study related] search terms:

(Limited or study or report or analysis or trial or test or RCT or diagnos*)

AND

Child OR adult

Additional search criteria for "Scoring criteria for sleep related breathing disturbances in Type III devices in children"

[Study related] search terms:

(Limited or study or report or analysis or trial or test or RCT or diagnos*)

AND

Child OR adult

Further restrictions:

Humans, English, Field area, 2007 onwards

Flow diagrams

Figure e1: Technical specifications of using Type III devices

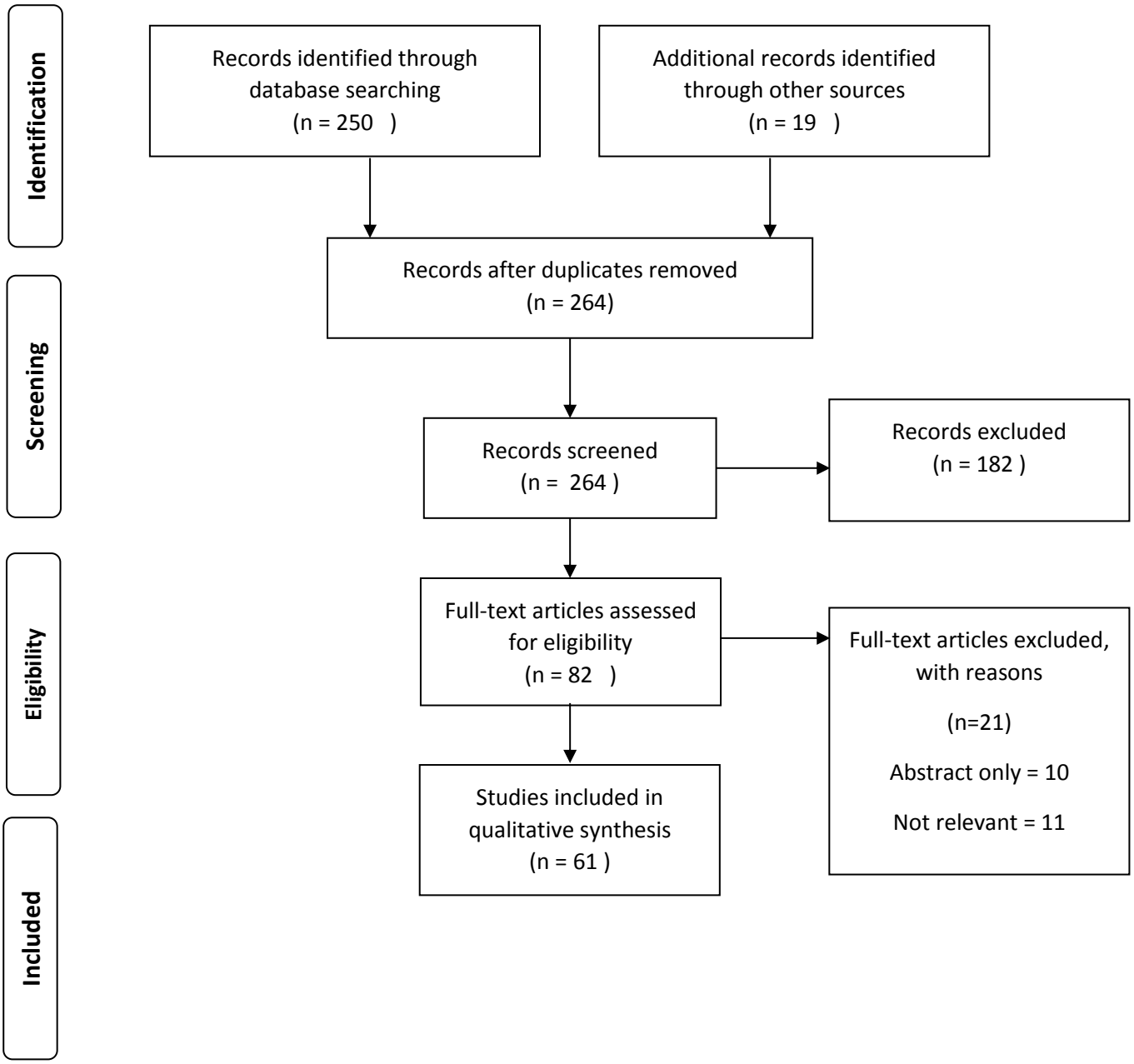


Figure e2: Scoring criteria for sleep related breathing disturbances in Type III devices in adults

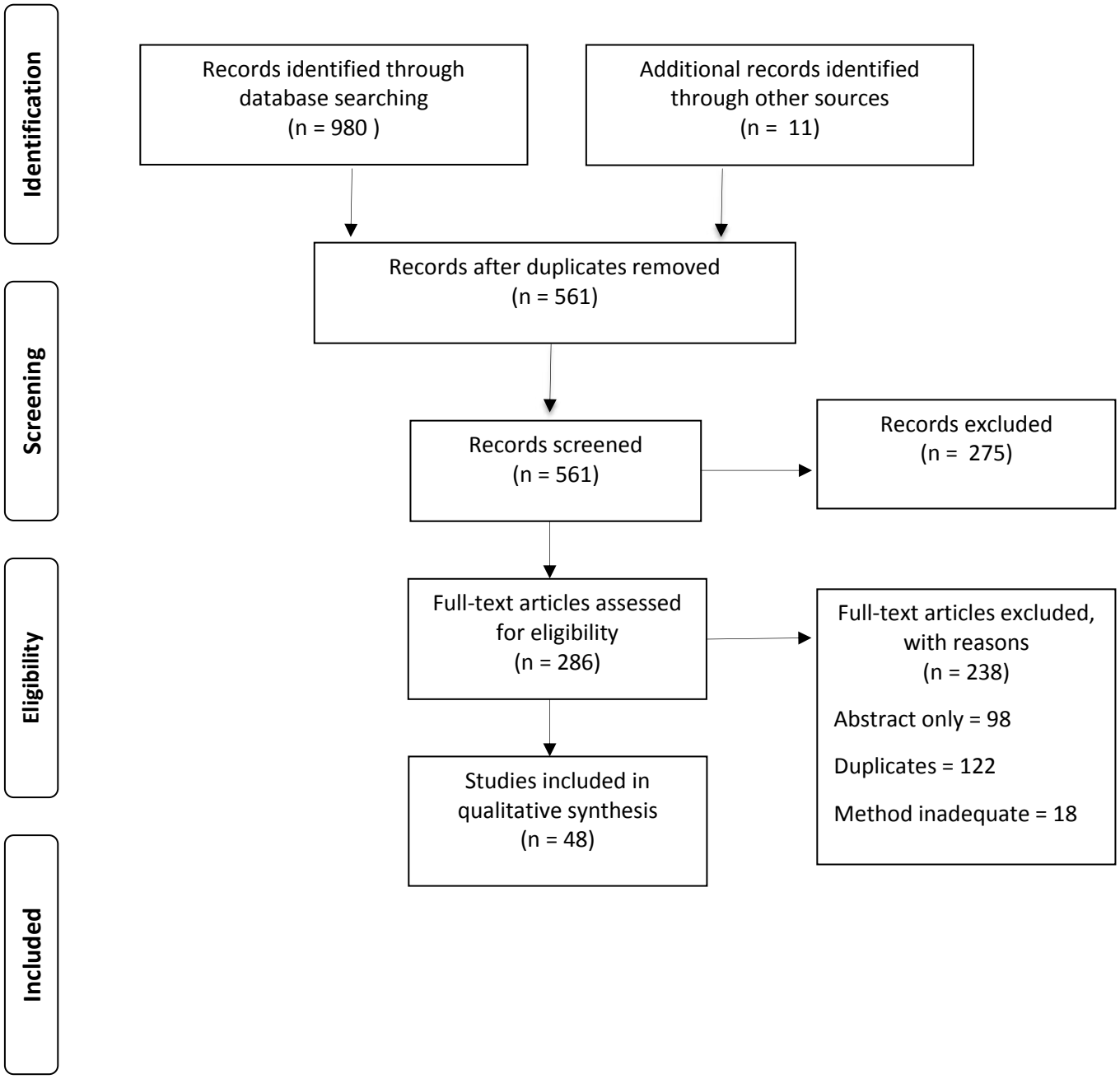


Figure e3: Type III devices in comparison to PSG for investigating sleep-disordered breathing

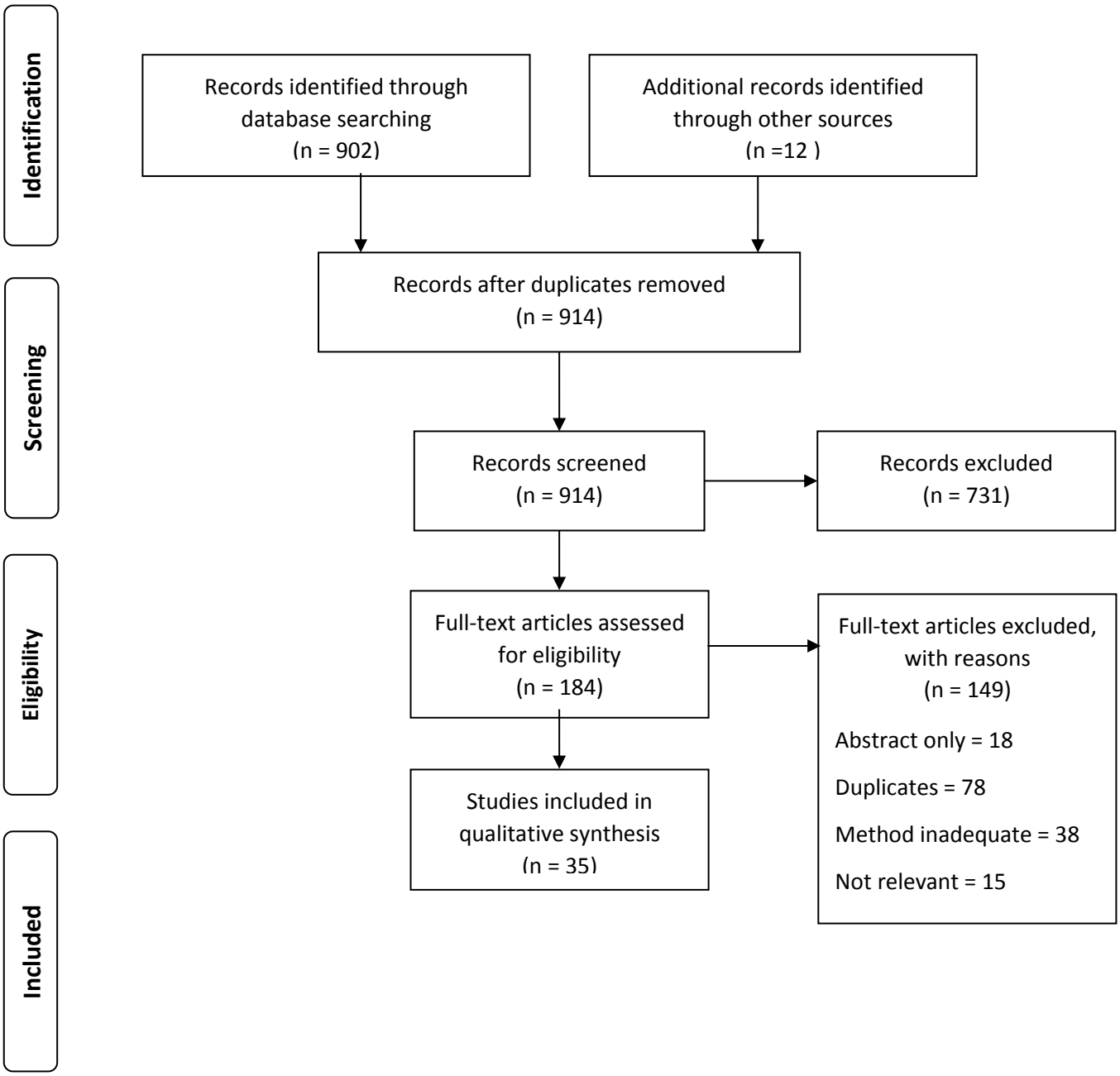
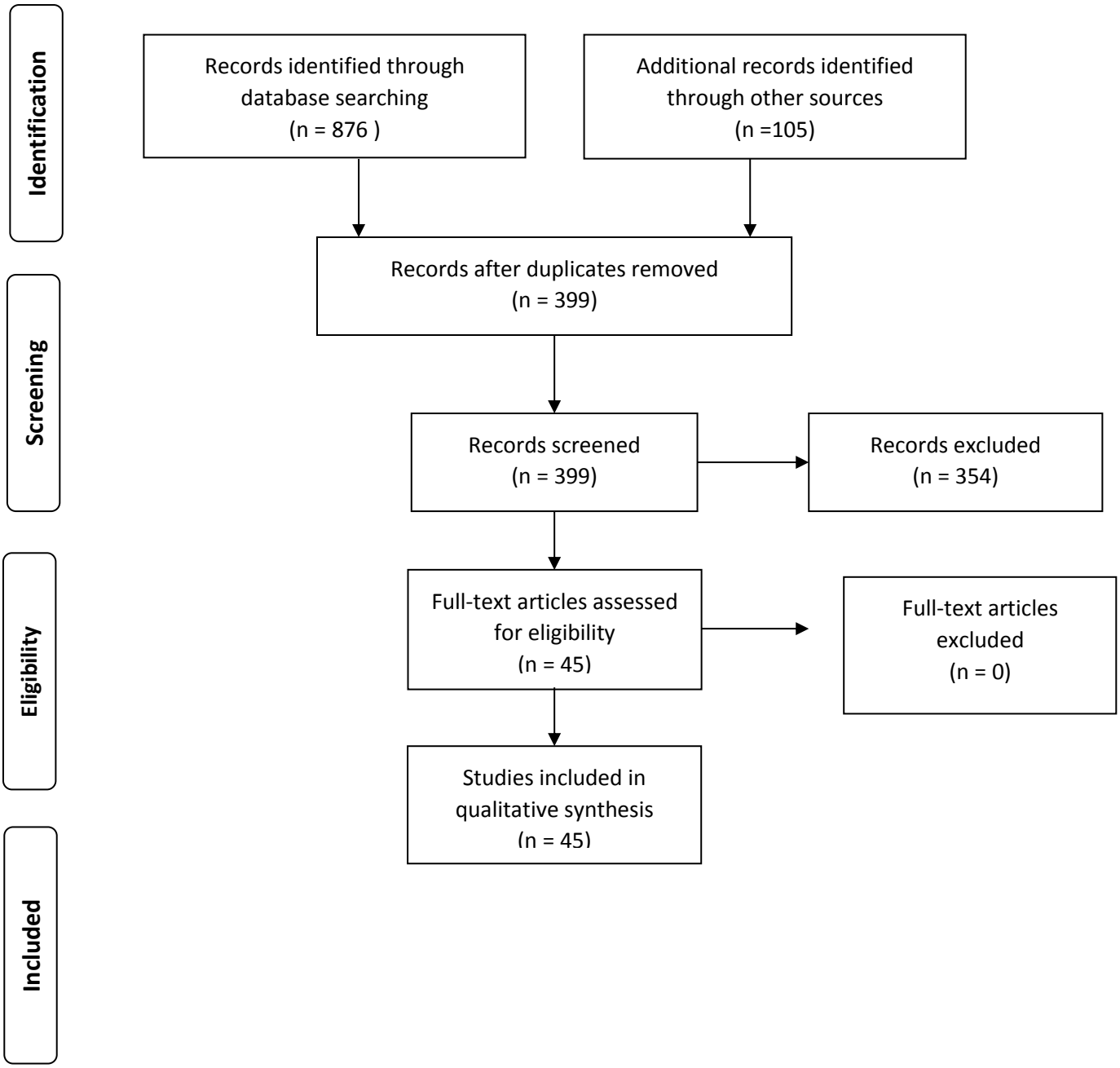


Figure e4: Scoring criteria for sleep related breathing disturbances in Type III devices in children



Additional Material on technical issues – sensors, channels, signals, sampling rates, filters

Home sleep apnea testing (HSAT), or respiratory polygraphy (RP), uses Type III recording devices or 4-channel (minimum) devices. The four channels often refer to airflow, respiratory effort, oxygen saturation, and pulse or heart rate. This refers to the concept, that one signal is one channel and one sensor. However, today with technical innovations, one sensor sometimes conveys several signals and several channels. One prominent example is the nasal cannula, which records pressure changes primarily. Derived from this one sensor, it is possible to calculated airflow, and if higher frequency vibrations are separated using adequate filtering, then snoring can also be derived. This, one sensor results in two signals and two channels. Another popular example is photoplethysmography on the finger used to derive oxygen saturation with two light sources. In addition, the pulses can be used to calculate pulse rate, which is equivalent to heart rate, normally. This gives two signals from one sensor already. Moreover, the changes in pulse amplitude vary with sympathetic tone (better known as peripheral arterial tonometry) and allow for the derivation of respiration or possibly respiratory effort. This gives a third channel or signal, as used in peripheral arterial tonometry when estimating respiratory events. Pulse amplitude variations also vary with respiratory effort from breath to breath, also denoted as pulsus paradoxus which is exploited in wearables / smartwatches when they combine information from oxygen desaturation events with pulse amplitude parameters to improve the prediction of respiratory events. Pulse wave and ECG recorded together allow for the calculation of pulse transit time, from which blood pressure can be derived. This method is used in some polygraphy devices and gives a fourth channel or signal from the same one sensor.

In 2011, a new classification system was proposed to evaluate the physiological information picked up by sensors [1]. The SCOPER (Sleep, Cardiovascular, oxygen, position, effort, and respiration) system tries to change the view on sleep recording from the type I to IV and channel counting perspective to a more physiologically oriented classification. This is more appropriate, as one sensor may deliver multiple physiological information. Primarily, this view allows for the definition of a reference standard per physiological

information (such as EEG/EOG/EMG for sleep)at the top end and a lower accuracy or surrogate measure (such as actigraphy for sleep) at the other end. Another example in the SCOPER categories is “cardiovascular”. According to the SCOPER table, the reference standard for this category would be a 12-lead ECG recording. This kind of recording is not even done as part of regular polysomnography recordings. According to SCOPER, a one-lead ECG recording is a somewhat reduced level of assessment for “Cardiovascular” and this is the standard recording approach used in clinical polysomnography recordings. However, the assessment and subsequent analysis of pulse wave as derived from oximetry, can also serve as a surrogate signal for cardiovascular information. The pulse wave can be used to derive pulse rate (a surrogate for heart rate), pulse transit time (a surrogate for blood pressure), pulse amplitude (a surrogate for sympathetic nerve activity). Similar to this “cardiovascular” example, for each physiological datum needed (e.g. respiratory flow, respiratory effort, and blood gases), a reference standard is defined and surrogates at lower levels are listed. This system is not evidence-based but provides a scheme to assess the quality of signals in terms of physiological information.

With standard electronics and early digital recordings, the definition of sampling rates was a big issue in order to find the optimal compromise between accuracy in time resolution (the higher the better) and storage capacity (the lower, the less space required). With new technology, digital storage capacity is no longer a limitation. Data transfer rates remain limitations, but not in our range of interest. Two important background principles need to be considered. First, at different steps in signal processing, different sampling rates are needed. As an example, for the calculation of oxygen saturation, first the pulse wave as obtained for the two optical wavelengths needs to be digitised. This is usually done with 100 Hz or higher sampling rates. A sampling rate of 2000 Hz is not unusual because it allows for better removal of unwanted noise from the raw optical signal. As oxygen saturation is then calculated by a microprocessor inside the sensor itself, the initial sampling rate chosen by the manufacturer is no longer visible to the user and not relevant for storage or any other evaluation. In the end, physiologically speaking, arterial oxygen saturation can only change from one heartbeat to the next heartbeat. Thinking of a heart rate of 60 beats per minute, a sampling rate of 1 Hz would be adequate for oxygen saturation. To be on the safe side, with a higher heart rate, 4 Hz or 10 Hz would definitely be high enough. Taking all the measurements and processing steps together, an overall error in oxygen saturation would be a much better-quality measure. Usually an overall error in oxygen

saturation is accepted as being +/- 2%. This is the reason desaturations are counted if they are at least 3%. The second piece of background information is that with digital pre-processing as it stands today, the concept of equidistant sampling, as assumed by the sampling rate, is not necessarily implemented in all cases. For example, even simple actimetry has a microprocessor to sample the accelerations and calculate activity. When no acceleration is detected (e.g. during slow wave sleep), then the microprocessor may fall into sleep mode with sample signals at a much lower rate, until wakefulness. Another very common example is the transmission of voice using cell phones. While the initial sound picked up by the microphone is digitised at very high sampling rates, it is pre-processed using algorithms, and only weight values of characteristic wavelets are transmitted. This results in enormous data compression and processing efficacy, not measurable by sampling rates or filter settings.

In conclusion, there are no studies which provide evidence on sampling rate or filter settings for any specific signal available. Instead, we as users of polygraphy, must define requirements on signal quality in terms of physiological measurement. One example is oxygen saturation with an accuracy of +/- 2% or respiratory amplitude for which we expect an accuracy of +/- 10%. However, neither standards nor studies are available for this kind of view on signals.

Table e 1: Sampling rates for signals according to physiological content

Signal content	**Sampling (Hz)**	**Filter settings (Hz)**
Pulse waveform	100 – 500	0.1 – 100
ECG	100 – 500	0.3 - 70
Airflow, air pressure, respiratory effort, respiratory movement	10 – 100	0.1 - 15
Oxygen saturation	4 – 10	DC – 10
Pulse rate / heart rate	4 – 10	DC - 10
Body position / activity	1	DC - 10
Optional signals		
EEG, EOG, EMG	200 – 500	0.3 – 35
Microphone / snoring amplitude (not raw sound)	10 – 100	10 - 100
Blood pressure (systolic / diastolic / mean)	4 – 10	0 -25

***

Additional material on scoring criteria for sleep related breathing disturbances in Type III devices in adults

Heading 1: Tools used to estimate sleep time

Zhao et al. compared respiratory event indices using either TST, an adapted TRT (TRT minus artefacts and periods of probable wakefulness, based on heart rate, breathing pattern, movements, oximetry and activity) and TRT (based on lights off and on) as denominator. The adapted recording time, provided a respiratory event index (REI) that was closer to PSG-AHI and reduced sleep apnoea severity misclassification from 27% to 10% [2].

Sabil et al. reported improved OSA severity classification by estimating TST with an algorithm (HypnoLighT, based on single-lead EEG, airflow, actimetry, light, snoring, suprasternal pressure and chest/abdominal belts). Using the algorithm correctly reclassified more than 50% of the patients who were initially misclassified using the TRT, with regard to OSA severity [3].

Aielo et al. reported good AHI and severity classification agreement when either actigraphy, button press or diary was used to shorten TRT, although actigraphy led to a shorter time available for analysis (difference between means ~15-20 minutes) [4].

Garcia-Diaz et al. compared TRT to sleep estimated time based on actigraphy. They concluded that wrist actigraphy improved agreement with PSG only slightly. They speculated this was due to actigraphy overestimating TST in patients with sleep-related breathing disturbances [5].

Norman et al. used a method that involved movement and respiratory signals to estimate TST, improving agreement with PSG-based AHI [6].

Peripheral arterial tone (PAT) devices have automated algorithm that use movements (actigraphy) for sleep/wake detection [7, 8] (Refs). Hedner et al. showed moderate agreement for epoch-by-epoch sleep/wake classification (Cohen's Kappa = 0.549 (95%CI = 0.544-0.553) against PSG (manually scored; AASM 1999 Criteria) corresponding to a TST ICC of 0.79 and minimal TST bias (690 ± 152 vs. 690 ± 154 epochs for PSG and PAT respectively) [7] (Ref). Massie et al. showed similar TST ICC of 0.78 against PSG (manual corrected automatic scoring; AASM 2012 Recommended criteria) with a mean bias of -17.9 ± 51.4 minutes, however there was no epoch-by-epoch comparison [8] (Ref).

Heading 3: Tools used for arousal scoring

Respiratory events associated with electroencephalography (EEG)-based arousals are considered important when scoring polysomnography (PSG), as they can lead to significant sleep apnoea symptoms, even without oxygen desaturation [9, 10]. As such, the recommended hypopnoea definition in the current American Academy of Sleep Medicine (AASM) PSG standard allows for hypopnoea scoring when airflow reduction is accompanied by EEG-based arousal without, or with minimal, oxygen desaturation [9, 10]. Additionally, EEG-based arousals are essential for scoring respiratory effort-related arousals (RERA's) in PSG for the calculation of the respiratory disturbance index (RDI) [9].

The study of To et al. used an ARES device worn on the forehead that measured blood oxygen saturation, pulse rate, airflow and respiratory effort, snoring levels, head movement and head position [11]. They compared the device using various hypopnoea definitions derived from automated analysis, to simultaneous PSG manually scored using Chicago criteria, in a population of suspected OSA patients. One RP hypopnoea definition required at least 1% desaturation accompanied by changes in pulse rate, head positions or snoring sounds, as a surrogate for arousal. When comparing to PSG apnoea hypopnoea index (AHI), the mean difference in AHI using the ≥1% definition was less compared to a ≥3% and ≥4% definition, and sensitivity for diagnosis at an AHI ≥5 was greater, however specificity was reduced with almost 40% of PSG normal patients classified as mild OSA.

Ayappa et al. also used an ARES device to make comparisons between indices derived from automatically scored RP (with limited manual review) and manually scored PSG, in a group that mostly included suspected OSA participants [12]. For one comparison, the RP AHI incorporated a hypopnoea definition that required ≥1% desaturation plus 1 surrogate arousal indicator (head movement, changes in snoring, or changes in pulse), whereas the comparison custom PSG RDI incorporated (i) hypopnoeas with greater than 50% flow reduction, (ii) hypopnoeas with 30-50% flow reduction accompanied by >=4% desaturation, and, (iii) RERA's which required a discernible flow change accompanied by an EEG arousal. That study reported high correlation between the RP AHI with simultaneous PSG RDI (ICC = 0.93; mean difference = 3.2/h, 95% CI: 1.2 to 5.3) and good diagnostic accuracy, with sensitivity using an RDI cut-off of 15 per hour of 95%, specificity of 94%, positive likelihood ratio (LR+) of 17.04, and negative likelihood ratio (LR-) of 0.06. The authors suggested the device used with the surrogate arousal event definition would be useful to rule out significant sleep disordered breathing.

Lachapelle et al. reported the AHI agreement and diagnostic accuracy was improved with RP hypopnoea scoring that used surrogate arousal defined as increase in pulse oximetry-derived heart rate ≥ 6 beats/min [13]. Bland-Altman analysis showed RP vs. PSG AHI mean difference of 11.2/h (95% CI 33.6, − 11.1) without vs. 7.2/h (29.6, − 15.4) with the surrogate arousal measure. Diagnostic accuracy improvement was reflected by an increased area under the receiver-operating characteristic (ROC) curve for AHI thresholds of 10 and 15 events/h, however AUC for threshold of 5 was not improved. Additionally, the study was conducted using Chicago criteria, in a targeted group of patients with a moderate-high pre-test probability of OSA after an inconclusive Type 3 study, and the comparison was conducted with PSG from a different night.

Masa et al. assessed RP hypopnoea definitions that either included or excluded a surrogate arousal, in patients with suspected OSA [14]. Surrogate arousal was defined as, "a clear resolution of airflow or band reduction by a sudden increase in amplitude and frequency ≥ 2 breaths". The most relevant comparisons was between AHIs derived from manually scored PM, with and without surrogate arousal, versus AHI from simultaneous PSG, manually scored using hypopnoea criteria equivalent to current AASM recommended definitions . With the use of surrogate arousal the study reported slightly smaller differences and slightly lower

agreement limits between PM AHI and PSG AHI. In terms of diagnostic agreement, with the use of surrogate arousal the area under ROC curves was also slightly improved. These same improvements were not seen when comparing home based RP to PSG conducted on a separate night, such that the authors suggested that the magnitude of improvements related to use of a surrogate arousal were not sufficient to overcome other factors that produce AHI differences between PSG and home RP. They concluded that incorporating an alternative arousal measure into PM did not substantially increase its agreement with PSG when compared with the PM without surrogate arousal. In a subset of PSGs (40) this study also reported good agreement between hypopnoea indices (HI) that were derived from hypopnoeas scored based on surrogate arousal alone (5.6/h ±4.8) compared to HI based on hypopnoeas with EEG arousal alone (4.6 ± 4.0), where the mean total HI was approximately 20/h.

Vat et al. used pulse-wave amplitude (PWA) drops as a surrogate for EEG arousal in large population-based sample stratified for OSA severity. PWA is a measure obtained from finger photoplethysmography during pulse oximetry and PWA drops are considered to be a marker of peripheral vasoconstriction associated with arousals from sleep [15]. Channels were removed from home based PSG recordings to simulate RP and compare various scoring methods. The study described a modest correlation between the EEG arousal index and the PWA drop index (30% drop from baseline) (r=0.2), with the PWA drop index overestimating the arousal index by 15.6/h ± 17.5. Using hypopnoea criteria requiring 3% desaturation or PWA drops led to an overestimation of the AHI compared to PSG scoring according to current the AASM 2012 hypopnoea definition (mean ± SD difference +3.5±5.4/h) as opposed to underestimation when PWA drops were not included in the criteria (−1.3/h ±4.8). Best diagnostic accuracy for mild-to-moderate OSA was obtained using hypopnoea criteria requiring at least 3% desaturation without PWA drops, however including PWA drops in the criteria resulted in a slight improvement in diagnostic accuracy for severe OSA.

Although surrogate arousal assessment was not the primary purpose, the study of Norman & Sullivan used body movement as surrogate for arousal in simulated RP, where body movement was defined as an abrupt change in the baseline pattern of respiration on the thoraco-abdominal traces [6]. On average there were 14 more respiratory events scored in PSG compared to simulated RP

(range −13 to 163). The events that were scored using PSG but not RP were almost all hypopnoeic events that were not associated with a ≥3% desaturation or a body movement.

PAT devices are atypical compared to other type 3 devices, in that they record peripheral arterial tone, pulse rate, pulse oximetry, and actigraphy but do not record respiratory flow or movement. The PAT signal detects pulsatile volume changes of the peripheral artery bed at the finger. Respiratory events are detected indirectly by detecting surges in sympathetic activity, signalled by attenuation of the PAT signal, accompanied by heart rate increase and desaturation at the end of a respiratory event [8, 16, 17]. Thus detection of respiratory events for PAT devices is reliant on signals that could be considered surrogates for EEG arousal, however recent WatchPAT validation studies have largely focused on sleep scoring and respiratory event detection rather than arousal detection [7, 17-24].

Table e2 a Scoring criteria of hypopnoeas

Type III studies do not allow for EEG-based arousal scoring. Multiple rule sets have been adapted as follows:

- AASM 1999 [31]: Hypopnoeas require either a >50% decrease in airflow, or lesser airflow reduction in association with oxygen desaturation of >3% [13, 32-34].
- AASM 2007 Recommended/AASM 2012 Acceptable [35, 36]: Hypopnoeas require ≥30% airflow reduction associated with ≥4% desaturation [15, 37-43].
- AASM 2007 Alternative [36]: Hypopnoeas require ≥ 50% airflow reduction with ≥3% desaturation [11, 44-48].

- AASM 2012 Recommended [35]: Hypopnoeas require ≥ 30% airflow reduction and ≥3% desaturation [2-4, 6, 14, 15, 29, 37, 38, 40, 43, 49-53].

Table e2 b Definitions of custom hypopnoea

In addition, various studies used custom hypopnoea definitions:

- ≥50% airflow reduction with ≥4% desaturation [5, 11, 12, 25, 26].
- Discernible airflow reduction with ≥3% desaturation [27] or ≥4% [28].
- ≥50% flow reduction with 1% desaturation and surrogate arousal [11, 12].
- ≥50% flow reduction (automatic scoring) [29].

One study did not describe hypopnoea criteria [30] and there is no distinction between hypopnoeas and other respiratory events in PAT studies [7, 8, 16-24].

Tables: Technical specifications of using Type III devices.

Table e3. Flow sensors: thermistors

Author	Design	Patient population	Technical findings	Comments
Hoppenbrouwers et al. 2019 [54]	An airflow signal compared to SpO_2 analysis in the screening for adults with OSA.	N=39 (23 males), age 47±12 yrs, BMI 27.6±5.6 kg/m2, AHI 10.0±8.8	Airflow characterisation: time domain features derived from the airflow signals were: mean, median, std and iqr of the airflow signal. For the frequency domain features, a frequency band of interest was defined between 0.025 Hz and 0.050 Hz, corresponding to events lasting 20 to 40 seconds (reported as the typical range in duration of apnea events [9]). Four spectral features were extracted from this 0.025-0.050 Hz band from the power spectrogram: mean median, std and iqr.	Studied the performance of airflow analysis using an integrated thermistor from nasal pulse oximetry. The information from an airflow signal provides additional information to identify adults with OSA.
Gutierrez-Tobal et al. 2017 [55]	The spectral analysis of 315 NPP and corresponding TH recordings is firstly proposed to characterise the conventional band of interest for SAHS	N=315 (71% male), age 49.9±12.0 yrs, BMI 25.5±9.5 kg/m^2. AHI<5 : n =39 AHI 5-15 : n=91 AHI15-30 : n=69	The thermistor sensor might be not necessary for SAHS severity estimation if an automatic comprehensive characterisation approach is adopted to simplify the diagnostic process	

	(0.025-0.050 Hz.). A magnitude squared coherence analysis is also conducted to quantify possible differences in the frequency components of airflow from both sensors. Then, a feature selection stage is implemented to assess the relevance and redundancy of the information extracted from the spectrum of NPP and TH airflow.	AHI≥30 : n=116		
Arifuzzman A et al. 2016 [56]	Description of a low-power thermal-based sensor system for low air flow detection (system architecture, physical model and temperature behaviour).	NA	The thermal-based air flow sensor comprises a heater and three pairs of temperature sensors that sense temperature differences due to laminar air flow. Detects airflow as low as 0.0064 m/s. The sensor is connected to the sensing mirror circuit. The ring oscillator is connected to the mirror circuit and is built with a three-stage inverter to make frequency variations with air flow. The output of the ring oscillator is amplified by a	The low-power dissipation, high linearity and small dimensions of the proposed flow sensor and circuit make the system highly suitable for biomedical applications.

			level shifter amplifier, and, finally, the driver circuit sends the output signal to the monitoring device.	
Gehring et al. 2014 [57]	Thermocouple and pressure cannula responses to oral and nasal breathing using a polyester model.	NA	During nasal breathing: thermocouple amplitude = 0.38 Ln [pneumotachograph amplitude] + 1.31 and pressure cannula amplitude = 0.93 [pneumotachograph amplitude](2.15); during oral breathing: thermocouple amplitude = 0.44 Ln [pneumotachograph amplitude] + 1.07 and pressure cannula amplitude = 0.33 [pneumotachograph amplitude] (1.72); (all range ~ 0.1-~ 4.0 L s(-1); r(2) > 0.7). For pneumotachograph amplitudes <1 L s(-1) (linear model) change in thermocouple amplitude/unit change in pneumotachograph amplitude was similar for nasal and oral airflow, whereas nasal pressure cannula amplitude/unit change in pneumotachograph amplitude was almost four times that for oral.	
Issa et al. 2013 [58]		NA	Thermal flow sensors are	Miniaturised thermal

			capable to detect very low air velocities by optimising the noise sources.	flow sensors have opened the doors for a large variety of new applications due to their small size, high sensitivity and low power consumption.

Table e4. Flow sensors: nasal cannula

Author	Design	Patient population	Technical findings	Comments
Gehring et al. 2014 [57]	Thermocouple and pressure cannula responses to oral and nasal breathing using a polyester model.	NA	During nasal breathing: thermocouple amplitude = 0.38 Ln [pneumotachograph amplitude] + 1.31 and pressure cannula amplitude = 0.93 [pneumotachograph amplitude](2.15); during oral breathing: thermocouple amplitude = 0.44 Ln [pneumotachograph amplitude] + 1.07 and pressure cannula amplitude = 0.33 [pneumotachograph amplitude](1.72); (all range ~ 0.1-~ 4.0 L s(-1); r(2) > 0.7). For pneumotachograph amplitudes <1 L s(-1) (linear model) change in thermocouple amplitude/unit change in pneumotachograph amplitude was similar for nasal and oral airflow, whereas nasal pressure cannula amplitude/unit change in pneumotachograph amplitude was almost four times that for oral.	

Table e5. Flow sensors: PVDF sensor

Author	Design	Patient population	Technical findings	Comments
Kryger et al. 2013 [59]	A PVDF airflow sensor in addition to the traditional thermal sensor and pressure sensor.	N=60 (28 males), age 49.9±13.6 yrs, BMI 34.8 ±8.1 kg/m2, AHI 29.8±29.9	The PVDF thermal data were acquired with a HFF of 15 Hz and LFF of 0.1 Hz. The PVDF pressure data were acquired with a HFF of 15 Hz and LFF of 0.05 Hz. The sampling rate for all the pressure and thermal sensor channels was 100 Hz.	

Table e6. Flow sensors: PneaVoX sensor

Author	Design	Patient population	Technical findings	Comments
Sabil et al. 2019 [60]	Comparison study	Age 66.7±15.3 yrs, BMI 30.1±4.6 kg/m2, neck circumference 42.8±4.1 cm.	PneaVoX is a stethoscope-like transducer with an acoustic sensor and a pressure senor inserted inside a 28-mm diameter and 15-mm thick protective housing. Filtering techniques are used to separate the high pitch (200 to 2000 Hz) tracheal flow sound from the low pitch (20 to 200 Hz) snoring sound.16 The intensity of the tracheal sound at high pitch allows the measurement of respiratory flow and the detection of apnoeas seconds, it can be assumed that there is no airflow through the trachea and therefore an apnea can be scored.	Popular sensor in France.
Glos et al. 2018 [61]			Detailed description of PneaVoX sensor : A threefold sensor that measures 1) respiratory flow, 2) the pressure variations	

			induced by the snoring sound and 3) the SSP (suprasternal pressure) variations due to respiratory effort. It combines an acoustic sensor and a pressure sensor. Both sensors are inserted into a protective plastic chamber that measures 24 mm in diameter and 13 mm thick. Pressure variations are measured via movements of the skin. In the absence of effort, the RIP signal as well as the SSP signal can be limited to high-frequency cardiogenic oscillations.	
Mlynczak et al. 2017 [62]	Implementation of sensor in 40 real-world, whole-night recordings.	N=16 (10 males) generally healthy subjects, age 25-75 yrs, students and university staff.	Wireless version of the PneaVoX sensor.	
Amaddeo et al. 2016 [63]			Detailed description of PneaVoX sensor : The surface of the transducer attached to the skin comprises a 2 mm-thick cuff, designed to ensure an airtight cavity between the skin and the transducers. Sounds in	

			the cavity related to respiratory flow and snoring are recorded by the microphone. Static pressure variations in this cavity, related to increasing deformation of the suprasternal notch during obstructed inspirations, are measured by the SSP. According to the intensity and frequency, three different signals are therefore recorded from the PneaVoX sensor: • Respiratory effort is recorded from the SSP with a frequency range between 0.02 and 20 Hz; • Snoring is recorded from the microphone with a frequency range between 20 Hz and 200 Hz and is defined by an acoustic intensity greater than 76 decibels in the transducer chamber; • Respiratory flow (in- and outflows) is recorded from the microphone with a frequency range between 200 and 2000 Hz during inspiration and expiration.	

Table e7. Effort sensors: Piezo-electric bands

Author	Design	Patient population	Technical findings	Comments
Lin et al. 2016 [64]		N=34	Due to the instability of the piezo sensor, the amplitude ratio was considered, instead of the amplitude, as a new feature; based on the nature of the piezo sensor, the frequency ratio was proposed as another new feature; the covariance of ABD and THO is considered as an auxiliary feature if both ABD and THO signals are used in the analysis.	Considerable potential of applying the proposed algorithm to clinical examinations for both screening and homecare purposes.
Vaughn et al. 2012 [65]	Mechanical test model	NA	PE belts perform similarly to RIP belts at distraction distances up to 10.0 centimetres.	Further testing on biological models is needed to determine if piezo-electric belts are a suitable alternative for RIP belts.

Table e8. Effort sensors: Strain gauges

Author	Design	Patient population	Technical findings	Comments
No studies available				

Table e9. Effort sensors: Respiratory induction plethysmography

Author	Design	Patient population	Technical findings	Comments
Vaughn et al. 2012 [65]	Mechanical test model	NA	PE belts perform similarly to RIP belts at distraction distances up to 10.0 centimetres.	Further testing on biological models is needed to determine if piezo-electric belts are a suitable alternative for RIP belts.

Table e10. Effort sensors: PVDF belts

Author	Design	Patient population	Technical findings	Comments
Koo et al. 2011 [66]	PVDF belts compared to RIP belts	N=50 (23 males) undergoing polysomnography, BMI 36.2±8.2 kg/m^2, AHI=26.	PVDF is a specialty fluoropolymer substance which reacts almost instantaneously to changes in temperature, pressure, strain, and impedance, making it a potentially useful substrate to sense respiratory flow or effort. Similar to inductance plethysmography, PVDF can be incorporated into a belt surrounding the chest and abdomen but unlike RIP, PVDF measures impedance and not inductance to estimate breathing and respiratory effort. Use of inductance technologies for respiratory measurement is based on the principle that the changes in current in the coiled wires surrounding the chest or abdomen induced by breathing are linearly proportional to changes in the cross-sectional areas occurring during breathing. In	PVDF belts may be used just as RIP often is, as a « back-up » signal for detecting respiratory events when nasal pressure signals become artefactual or are lost.

			contrast, impedance measures changes in electrical resistance, which usually are not linearly related to changes in cross-sectional dimensions.	

Table e11. Effort sensors: Jaw movement sensor

Author	Design	Patient population	Technical findings	Comments
Cheliout-Heraut et al. 2011 [67]	Portable monitoring device with mandibular movement signal versus polysomnography	N=90 (60 males), age 55.4±8.7 yrs	The principle of the measure of the jaw movements is based on the mutual electromagnetic induction of two electromagnets (the sensors). The probes are placed on the vertical midline of the face, parallel to each other, one on the forehead and one below the lower lip. The output voltage is a monotonic cubic function of the distance between the two probes. The voltage is sampled at 10 Hz, digitally linearized and the corresponding mouth opening is stored on the computer synchronously along with the other parameters recorded by the recorder. Jaw movement data can be expressed in absolute values (millimetre) or in normalised value (percentage of mouth opening), the reference value (zero) being the fully closed	

			mouth level.	
Senny et al. 2012 [68] Maury et al. 2013 [69]			The recording of this mandible movement signal was performed by a distance meter based on the principle of magnetometry. The sensors were composed of two coils and capacitors, each embedded in a small cylinder (7 mm diameter; 25 mm main axis). They were disposed, parallel to each other, perpendicular to the midline of the face, and fixed with plasters, one in the dimple above the chin and the other on the forehead. They were connected to an electronic circuit by two cables. The electronic circuit converted distance into voltage. The signal was digitalised with a sampling frequency of 10 Hz. Physical calibration was done by asking the patient to first close his/her mouth and then to open it fully.	

Table e12. Effort sensors: Peripheral arterial tonometry (PAT)/photoplethysmography (PPG)

Author	Design	Patient population	Technical findings	Comments
Peripheral arterial tonometry				
Penzel et al. 2020 [70]	Polysomnography versus PAT recorder device (WatchPAT)	N=85 (50 had cardiac problems such as heart failure (n=33) or atrial fibrillation (n=9) or both conditions (n=8)). Age 17-90 yrs.	New algorithm to distinguish between central and obstructive sleep apnoea.	Until recently, PAT was not able to distinguish between central and obstructive sleep apnoea. Patients with alpha blockers and short-acting nitrates were excluded. The algorithms for distinguishing central and obstructive sleep apnoea events are protected and not open to the investigators or the public.
Massie et al. 2021 [71]	Polysomnography versus PAT recorder device (NightOwl)	N=261, age 54±14 yrs, AHI 31.9±25.6, BMI 30.±5.9 kg/m^2.	NightOwl, similar to WatchPAT, is built around a fingertip-mounted PPG probe. From the PPG measurement, the arterial blood oxygen saturation, pulse rate and PAT are derived. The probe also contains an accelerometer for the	The sensor and algorithm is able to identify whether or not a sleep epoch comprises REM sleep.

			detection of limb movement to determine sleep/wake based on actigraphy.	
Massie et al. 2018 [8]	Polysomnography versus PAT recorder device (NightOwl)	N=101 (56% male), age 53±13 yrs, BMI 28.8±4.9 kg/m^2, AHI 26.87±20.87.	The NightOwl sensor acquires accelerometry and PPG from which it derives actigraphy, SpO2, PAT and pulse rate, among « other features ».	The NightOwl software derives the Respiratory Event Index (REI) as well as the total sleep time as main clinical parameters.
Hedner et al. 2011 [7]	Validation study Polysomnography versus simultaneous PAT recorder device (WatchPAT100)	N=228 (17 normals, 139 referred subjects, 71 randomly drawn from a population based cohort), age 49±14 yrs, BMI 29±6 kg/m^2, RDI 30±23).	Analysis of autonomic signals from PAT recorder can detect sleep stages with moderate agreement to more standard techniques : Sleep/wake detection is based on assessment of movements and their occurrences (periodic or sporadic) while the sleep stage detections (REM, deep/light sleep) are based on the spectral components of the PAT signal.	
Photoplethysmography				
Ye et al. 2021 [72]	Algorithm development	NA	RR-interval based OSA detection was advanced by considering its real-world practicality from energy perspectives. The energy efficiency of the	A novel model, called FENet, was studied that extracts features from different frequencies of the input RRinterval

			detection model is crucial to fully support an overnight observation on patients. This creates challenges as the PPG sensors are unable to keep collecting continuous signals due to the limited battery capacity on smart wrist-worn devices. A novel Frequency Extraction Network (FENet), was proposed which can extract features from different frequency bands of the input RR interval signals and generate continuous detection results with down sampled, discontinuous RR-interval signals. With the help of the one-to-multiple structure, FENet requires only one-third of the operation time of the PPG sensor, thus sharply cutting down the energy consumption and enabling overnight diagnosis.	signals to perform OSA detection in an energy efficient manner. We constructed a dilated convolutional neural network with a set of filters for different frequency bands.
Lazazerra et al. 2021 [73]	Algorithm development	N=96 overnight recordings of patients suspected to suffer from OSA and without any cardiovascular co-morbidity.	DAP (“Decreases in the amplitude fluctuation of PPG”) detector: a DAP event was identified when the PPG envelope was lower than the predefined adaptive threshold ζ(n), for a minimum time	Due to the noisy nature of the PPG signal, it would not replace clinical devices like those based on the nasal pressure signal.

			duration (∆nDAP), set a priori. The DAP detector was designed to highlight PPG signal shape variations and then those detections were discriminated, by verifying if an oxygen desaturation occurred in those time instances. PPG and SpO2 signal sampled at 500 Hz.	
Li et al. 2021 [74]	Algorithm development	Elderly men with both current and past Post-Traumatic Stress Disorder (PTSD).	Combined PPG and actigraphy-based sleep stage classification approach using transfer learning from a large ECG sleep database. Results demonstrate that the transfer learning approach improves estimates of sleep state.	The use of automated beat detectors and quality metrics means human over-reading is not required, and the approach can be scaled for large cross-sectional or longitudinal studies using wrist-worn devices for sleep staging.
Hayano et al. 2020 [75]	PPG was recorded simultaneously with a wearable watch device.	N=41 patients referred for PSG	The wearable watch device (E4 wristband) emitted green light and recorded PPG as the inverted intensity of reflected light at a sampling frequency of 64 Hz and a resolution of 0.9 nW/digit. The	Algorithm could be used for the quantitative screening of sleep apnoea.

			PPG data were uploaded offline to the manufacturer's cloud via the Internet (E4 connect, Empatica, Milan, Italy), where the PI time series were measured as the foot-to-foot intervals of the pulse waves with motion artefacts removed Pulse interval data were analysed by an automated algorithm called auto-correlated wave detection with adaptive threshold (ACAT).	
Abdul Motin et al. 2020 [76]	Algorithm development	NA	A PPG-based sleep–wake classification model divided into four sections: (a) pre-processing (with removal of the unwanted overshoot from PPG), (b) feature extraction, (c) feature selection, and (d) sleep–wake classification (using the mRMR algorithm selected time features). The sampling frequency of PPG was 128 Hz.	An automated approach for sleep-wake classification using a wearable fingertip photoplethysmographic signal. It allows to perform online and real-time classification, since it uses only computationally efficient features.
Liao et al. 2020 [77]	Design of a new device	Presentation of a wearable device built on an Adafruit Circuit Playground Express	It achieves substantially improved performance compared to the commercially	The component cost also remained low (under USD $5 for

		(CPE) board and integrated with a photoplethysmographic (PPG) optical sensor for heart rate monitoring and multiple embedded sensors for medical applications—in particular, sleep physiological signal monitoring.	available Philips ActiWatch2 wearable device. It has an open architecture. The device is easily scalable and has low commercialisation costs. The device is based on the IoMT infrastructure. We used a DFRobot heart rate sensor (SON1303, DFRobot, Shanghai, China) from the DFRobot Gravity Series, which uses a green LED with a 570-nm wavelength. The sensor includes built-in noise filters and issues an alarm when the HR is abnormal. According to the manufacturer's specifications, the HR sensor has an accuracy of 98.5%. The working voltage of the HR sensor is 2.1 V, but it can withstand a maximum of 5.5 V and temperatures of −40~85 ◦C. The chip used in the sensor is able to achieve a bandwidth of 1 MHz at a low current consumption of 60 μA and has low input bias currents of 10 pA. Thus, our device consumes less power (operating current < 10 mA) than other devices that	each component).

			incorporate multiple sensors using a development board.	
Betta et al. 2020 [78]	Algorithm development	N=16 patients randomly sampled from the HypnoLausSleep cohort database, 3 males, age 50.9±6.33 yrs,	A novel automated approach to detect and characterise significant drops in the PWA signal. First, the PWA-time-series is extracted from the raw PPG-signal and potential artefactual segments are identified and excluded from subsequent evaluations. Then, candidate PWA-drops corresponding to local peaks in the variance of the PWA time-series are identified. Finally, significant drops are selected among all candidates, based on a-priori defined criteria, and their main characteristics (e.g., timing, amplitude, slopes, duration, etc.) are stored for further evaluation.	
Motin et al. 2019 [79]	Algorithm development	NA	An automated approach for classifying sleep-wake stages using finger-tip photoplethysmographic signal.	
Garcia-Lopez et al.			The results show that the HR estimated from signals	Respiratory frequency is more predominant in

2018 [80]			obtained with the neck sensor are strongly correlated to the output of the reference finger (R=0.862, MAE=1.27 BPM), whereas SpO_2 measurements are not that accurately predicted (R=0.129, MAE=11.7%).	neck PPG than in finger, which has a great potential for respiratory rate (RR) extraction. These are very promising results for the suitability of the neck as an alternative location for monitoring of respiratory diseases, and specifically for sleep apnoea.
Beattie et al. 2017 [81]	Algorithm development	N=60 (36 males), age 34±10 yrs, BMI 28± 6 kg/m^2.	A peak detector algorithm has been developed to find the peaks in the PPG signal. The time between PPG peaks (PP-interval) is taken as a surrogate for the RR intervals obtained from an ECG. In general PPG signals are more prone to motion artefact than ECG, and in the case of excess motion, the peak detection algorithm does not return any estimated peaks.	
Jayawardhana et al. 2017 [82]	Algorithm development	N=52, AHI 1-82, N=46 AHI≥10, N=6 AHI<10.	The PPG signal provides an optically obtained time-varying measurement of the blood volume in the tissue at the measurement location. Due to the pumping of blood into peripherals by the heart,	

			the PPG waveform varies with the heart cycle. The PPG waveform is also influenced by the breathing cycle. During inspiration, the intra-pleural pressure decreases uncompressing the heart in the process, which results in a decreased stroke volume and therefore decreased blood volume into the peripherals. During expiration, the pressure in the thorax is increased compressing the heart in the process thereby increasing the stroke volume. Hence, the baseline as well as the amplitude of the PPG signal fluctuates in the low frequency region that corresponds to the breathing rate.	
Papini et al. 2017 [83]	Algorithm development	NA	The PPG signals are susceptible to be corrupted by noise and artefacts, caused, e.g., by limb or sensor movement. These artefacts affect the morphology of PPG waves and prevent the accurate detection and localisation of beats and subsequent cardiovascular feature	Algorithm is designed for wristworn reflective PPG sensors for sleep research, in which the cardiac signal can be more easily corrupted by, e.g., motion artefacts and pressure applied to the sensor.

			extraction. The quality index is used to discard corrupted pulse beats. A new algorithm for offline beat detection with single pulse QI assessment is presented. The QI is obtained by comparing each pulse with a dynamic template. The template is derived from the pulses contained in the PPG signal via DTW barycentre averaging (DBA) [9]. Each pulse is warped on the template, using DTW, in order to reduce QI underscoring due to physiological pulse deformations. Then the QI is calculated from the mismatch between the template and the warped pulses. The algorithm is tested on two public datasets.	
Khandoker et al. 2013 [84]			This paper offers a description of the PPG signal : Eight features can be extracted from the PPG signal (Figure 2). - Peak amp: Amplitude of peak point of each	

			pulse. - Valley amp: Amplitude of trough point of each pulse. - PWA: pulse wave amplitude (vertical distance. between Peak amp and Valley amp) during systole. - Pp Interval: Pulse to pulse time interval. - Area: Triangular area between one Peak amp and two neighbouring Valley amp points. - Upslope: gradient towards Peak. - Downslope: gradient towards Valley. The PPG signal oscillates with the heart cycle period, due to the systolic increase in the tissue blood volume, resulting in a lower transmission of light. Peak amp is inversely related to the tissue blood volume, PWA is directly related to the tissue blood volume increase during systole and pp Interval is actually the heart cycle period.	

Table e13. Effort sensors: Pulse transit time

Author	Design	Patient population	Technical findings	Comments
Arslan Tuncer et al. 2019 [85]		N=100 (50 patients and 50 healthy individuals; 50% males), age 63-77 yrs, AHI « moderate ».	In this study a decision support system was developed to determine OSA. The suggested method can perform feature extraction from PTT signals by means of deep-learning method.	Although deep-learning methods used in the study have been very successful, the complex internal mechanisms are a disadvantage of the system. The accuracy of the system can be improved with different parameters and machine learning algorithms.
Chouchou et al. 2011 [86]	PTT continuously monitored during polygraphy.	N=780 (43% male), age 65.8±1.1, BMI 25.4±4.0 kg/m^2.	PTT was calculated as the time interval between the electrocardiographic R wave and a point on the pulse waveform (detected by a plethysmographic finger probe) that was 50% of the height of ascent of the pulse wave. The electrocardiogram and pulse were sampled at	This paper reports sampling features and accuracy details.

			500 Hz. PTT is typically about 250 milliseconds and is measured to an accuracy of 2 milliseconds. PTT values available with every heart beat were oversampled at 5 Hz. The PTT was continuously monitored and an autonomic activation index (AAI) was obtained from the PTT signal and was broken down into total, respiratory, and non-respiratory autonomic activations. The scoring of autonomic activations was obtained using the manufacturer's analysis software.	

Table e14. Effort sensors: Accelerometers

Author	Design	Patient population	Technical findings	Comments
Schipper et al. 2021 [87]	Chest accelerometry versus respiratory inductance plethysmography.	N=20 healthy volunteers, age 46-65 yrs, BMI 20-30 kg/m^2.	Chest-worn accelerometery, which may be quite suitable for non-intrusive measurement. It only requires a single point of mechanical contact with the chest. There is no need for electrical contact and, therefore, it may be worn in clothing, over clothing, or integrated in an adhesive patch. It can be very compact and, due to low power consumption, allow for a long operating time, even when using a small battery. The accelerometer orientation changes during the respiratory cycle are clearly measurable. The 3D acceleration signal is acquired with a sampling rate of 250 Hz.	Chest-worn accelerometry can be a robust and accurate method for measurement of respiratory features under realistic conditions.
Lee et al. 2018 [88]	Evaluation of a device that obtains a continuous tidal	Pilot study on one OSA volunteer	A motion chip (MPU-9250, InvenSense, USA) including a three-axis gyroscope, three-	

	volume signal from real-time lung ventilation images produced by an electrical impedance tomography (EIT) technique.		axis accelerometer and three-axis magnetometer is used to measure respiratory effort and body position signals. The acquired data are transmitted to the main body with a maximum sampling frequency of 250 Hz. A differential buffer amplifier module connected to each electrode. Sixteen of them are soldered on the fPCB near the eyelet connectors. Two operational amplifiers (OPA2140, Texas Instrument, USA) and one difference amplifier (AD8139, Analog Device, USA) are used to implement each differential buffer amplifier with a high common-mode rejection ratio (CMRR) and low output impedance. Its high-pass cut-off frequency is 100 Hz to reduce low-frequency noise and motion artefacts at the electrode–skin interface.	

Table e15. Effort sensors: PneaVoX

Author	Design	Patient population	Technical findings	Comments
See table e6				

Table e16. Snoring sensors: Nasal cannula

Author	Design	Patient population	Technical findings	Comments
Pérez-Warnisher et al. 2017 [89]	Comparison between nasal cannula and a microphone as the point of reference.	N=75 patients (65% males), age 55±13.8 yrs, BMI 30.2 kg/m^2 (27-33), AHI 25.5 (11.7-41.8).	The nasal cannula detects sound indirectly through changes in pressure. A disposable cannula was used with a tube made of polyvinyl chloride that measures 90 cm and has nasal prongs made of silicone (Adult Nasal Cannula 5012, Salter Labs, Lake Forest, Illinois, U.S.A.).	The nasal cannula showed poor reliability and accuracy for measuring snoring. The continued use of a nasal cannula was called into question. The cannula has several problems that remain to be solved. In the first place, the 200 Hz16 sampling rate of the cannula is

				a limitation on the detection of snore events. Therefore, the cannula only detects frequencies up to 100 Hz as compared to the 4 KHz that the microphone can capture. Additionally, the AASM recommends filtering the cannula signal in the range of 10 to 100 Hz, which removes part of the snore events. Furthermore, the material and length of the cannula also may affect snore detection, and the nares may clog with mucus, lowering sensitivity. Moreover, there is a lack of standardisation regarding the type of cannula and pressure transducer that should be used. Ultimately, the algorithm in which the software transforms the pressure measured by

				the cannula into sound is inaccessible. The main limitation of the microphone is that the patient may cover it, thus diminishing the sound waves sensed and decreasing its sensitivity. In addition, microphones may register certain ambient sounds, such as the snoring produced by a bed partner, as the patient's own snoring. To reduce the influence of these limitations, subjects have to be specifically instructed not to cover the box and to sleep alone.
Arnardottir et al. 2016 [90]	Snoring was assessed by listening to the air medium microphone located on a patient's chest, compared to listening to two overhead air medium microphones (stereo) and manual scoring of	N=10, age 53.1±16 yrs, BMI 28.7±2.3 kg/m^2, AHI 10.3±15.8.	The chest audio was capable of detecting snore events with lower volume and higher fundamental frequency than the other sensors. The 200 Hz sampling rate of the cannula and piezoelectric sensor was one of their limitations for detecting snore	The lack of consistency between snoring sensors will affect future research on the clinical significance of snoring. Standardisation of objective snore

	a piezoelectric sensor and nasal cannula vibrations.		events. The different snore sensors do not measure snore events in the same manner.	measurements is therefore needed. Based on this paper, snore measurements should be audio-based and the use of the cannula as a snore sensor be discontinued, but the piezoelectric sensor could possibly be modified for improvement.

Table e17. Snoring sensors: Microphone

Author	Design	Patient population	Technical findings	Comments
Freycenon et al. 2021 [91]	Two filtering techniques were tested: noise reduction adaptive filter and adaptive prediction filter.	NA	PneaVoX sensor developed and patented by Cidelec (Saint Gemmes sur Loire, France). This sensor simultaneously records three physiological parameters such as nose and mouth breathing, respiratory efforts and snoring. The acoustic pressure is picked up thanks to a MEMS microphone (SPU140LR5H Knowles Acoustics-bandwidth [10, 10000] Hz) positioned inside the acoustic chamber of the probe itself. The acquired signal is then analogically filtered in the frequency band [200, 2000] Hz and subsequently digitised by a 16-bit analogue-to-digital converter (AD1845) at the sampling frequency of 4 kHz. Sensor positioned in the suprasusternal notch of the neck (see Fig.6) as this	

			region offers an optimal sensitivity to respiratory sounds generated by breathing.	
Pérez-Warnisher et al. 2017 [89]	Comparison between nasal cannula and a microphone as the point of reference.	N=75 patients (65% males), age 55±13.8 yrs, BMI 30.2 kg/m^2 (27-33), AHI 25.5 (11.7-41.8).	The microphone of the T3 device (Nox Medical) has a sampling frequency of 8 kHz. It detects frequencies of 50 to 3500 Hz and gathers audio data measured in decibels.	
Castillo et al. 2017 [92]	Comparison between an intraoral microphone and a tracheal microphone.	NA	A microphone integrated into the subject's personal oral appliance. Sampling frequency of 44.1 kHz and 16-bit resolution, and captured by Adobe Audition CC (release 2016). Gain to the signal was provided by Amalfi Acoustics preamplifiers prior to the capture by Audition. The intraoral position of the tooth microphone induces the capture of higher frequencies which cannot be recorded with an external contact microphone. In fact, the temporal and spectral patterns of the signals from the oral appliance microphones are like the ones of the tracheal microphone	It is a feasible solution to record OSA-related sounds, providing signals of a good signal-to-noise ratio, comparable to that of commercial tracheal microphones. Moreover, it captures high-frequency components that are not found with external sensors and which could contain acoustic information relevant for the study of the disease.

			after applying a high-pass filter	
Kim et al. 2016 [93]			Embedded microphones recorded snoring sound (8 kHz sampling). Automatic analysis of snoring was performed using Noxturnal software. Briefly, using the adaptive threshold method, snoring episodes were detected when they met a relative threshold (four times higher than the background noise of the signal) and duration (up to 3 s). Other techniques that increased the specificity of detection of snoring included determination of synchronisation with inspiration below a maximal frequency level (500 Hz) and exclusion of any noise resulting from movement.	
Saha et al. 2015 [94]		Men 20-70 yrs, BMI <30 kg/m2.	Tracheal sounds recorded by a Sony EMC-44B omni-directional microphone, placed over suprasternal notch of the neck. Snore sounds were amplified and filtered by a low-pass filter (cut off frequency: 5 kHz)	

			using Biopac DA100C, and digitised at the sampling rate of 12.5 kHz using MP150 Biopac System.	
Azarbarzin et al. 2011 [95]		N=30 (23 males), age 50.6±9.96 yrs, N=23 OSA (16 males), age 49.9±10.2 yrs, BMI 34.1±7.2 kg/m^2, AHI 26.1±22.9. N=7 simple snorers (all males), age 53.1±9.3 yrs, BMI 30.0±3.8 kg/m^2 in simple snorers, AHI 2.3±1.5.	ECM-77B microphone with the high-performance frequency response of 40 Hz–20 kHz, embedded in a chamber (diameter of 6 mm) amplified with a gain of 200 and bandpass filtered with the cut-off frequencies of [0.5–5000 Hz] using Biopac (DA100 C) amplifiers. The amplified signals were digitised at a sampling rate of 10240 Hz using NI9217 data-acquisition module and a custom written LabView program.	

Table e18. Snoring sensors: Piezoelectric vibration sensor

Author	Design	Patient population	Technical findings	Comments
Erdenebayar et al. 2017 [96]		N=45 OSA	A piezo-electric sensor (REF 1420610; Embla Systems Inc.) was attached to the neck of a patient to measure vital signs during nocturnal PSG. Signals were recorded at a sampling rate of 200 Hz. A support vector machine (SVM) was used as a classifier to detect OSA events.	Cannot only provide data on snoring, but also on movement and heartbeat during sleep. Is useful for monitoring sleep and diagnosing OSA.
Lee et al. 2013 [97]	Algorithm development	N=21 OSA	The piezo snoring sensor used was piezoelectric in type, but was made to acquire the vibration related to snoring with frequencies in the range of 5 to 50 Hz. An automated method for snoring detection based on Hidden Markov Models was tested.	Using a vibration signal acquired from a piezo snoring sensor for automatic snoring detection is feasible.

Table e19. Snoring sensors: PneaVoX

Author	Design	Patient population	Technical findings	Comments
See table e6				

Table e20. Body position sensor

Author	Design	Patient population	Technical findings	Comments
Alinia et al. 2020 [98]	Algorithm development	NA	A comprehensive approach using a single accelerometer along with machine learning algorithms for in-bed lying posture classification.	Accelerometers are ubiquitous and inexpensive sensors.
Doheny et al. 2020 [99]		N=11 (9 males), age 47.82±14.14 yrs, BMI 30.9±5.27 kg/m^2, AHI 5.77±4.18.	Low profile wearable inertial sensors (BiostampRC, MC10 Inc.) were programmed to record tri-axial acceleration data at 125 Hz (±4g), and were attached to the chest and upper abdomen. Change in mean acceleration	A low-cost solution for in-home, long term sleeping posture and respiration monitoring. A limitation of the study is the lack of prone position data in the collected dataset

			and orientation are expected as patients move between sleeping postures due to the changing influence of gravitational acceleration on each sensor axis. Accelerometer data was segmented into no overlapping 30 s epochs, and respiration rate and sleeping position was estimated for each epoch.	collected.
Pillar et al. 2020 [100]		N=84	The system used in this study is a home sleep testing system based on a wrist-worn device and a finger probe which acquires Peripheral Arterial Tone (PAT) signals and arterial oxygen saturation levels, together with actigraphy data from a 3D accelerometer that is embedded in the wrist unit, and an optional snoring and body position (SBP) sensor that is positioned under the sternal notch. The SBP sensor encapsulates together a microphone and a 3D accelerometer used to derive the spatial body orientation	A potential disadvantage of the SBP in comparison with the other methods is its location on the very upper chest where respiratory movement is subtle and therefore the signal-to-noise ratio is lower. Provides a more robust and comfortable attachment than respiratory belts which tend to slide and become displaced and may therefore be advantageous for

			relatively to the vertical gravity from which district position can be calculated (supine, left side, right side, prone, or sitting).	home sleep testing.
Lee et al. 2018 [88]		NA	A motion chip (MPU-9250, InvenSense, USA) including a three-axis gyroscope, three-axis accelerometer and three-axis magnetometer is used to measure respiratory effort and body position signals. The acquired data are transmitted to the main body with a maximum sampling frequency of 250 Hz.	
Yoon et al. 2015 [101]	Algorithm development	N=13 subjects (4 males), age 46.4±9.9 yrs, BMI 25.7±3.4 kg/m^2.	A sleep posture estimation algorithm using 3-axis accelerometer signals measured from a patch-type sensor. A 0.1Hz low-pass filter was applied to eliminate undesired sources which were reflected on the recording of accelerometer signal such as respiration and snoring.	Suggested algorithm based on 3-axis accelerometer signal sis capable of estimating sleep posture.
Selvaraj et al. 2014 [102]		N=53 volunteers of healthy and untreated SAS patients (29 males and 24 females) with age range of 22−73 years. The inclusion	The HealthPatchTM sensor is a disposable adhesive patch biosensor worn on the chest that incorporates two surface electrodes with hydrogel on	

		criterion to participate in the study was the age limit of ≥18 years. The exclusion criteria included surgical treatment for SAS and major behavioural and neurological disorders. The AHI had a range of 0.1−85.8 among these subjects.	the bottom, a battery, an electronic module with the embedded processor, microelectromechanical system (MEMS) tri-axial accelerometer and Bluetooth Low Energy (BLE) transceiver. The patch sensor facilitates continuous monitoring of ECG and actigraphy signals at a sampling rate of 125 Hz and 62.5	
Lee et al. 2013 [103]		N=13 healthy subjects (11 males), aged 28.08 ± 3.20 yrs	Approach was based on the fact that the QRS complex changes as body positions change, because the body position influenced the heart's position. Unconstrained ECG data measured from 12 CC electrodes on a bed were used for classification of four basic lying postures. The features were extracted based on the fact that the morphology of ECG varies according to the body posture.	The advantages of the method are: 1) the system can be applied for multiple subjects without individual calibration; 2) users do not care about the battery because the system used power adaptor; 3) users do not need to wear any devices on their body; and 4) the system is made up one modality and fewer sensors.
Skarpsno et al. 2017 [104]		N=664, age 44.4±10.1 yrs, BMI 27.4±4.9 kg/m^2.	Body positions (front, back, and side) were recorded from the accelerometer on the	

			upper back and classified based on “mode” filtering, requiring that a position must last for at least 1 minute to be registered. The inclinometer features make it possible to identify the orientation of the device relative to the line of gravity, and to be quantified as a position shift (eg, a shift from front to back position) an angle change of at least 30° was required.	

Table e21. Oximetry

Author	**Design**	**Patient population**	**Technical findings**	**Comments**
Gumb et al. 2018 [105]	Data from a widely used stand-alone level IV device, the Nonin WristOx$_2$ (Model 3150, Plymouth, MN, USA) and a level III device, the Apnea Risk Evaluation System (Ares^{TM}), Watermark Medical, Boca Ration,	N=178 (152 males), age 52.5±8.9 yrs, BMI 30.3±5.6 kg/m^2, ESS 8.5±5.	Good agreement between indices of OSA that required ≥4% oxygen desaturation. Significant and systematic device-dependent differences in the %time<90%SaO$_2$. The ARES device had significantly greater data loss than the WristOx$_2$.	Differences in cumulative oxygen desaturation measurements between the devices (%time<90%SaO$_2$, baseline SaO$_2$) suggests that caution is needed when interpreting this metric

	FL, USA) were compared.		AHI4 ARES : 12.8±14.1 AHI WristOx$_2$: 12.1±13.3 SaO2<90% ARES : 2.6±5.0 % SaO2<90% WristOx$_2$: 9.6±17.1 % Baseline SaO$_2$: ARES : 96.2±1.6 % WristOx$_2$: 92.2±2.1 % %Artefact ARES : 14.6±15.9 WristOx$_2$: 2.1±7.9%	particularly in populations likely to have significant hypoxia.
Ng et al. 2017 [106]	Comparison between oximetry recording using ResMed ApneaLink Plus (software version 9.20 ResMed, Sydney, Australia) and Compumedics Profusion PSG3 system (software system version 3.4, build 401 Compumedics Limited, Abbotsford,	N=106 (62 males), age 47±15.5 yrs, BMI 31.9±7.1 kg/m2, ESS 7.8±4.8, AHI 23.2±25.0	ApneaLink Plus reports higher ODI values, both for ODI3% and ODI4%, but with wide limits of agreement : ODI4% : ApneaLink : 15.0±17.8 Compumedics : 10.6±16.3 ODI3% ApneaLink : 22.0±19.4 Compumedics : 14.9±18.8	Clinically significant difference in ODI values generated by the two systems, likely due to device signal processing, rather than difference in ODI calculation algorithms. The differences are large enough to significantly affect diagnostic thresholds for OSAS. Caution is advised when

	Victoria, Australia)			comparing ODI between patients or when performing posttreatment reassessment in the same patient, unless the same oximeter and software algorithm have been used.
Cross et al. 2015 [107]	Study at high altitude: data from wrist-worn oximetry. Simple moving averages were retrospectively applied to raw, pre-processed beat-to-beat SpO_2 time series.	N=5 healthy adult males, age 35±5 yrs,	The artefact index decreased with progressively wider moving average windows (reported in a figure). SpO_2 low was higher for all window lengths than those values derived from the « artefact-free » beat-to-beat SpO_2 data.	Increasing oximeter averaging window length progressively underestimates the frequency and magnitude of respiratory events at high altitude based on ODI.
Vagedes et al. 2014 [108]	Development of a formula that allows conversion between desaturation rates obtained using different averaging times for various desaturation levels and minimal desaturations.	N=15 infants, age 32-33 weeks.	A linear relationship between the logarithm of the desaturation rate per hour and the logarithm of the averaging time. Log to base 10.	

Table e22. ECG

Author	**Design**	**Patient population**	**Technical findings**	**Comments**
No studies available				

Table e23: Properties of different sensors to assess ventilation during cardiorespiratory sleep studies [109, 110]

Type	Physical characteristics	Advantages	Disadvantages
Airflow sensors			
Pneumotachography [111-113]	The gold standard for measuring flow. It quantifies airflow by measuring the pressure drop across a linear (constant) resistance.	- very precise - quantitative (the only sensor able to detect changes in the percentage of ventilation) - also assesses the morphology of the flow signal and flow limitation - linear - easy to calibrate	- requires a snug-fitting face mask with the pneumotachometer - not used for routine diagnosis - thermal drift - mask leakage - condensation problems - not able to detect flow limitation - cumbersome and uncomfortable for subject to wear - the procedure itself may potentially increase respiratory effort to some extent
Thermistor [9, 113-116]	Has electrical characteristics (resistance and voltage) that	- monitors both nasal and oral	- signal is not well correlated

	depend on its temperature: - Thermocouple (change in voltage output with temperature) - Thermistor (change in resistance with temperature) Records temperature variations caused by breathing (inhaled air cooler, exhaled air warmer). Changes are dependent on environmental temperature. Changes dependent of mass / inertia of temperature probe. Non-linear flow/temperature changes – extremely difficult to get linear	airflow - cheap (in sales and during use) - less signal when breathing through the mouth - the response time is sufficiently fast enough for detecting and displaying wave-forms consistent with apnoeas. - do not require sensitive pressure transducers	with breath amplitude (nonlinearity between flow/temperature changes) - hence, semi-quantitative, therefore useless in practice - usually resulting in overestimation of ventilation (underestimation of flow reduction) as flow decreases - not able to detect flow limitation. - too slow for displaying hypopnoeas - main limitation is the slowness of its dynamic response - probe often shifts and misses signal - harder to use with a CPAP mask - uncomfortable for subject to wear
Nasal cannula [117-123]	Records air pressure changes Non-linear flow/pressure changes (underestimates flow at low flow rates and overestimates flow at higher flow rates) – Can be made linear mathematically by a square root transformation	- quantitative - fast response - able to detect flow limitation - signal can be made linear mathematically	- may not provide an absolutely accurate estimate of total flow over the entire night: - no signal when breathing through the mouth - changes in cannula position - obstruction of the cannula by nasal secretions

	(flow is proportional to the square root of the nasal pressure signal), which is a more accurate estimate of nasal flow. More sensitive than thermal sensors for detecting hypopnoeas. Very sensitive for hypoventilation – tendency to overestimate hypoventilation. Nasal cannula has a better negative predictive value and a poorer positive predictive value than RIP. Must have appropriate low and high filter settings to properly display the useful information provided by the recording.		- not linear (but can be made linear mathematically) - overestimates flow reduction, due to the quadratic pressure-flow relationship, compared to the untransformed signal, but the difference is usually not clinically significant. - deflection in the nasal pressure signal may occur during mouth breathing: the relationship of nasal pressure and flow is ten altered with the signal no longer proportional to the flow squared. - more expensive due to disposables - harder to use with a CPAP mask - uncomfortable for subject to wear - no adequate signal if mouth breathing: if the patient is mouth breathing during a hypopnoea, the nasal cannula tracing may falsely suggest that an apnoea is present. - need for an accurate pressure transducer - nasal prongs can markedly increase nasal resistance in subjects presenting with nare narrowness and/or deviated nasal septum. - some patients have difficulty tolerating the cannula for the whole night

			- flow limitation is common and 30% (based on a shape criterion) could be used as the upper limit for normal breathing during sleep
Polyvinylidenefluoride (PVDF) [59, 112, 124]	Change in PVDF signal with temperature (proportional to the difference in temperature between the two sides of the film) or pressure. PVDF is a polarized fluoropolymer whose electrons are aligned (similar to a magnet), and any force that disturbs this alignment causes the film to generate a measurable voltage.	Responds linearly and nearly instantaneously to changes in temperature. More sensitive to detect changes in airflow during sleep compared with traditional thermal devices, due to a faster response time (around 0.005 s) than traditional thermal devices (around 1 s) and a comparable response time to pressure-based airflow devices. 150.000 times higher sensitivity than thermistors. Ability to detect events over the course of the night does not vary.	Not well spread. Disposable sensors: behaviour of the sensors could vary from batch to batch, but this possibility could also be present in multiple-use devices. The PVDF sensor does not show flow flattening during airflow limitation.
Tracheal sound sensors [125]	Tracheal sounds correlate well with respiratory flow and could be used in the same way as any other flow signal for the analysis of respiratory events. Tracheal sounds are a measure of the body surface vibrations set into motion by pressure fluctuations. The sensor consists of an acoustic	If a third band pass filter is used at a much lower frequency range, the SSP variations could be extracted, which could be used for the detection of respiratory effort and help characterise the respiratory events Tracheal sound intensity is robust enough, even at low airflow rates, so an acceptable signal-to-noise ratio can be achieved without pre-	Incorrect positioning of the sensor could result in poor quality or absence of the signal. Detecting hypopnoeas is more difficult.

	sensor inserted into a thick protective plastic chamber, with deep cuff creating an airtight space between the transducer and the skin of the patient, usually 2-3 mm between the sensor and the contact surface of the chamber. Amplification may not always be necessary given the high intensity of the tracheal sound signals. The signal is band pass filtered to separate the high-pitched frequencies of the breathing sound from the low-pitched frequencies of the snoring sound. Sensor should be place on the skin above the sternal notch and then secured in place using adhesive tape.	amplification. The response characteristics are linear over a wide range of frequencies.	
End-tidal capnography [126, 127]	CO2 levels low during inspiration, high during expiration. Gas is suctioned via a nasal cannula with tips located in the nostrils to an external sensor at bedside (side stream method). Can indicate an apnoea (absence	- a delay in the start of an apnoea by the capnograph is caused by the time required for exhaled PCO_2 to reach the sensor at bedside. - provides breath to breath information, can replace the respiratory flow signal, negating	- only a qualitative indicator of flow - not quantitative - only useful for detection of apnoeas - rather expensive - not useful to detect flow limitation - in the side stream method, there is a

	of exhaled PCO2).	the need for an additional sensor.	delay in exhaled gas reaching the sensor (the CO2 tracing is delayed compared with the exhaled airflow). - the end-tidal PCO_2 is generally around 5 mmHg lower then the arterial PCO_2 (anatomic and physiologic dead space dilutes the alveolar PCO_2). - when lung disease is present (especially in patients with increased dead space and small tidal volumes), they have more variability in CO2 levels (higher differences between end-tidal PCO_2 and arterial PCO_2). - rarely used in an ambulatory setting - mouth breathing not assessed - nasal cannula can occlude with secretions
Effort sensors			
Piezo [65, 128, 129]	Made from piezo-crystals (quartz crystals and some manufactured ceramics), Have the characteristic of inducing an electric charge when they are stretched. When connected to an amplifier, they can reflect respiratory movement (no battery required). The sensors can be attached	- very robust - cheap - comfortable for subject to wear - very easy to use in sleep studies	- provide only qualitative information on changes in ventilation or airflow. - not quantitative - Cannot be used to reliably distinguish between central and obstructive respiratory events. - poorly validated - currently positioned as an obsolete

	to belts that are placed around chest and abdomen.		technique - use a single sensor between belt material surrounding the thorax and abdomen: the signal depends on variations in the tension on the sensor which may or may not reflect the magnitude of thoracoabdominal excursions. - may provide misleading information (false absence of respiratory effort) if not properly positioned and tensioned (sized). - belts have to be tightly secured around chest and abdomen, which can impair the respiratory effort itself in susceptible individuals - inversion of the signal can occur due to overstretching of the piezoelectric crystals. - due to instability of the piezo sensor, these channels are not recommended as the first-line sensors by the AASM. - the instability nature is complicated by its non-stationarity nature, and suitable analysis tools are unavailable.
Strain gauge [65, 128, 129]	A strain gauge (or load cell) is a device used to measure strain on an object. Small voltages are generated in	- cheap - comfortable for subject to wear	- the sensory element is located on only a very small section of the band's length (near the nipple line or mid-chest and just

	response to movement. The sensors are attached to belts that are placed around chest and abdomen.		above the belly button) - poorly validated - currently positioned as an obsolete technique - only a qualitative indicator of flow, not quantitative - problems may occur when a patient lies on top of the piezo crystal or in morbidly obese patients. Signals may not accurately reflect changes in ribcage and abdominal volume during respiration
Respiratory induction plethysmography (RIP) [1, 130-135]	Based on detection of changes in volume of the chest and abdomen during inspiration and expiration. The RIP belts are embedded with copper wires, woven in a sinusoidal pattern that encircle the body. A very low electrical current is applied from a high-frequency electrical oscillatory circuit to the wires, generating an oscillating signal, in response to variations in resistance associated with expansion and contraction of	- quantitative (theoretically: when the RIP is calibrated, it permits the measurement of the volume of the breathing cycle) - quite linear - no problem when breathing through the mouth - also substitutes belts for measurement of respiratory effort - comfortable for subject to wear - if one takes the time derivative of the RIPsum signal, the result is an estimate of airflow	- calibration is difficult (especially in obese patients) and usually impaired during sleep, and therefore RIP does not measure ventilation completely quantitatively - sum of thorax and abdomen has to be chosen such that this is zero in case of an obstructive apnoea - the signal is affected by the position of the subject and the belts can move on the patient over the night - costs of belts can rise - the estimate of airflow has less accuracy for detecting flow limitation than the pneumotachograph and nasal cannula.

	the belts due to changes in body circumference (it behaves like an induction spindle or coil). Changes in inductance are converted into a voltage output. The RIP belts consist of wires attached to a cloth band in a zig-zag pattern (which produces a larger change in inductance for a given change in band circumference). If properly calibrated, the sum of the chest and abdominal signal can provide a measure of tidal volume. In the absence of a sum signal, analysis of the RIP chest and abdominal channels can indicate a hypopnoea, provided that there is a good baseline from which to make this judgment.		
Effort belts with polyvinylidenefluoride (PVDF) sensors [66]	PVDF can be incorporated into a belt reflecting respiratory effort (detects the movements of rising and falling of thorax and abdomen corresponding to inhalation and expiration	- Responsive to both airflow temperature and pressure - The amplitude and frequency of the signal is directly proportional (linear) to the mechanical deformation of the PVDF film.	- Both PVDF belts and RIP modestly underestimate breathing events when compared to the reference standard pneumotachography. - Results comparable with RIP - Not fully incorporated in clinical practice

	phases). Excellent sensor: very sensitive towards changes in the strain applied on it: Responds to mechanical changes (piezoelectric properties) and to thermal changes (pyroelectric properties).	- As the film is heated and cooled, the resultant expansion and contraction induces secondary piezoelectric signals. - No additional external power supply is required to generate or amplify the signal. -	
Midsagital jaw movement sensors [67, 136]	Mandibular movement sensors: use of a distance meter based on the principle of magnetometry (mutual electromagnetic induction of two electromagnets, called the sensors). The sensors are composed of two coils and capacitors, each embedded in a small cylinder. Probes are placed on the vertical midline of the face, parallel to each other (one on the forehead, one below the lower lip in the dimple above the chin). An electronic circuit converts the distance between the two probes into an output	- The signal is digitally linearized. - Can separate wake (high jaw activity) from healthy sleep (no jaw movement most likely) - Can be used as indirect marker of total sleep time. - Non-invasive technique that mirrors oesophageal pressure swings (oscillating jaw movements if respiratory events occur) - Improves the respiratory index calculation accuracy compared with an airflow and oxygen saturation analysis - Convenient - Cost-effective - non-invasive	- a lack of sensitivity in wake recognition because quite activity does not necessarily imply sleep-related brain activity - A decrease in performance when sleep is disturbed.

	voltage. The output voltage is a monotonic cubic function. Data can be expressed in absolute values (mm) or in normalised value (% of mouth opening. The reference value is the fully closed mouth level (zero).		
Tracheal sound sensors	See also airflow sensors. These sensors can also record suprasternal pressure (SSP), a good surrogate for evaluation of respiratory effort (based on tracheal wall vibrations). The low-pitched snoring sound signal and the high-pitched breathing sound signal can be extracted at different frequency bands from a raw tracheal sound signal. In addition, a non-audible, much lower frequency signal corresponding to SSP can also be derived using band pass filtering. This signal corresponds to pressure variations induced by	Non-invasive approach of respiratory effort.	Not well studied in adults.

	respiratory effort.		
Oesophageal pressure catheter [128, 137]	The gold standard for measuring respiratory effort, but usually not applied during cardiorespiratory polygraphy Changes in oesophageal pressure are estimates of changes in pleural pressure that occur during respiration. Utilizes air-filled balloons, fluid-filled catheters, or catheters with pressure transducers on their tips.	- direct measurement - can detect feeble respiratory efforts even when ribcage and abdominal movements are minimal - no requirement of special equipment	- Can influence the dynamics of the upper airway - Insertion uncomfortable and not always acceptable- Can interfere with the sleep and breathing pattern - Only available in a limited number of sleep centres. - currently considered as a research tool
Peripheral arterial tone (PAT) [8, 19, 20, 138-140]	Utilises the changes in peripheral vascular resistance and oximetry as indirect measures of respiratory signals	- comfortable for subject to wear - few loss of signals	- Relatively expensive compared to other respiratory sensors - contraindications with certain conditions (atrial fibrillation)/medications (alpha blockers) - devices utilise finger probes
Pulse Transit Time [141-145]	Is a surrogate marker for respiratory effort. It calculates changes in pleural pressure and detects autonomic arousals. Drops in systolic blood pressure occurring with inspiration (pulsus paradoxus) correlates well	- Some expertise is needed to perform it.	

	with the degree of inspiratory effort.		
		-	
Snoring sensors [90, 97, 146-150]			
Nasal cannula	Snoring may be detected as a high-frequency oscillation in the (unfiltered) nasal pressure signal. Measures snoring by pressure vibrations in the nares.		- Unfiltered signal must be used to detect snoring (the 200 Hz sampling rate of the nasal cannula is a limitation for detecting snore events). - Probable filter dampening effects - Effects of nasal cannula length, mouth breathing and movement in nares. - Snore events with lower strength (small deflection caused by snore vibrations compared with respiratory flow) and high fundamental frequency not all picked up. - Nasal cannula is capable of detecting only 55% of the snore events, while it's prone to also detect snore events when no snoring occurs (false positives). - No scoring rules for manual review of data.
Acoustic sensors (microphones)	Different types exist: - Capacitor (or condenser) microphones: have a membrane or flexible plate that moves with the air pressure	- Electret microphones are small-sized and cheap. - Contact microphones might be used in screening devices, while for appropriate analysis of snoring sound, the use of air-coupled microphones is	Non-standardised location of microphone (above the patients, on forehead, chest or on wall) Condensor microphones are rather

	variations. A direct-voltage is applied between the plates. The capacity changes that result from the movement of the membrane are converted into voltage. - Prepolarised or electrets microphones: do not require an external polarizing voltage. - Piezo-electric or ceramic microphones: make use of a membrane that is connected to a piezoelectric element. When the piezoelectric element moves, a voltage is generated.	mandatory. - Audio sensors require a high sampling rate of 8-10 kHz to analyse the acoustic characteristics of the snoring sound (compared to 200 Hz for the nasal cannula and the piezoelectric vibrational sensor) - Investigation of the effect of temperature conditions on the sensor response indicates that the output of the sensor needs no temperature correction over a moderate range of operating temperatures in spite of the fact that the piezoelectric coefficients are strongly temperature dependent. - Less influenced by the ambient noise. - Can also be used as tracheal sound sensor (detecting tracheal wall vibrations)	expensive. Their performance can vary depending on the position of the microphone: recorded sound frequency range depends on the placement of the microphone; higher frequencies are lost with microphones that have contact with the skin. Ambien noise also recorded (recording needs to be reviewed for false positive signals). Probable dampening effects of bed cover. Snore events with lower strength not all picked up. High processing cost of the necessary microcontroller
Piezo-electric vibration	Placed on the neck to detect	A piezo snoring sensor is also	- Snore events with lower strength

sensors	vibration. Transduce vibration into an electronic "sound" signal.	piezoelectric in type, but was made to acquire the vibration related to snoring with frequencies in the range of 5 to 50 Hz.	and high fundamental frequency not all picked up. - Detect 78% of the snore events, and almost all events detected are truly snore events. - Affected by low sampling rate (0-200 Hz). - Sometimes difficult to distinguish false positives from true snore events. - No scoring rules for manual overview. - Performance of piezoelectric sensors surpasses that of conventional foil type strain gages, with much less signal conditioning required. -
Body position [151-153]			
Mercury-switch based sensors	This sensor is a multicontact tilt switch containing a small mercury ball. Movement of the mercury ball resulting from body motion, causes openings and closures within the sensor as the ball touches the numerous sensor contacts. Sometimes incorporated	- Provides a signal directly proportional to the patient's position.	- Only coarsely resolve position ranges: categorise torso posture as supine, left-lateral, right-lateral or prone, each with a 90-degree range - Body position may not always represent the more important position of the head. - lack of an analytical technique to display and interpret finer resolution positional data and its relationship with sleep apnoea severity

	within the recording device itself or may be a separate belt-mounted device. Usually gravity-sensitive.		
Accelerometers:	An accelerometer is a damped mass, a proof mass, on a spring. When the accelerometer experiences an acceleration, the mass is moved to the point that the spring can push (accelerate) the mass at the same speed as the casing. The measurement of the spring's compression measures the acceleration. The system is damped so that oscillations (wiggles) of the mass and spring do not affect the needed measurements. Because of the damping, accelerometers always respond in different ways to different frequencies of acceleration. This is called the "frequency response." In mechanical accelerometers, measurement is often electrical, piezoelectric, piezoresistive or capacitive. Piezoelectric accelerometers		

	use piezoceramic sensors (e.g. lead zirconate) or single crystals (e.g. quartz, tourmaline). They are unmatched in high frequency measurements, low packaged weight, and resistance to high temperatures. Piezoresistive accelerometers resist shock (very high accelerations) better. Capacitive accelerometers typically use a silicon micro-machined sensing element. They measure low frequencies well.		
Pulse oximetry [105-107, 154-162]			
	Two types: Transmissive pulse oximetry (or transmission pulse oximetry): in this approach a sensor device is placed on a thin part of the patient's body (fingertip, earlobe, or an infant's foot), with the light source and light detector placed opposite to one	- Simplicity of use and ability to provide continuous and immediate oxygen saturation values - Connectivity advancements allow to monitor without a cabled connection to a monitor. - Accurate down to about 70-80%. Values below 70-80% probably do not correspond accurately to the actual SaO_2.	- Erroneously low reading or false reading may be caused by hypoperfusion of the extremity or from vasoconstriction, incorrect sensor application, highly calloused skin, nail polish, extraneous light intrusion, misalignment/misplacement, movement. - Error rates may be higher for adults with dark skin colour - COPD may cause false readings - Pulse oximeters differ in their

	another. Reflectance pulse oximetry oximetry (or reflection pulse oximetry): does not require a thin section of the person's body (application on the feet, forehead and chest). The light sources and the photodetector are located on the same surface of the skin, with the light source and light detector placed next to one another. Oxygen saturation changes are estimated by changes in the absorption of light of (at least) two distinct wavelengths (typically one in the red and the other one in the infrared spectrum) by oxygenated and deoxygenated blood. The oximeter probe emits a light that shines through the nail bed and is picked up by a light detector on the opposite side of the finger. Use of finger or ear probes: ear oximetry responds faster than finger oximetry. Maximum acceptable signal	- Pulse rate can also be derived, as well as the beat-to-beat signal derived from finger-plethysmography. - Variation in amplitude in the photoplethysmogram (PPG) is from variation in blood volume in the skin (caused by the pressure pulse of the cardiac cycle) and from respiratory induced variation. - The height of the photoplethysmogram is proportional to the pulse pressure, the difference between the systolic and diastolic pressure in the arteries. - Multisite PPG offers significant potential for data mining (deep learning), as well as a range of other innovative pulse wave analysis techniques. - Many inexpensive consumer models are available, and tend to be accurate within a few percentage points. - Maximal sample rate, resolution, accuracy in the area of 70-100% SaO2: - Model 3100 WristOx	ability to provide accurate data during conditions of motion or low perfusion. - Reflectance pulse oximetry: vasodilatation and pooling of venous blood in the head due to compromised venous return to the heart causes a combination of arterial and venous pulsations in the forehead region, and can lead to spurious SaO2 results. - In patients with a slow heart rate, a little longer averaging time may be needed (at least a 3-beat average). - Nadir in SaO_2 usually follows apnoea or hypopnoea termination by approximately 6 to 8 seconds, secondary to circulation time and instrumental delay. - Oximeters average over several cycles before producing a reading - Oximetry estimate of heart rate may be much lower than the actual rate if the patient has atrial fibrillation or frequent premature beats (only every other beat may provide sufficient signal to oximetry probe for the oximetry software to detect a heartbeat - Relationship between SaO_2 and PaO2 is dependent on many factors (including $PaCO_2$, hydrogen ion concentration, 2,3-diphosphoglycerate concentration, temperature, changes in acid-

	averaging time of 3 seconds or less (at a heart rate of 80 beats per minute). Longer averaging times may reduce the ability to detect oxygen desaturations. Determines the absorption of two wavelengths of light (most commonly 660 nm and 940 nm, red and infrared respectively) by capillary blood. Oxygenated haemoglobin absorbs more infrared light and allows more red light to pass through. Deoxygenated haemoglobin allows more infrared light to pass through and absorbs more red light. The change in volume caused by the pressure pulse is detected by illuminating the skin with the light from a light-emitting diode (LED) and then measuring the amount of light either transmitted (at the finger tip) or reflected (as on the forehead) to a photodiode.	(Nonin, Plymouth, MN, USA): 1 Hz, 1%, ±2% - PSG system Somnolab (Weinmann, Hamburg, Germany): 16 Hz, 1%) - Model Pulsox 300i (Konica Minolta Sensing Europe, Nieuwegein, The Netherlands): 1 Hz, 0.1%, ±2% - PSG system (Compumedics, Singen, Germany): 1 Hz, %, ±2% - Model ChipOx (MCC, Gesellschaft für Diagnosesysteme in Medizin und Technik mbH & Co, KG, Karlsruhe, Germany): 100 Hz, 1%, ±2%. This device allows for the adjustment of the measuring dynamics. The user can choose between sensitive, normal and stable mode. - The relationship between the measured parameter and SaO2 is determined experimentally for each type of commercial pulse oximeter sensor by calibration. - The reflection pulse oximeter can be applied on any accessible site, and is an advantage in low	base status and abnormal haemoglobin) - Photoplethysmogram is rarely displayed and is nominally only processed to determine heart rate. - The shape of the photoplethysmography waveform differs from subject to subject, and varies with the location and manner in which the pulse oximeter is attached. - The camera of mobile app pulse oximeters can't measure the light reflection at two wavelengths, so the oxygen saturation readings are inconsistent for clinical use. - The bias (mean SpO2-SaO2 difference) and the error in precision (SD of the differences) were both below 4%. - In measuring dynamic events (apnoeas), different pulse oximeters do not record identical values. - There is a fluctuation range of up to factor 1.42 between devices. - The low level of accuracy in pulse oximetry can be attributed to the empirical calibration that is essential for the execution of conventional pulse oximetry. - Increasing the window length of an oximeter's moving average leads to a progressive underestimation of the ODI's obtained during sleep.

	Introduction almost 40 years ago.	peripheral perfusion conditions. - There is good agreement between indices of OSA that require ≥4% oxygen desaturation, calculated from either finger pulse oximetry or foread pulse oximetry.	- Caution is advised when comparing ODI between patients or when performing posttreatment reassessment in the same patient, unless the same oximeter and software algorithm have been used. - Differences in ODI between different oximeters are likely the result of signal processing rather than patient factors or manufacturer algorithms for scoring desaturations. - Significant differences in SaO2 and %time<90% SaO2 between transmission and reflectance pulse oximetry. - SaO2 may be different at the forehead and finger as temperature and blood flow influence SaO2 and may differ. - Forehead pulse oximetry is particularly susceptible to data loss during movement (as during position changes) possibly due to changes in venous blood flow to the forehead.
ECG [163, 164]			
	Conductive pads attached to the body surface. These electrodes detect the	- Enables to detect heart rhythm abnormalities and ST segment changes - Electrode position and lead	- Changes in body position are known to result in ST segment fluctuations, rendering the tracing unreliable for assessment of

	small electrical changes that are a consequence of cardiac muscle depolarisation followed by repolarisation during each cardiac cycle (heartbeat). Two types of electrodes are in common use: - a flat paper-thin sticker - a self-adhesive circular pad. The former are typically used in a single ECG recording while the latter are for continuous recordings as they stick longer. Each electrode consists of an electrically conductive electrolyte gel and a silver/silver chloride conductor. The gel typically contains potassium chloride – sometimes silver chloride as well – to permit electron conduction from the skin to the wire and to the electrocardiogram. A single lead II using torso electrode placement is preferable for routine sleep	may be adjusted at the practioner's discretion.	ischemic changes - The amplitudes of clinically significant ST changes are as small as 1 mm, therefore muscle or movement artefacts would hinder reliability in assessing waveform changes

	studies. ECG voltages measured across the body are very small. This low voltage necessitates a low noise circuit, instrumentation amplifiers, and electromagnetic shielding.		

Table e24: Type III device studies with PSG Comparison or different scoring methods

Study	Demographics: n/ sex (%Female)/ Age/	Device (Channel n)/Comparison	Sensitivity/ specificity (%)	Level of Device to PSG AHI	Positive Predictive Value (%)	Negative Predictive Value (%)	AHI Correlation: Comments

	Population			**Agreement**			
Vat et al. (2015) [15]	312 (50%) Age 61.2 ± 10.7 Subset of population-based sample – stratified for age, gender and AHI severity.	**PSG (Titanium, Embla)** with some channels removed**.** (4: nasal pressure, pulse oximetry, 2 effort belts, body position) **Reference:** **PSG AASM 2012 Recommended** Denominator: MT based on self-reported lights. Comparison: PSG with channels removed vs. PSG Comparison of different hypopnoea rules: a) 3% desat b) 3% desat or 30% PWA drop	**AHI≥5** a) 99.1/81.3 b) 100/22.5 c) 77.2/98.8 d) 98.7/51.3 **AHI≥15** a) 90.1/96.3 b) 99.3/76.3 c) 54.0/100 d) 77.6/96.9 **AHI≥30** a) 70.3/100 b) 89.2/95.8 c) 29.7/100 d) 50/99.6	**PSG AHI – PM AHI** a) ΔAHI: 1.3 (CI: -8.2-10.8) b) ΔAHI: -3.5 (CI: -14-7) c) ΔAHI: 7.6 (CI: -7-22.2) d) ΔAHI: 2.8 (CI: -11.7-17.4) **a, c and d underestimated PSG-determined AHI; b resulted in overestimation.**	**AHI≥5** a) 93.9 b) 78.9 c) 99.4 d) 85.5 **AHI≥15** a) 95.8 b) 79.9 c) 100 d) 95.9 **AHI≥30** a) 100 b) 86.8 c) 100 d) 97.4	**AHI≥5** a) 97.0 b) 100 c) 59.9 d) 93.2 **AHI≥15** a) 91.1 b) 99.2 c) 69.6 d) 82.0 **AHI≥30** a) 91.5 b) 96.6 c) 82.1 d) 86.5	PM-derived AHI values using standard hypopnoea criteria (3% desat without PWA drops) can correctly classify OSA patients with an accuracy of > 93%. Including PWA drops as a surrogate for EEG arousal showed a higher sensitivity only for identification of severe OSA but do not seem to substantially increase overall accuracy.

		c) 4% desat d) 4% desat or 30% PWA drop					
Xu et al. (2017) [43]	80 (22.5%) Age 47.6 ± 14.0 Chinese adults with suspicion of OSA	**Nox T3** (Nox Medical, Iceland) (7: nasal pressure, pulse oximetry, 2 effort belts, snoring, body position, activity and heart rate) **Reference: PSG hypopnoea: 30% flow reduction + 4% desat (a) OR 30% flow reduction + 3% desat and/or arousal (b)** Denominator: MT based on questionnaire and activity. Comparison: Simultaneous device – PSG measurement.	**AHI≥5** a) 97/75 b) 100/40 **AHI≥10** a) 96/100 b) 92/73 **AHI≥15** a) 100/94 b) 96/83 **AHI≥30** a) 97/98 b) 94/95	**AHI correlation** a) R^2: 0.96 c) R^2: 0.98 vs. manual scoring (a)	**AHI≥5** a) 95 b) 96 **AHI≥10** a) 100 b) 95 **AHI≥15** a) 95 b) 93 **AHI≥30** a) 97 b) 94	**AHI≥5** a) 82 b) 100 **AHI≥10** a) 92 b) 62 **AHI≥15** a) 100 b) 91 **AHI≥30** a) 98 b) 95	Close agreement between PSG and simultaneous PM recording. Close agreement between automatic and manually edited scoring (< 2 events/h) potentially increases testing efficiency by reducing amount of time needed to edit the automatic score.

		a) manual scoring: (hypopnoea: 30% flow reduction + 4% desat) b) manual scoring: (hypopnoea: 30% flow reduction + 3% desat) c) automatic scoring (Noxturnal software): (hypopnoea: 30% flow reduction + 4% desat) vs. manual scoring (a)					
Zhao et al. (2017) [2]	30 (NA) Age 68.2 ± 9.1 Selected PSG studies from MESA study: - AHI<10 (n=10) - AHI 10-15 (n=10) - AHI 15-30 (n=10)	**PSG (Somte System, Compumedics)** with some channels removed**.** (6: nasal pressure, thermistor, pulse oximetry, 2 effort belts, ECG, limb movements) **Reference: hypopnoea: 30% flow reduction + 3% desat** Denominator: MT based on estimated wake/artefact or MT based on lights Comparison: PSG with channels removed vs.	**AHI≥5** a) 100/100 b) 70/100 **AHI≥15** a) 91/100 b) 40/100	ICC for AHI: a) 0.98 b) 0.98	**AHI≥5** a) 100 b) 100 **AHI≥15** a) 100 b) 100	**AHI≥5** a) 100 b) 78 **AHI≥15** a) 87 b) 77	Manual editing of TRT reduces the overestimation of TST and results in a respiratory event index that more closely approximates the AHI from PSG.

		PSG a) TST = TRT – periods of probable wakefulness and artefact (sleep onset based on artefact, heart rate and rhythmic breathing) b) TRT based on lights on and lights off					
Sabil et al. (2018) [3]	160 (22.3%) Age 48.8 ± 13.7 Adults with suspicion of OSA	**PSG (CID102L8D, Cidelec)** with some channels removed**.** (9-11: thermistor, nasal pressure, pulse oximetry, 2 effort belts, body position, limb movements, actigraphy, light, snoring (a) + 1 EEG and suprasternal pressure for (b)) **Reference:** **PSG AASM 2012 Recommended** Denominator: TRT or HypnoLight algorithm sleep time Comparison:		a) 48 pat changed category (no/mild, moderate, severe OSA) b) 27 pat of these 48 pat were successfully reclassified			Fully automated analysis of single-lead EEG channel combined with HSAT was reliable for sleep/wake identification and improved AHI calculation compared with HSAT.

		PSG with channels removed vs. PSG a) AHI using TRT; hypopnoeas: 3% desat b) AHI using TST detected by HypnoLight algorithm; hypopnoeas: 3% desat					
Zhang et al. (2020) [17]	262 (55.7%) Age 48.4 ± 14.9 Adults with suspicion of OSA.	**WatchPAT 200** (Itamar Ltd, Israel) (5: PAT, pulse rate, pulse oximetry, actigraphy, snoring) **Reference: PSG AASM 2012 Recommended** Denominator: TRT Comparison: Simultaneous device – PSG measurement. a) automatic PAT algorithm b) automatic PAT algorithm with manual adaptations	**AHI≥5** a) 96/29 b) 93/60 **AHI≥15** a) 88/66 b) 93/64 **AHI≥30** a) 65/84 b) 51/98	**AHI Spearman's correlation** a) 0.65 b) 0.81			Manual editing of WatchPAT automated scoring improves agreement with PSG-derived AHI across age and sex strata.

Cairns et al. (2014) [39]	32 (44%) Age 46.8 ± 12.3 Adults with suspicion of OSA.	**Nox T3** (Nox Medical, Iceland) (6: nasal pressure, pulse oximetry, 2 effort belts, snoring, body position, activity) **Reference: PSG AASM 2007 Recommended** Denominator: MT based on PSG lights. Comparison: Simultaneous device – PSG measurement. a) Autoscoring b) Manual scoring (hypopnoeas 4% desat)	**AHI≥5** a) 100/70 b) 100/70 **AHI≥15** a) 92/85 b) 92/90	**AHI correlation** a) R^2: 0.86	**AHI≥5** a) 88 b) 88 **AHI≥15** a) 79 b) 85	**AHI≥5** a) 100 b) 100 **AHI≥15** a) 94 b) 95	Very good agreement between T3 and PSG with an accurate autoscore algorithm.

Driver et al. (2011) [44]	73 (59%) Age 53 ± 12 Adults with suspicion of OSA.	**Medibyte** (Braebon Medical Corporation, Canada) (4: nasal pressure, oximetry, 2 effort belts, body position) **Reference: PSG AASM 2007 Alternative.** Denominator MT based on PSG lights. Comparison: Simultaneous device – PSG measurement. AHI (hypopnoeas 30% flow reduction + 3% desat)	**AHI≥5** 61/97 **AHI≥15** 80/97 **AHI≥30** 70/100	**PM AHI – PSG AHI** a) ΔAHI: 5.9 ± 11.2	**AHI≥5** 94 **AHI≥15** 97 **AHI≥30** 100	**AHI≥5** 80 **AHI≥15** 76 **AHI≥30** 88	Correct evaluation of absence or presence of severe OSA in 88% and 100% of patients.
Chang et al. (2019) [40]	88 (11%) Age 66.5 ± 7.8 Adults with	**Nox T3** (Nox Medical, Iceland) (7: nasal pressure, pulse oximetry, 2 effort belts, snoring, body position, activity and heart rate)	**AHI≥5** a) 96/100 b) 94/73 **AHI≥10**	**AHI correlation** a) R^2: 0.949	**AHI≥5** a) 91 b) 91 **AHI≥10**	**AHI≥5** a) 93 b) 80 **AHI≥10**	Good performance in diagnosing OSA in COPD. Significantly lower mean oxygen saturation

	COPD.	**Reference PSG hypopnoea: 30% flow reduction + 4% desat (a) OR 30% flow reduction + 3% desat and/or arousal (b)** Denominator: MT based on questionnaire and activity. Comparison: Simultaneous device – PSG measurement. Automatic + manual scoring. a) hypopnoea: 30% flow reduction + 4% desat b) hypopnoea: 30% flow reduction + 3% desat	a) 95/98 b) 90/91 **AHI≥15** a) 95/98 b) 95/93 **AHI≥30** a) 96/98 b) 93/98	b) R^2=0.933	a) 98 b) 94 **AHI≥15** a) 97 b) 93 **AHI≥30** a) 96 b) 96	a) 96 b) 86 **AHI≥15** a) 96 b) 95 **AHI≥30** a) 98 b) 97	compared to PSG, likely due to differences among pulse oximeters.
De Oliveira (2009) [32]	121 (31%) Age 45 ± 11 Adults with suspicion of OSA.	**Somnocheck** (Weinmann, Germany) (4: nasal pressure, pulse oximetry, body position and heart rate) **Reference PSG AASM 1999**	**AHI≥5** 96/65 **AHI≥10** 91/83	Kappa AHI: 0.53	**AHI≥5** 94 **AHI≥10** 93	**AHI≥5** 73 **AHI≥10** 68	

		Denominator: MT based on questionnaire and position. Comparison: Device – PSG measurement on different nights	**AHI≥15** 81/83 **AHI≥30** 80/92		**AHI≥15** 88 **AHI≥30** 86	**AHI≥15** 73 **AHI≥30** 89	
Ayappa (2008) [12]	92 (29%) Age pat: 46 – volunteers: 36 Adults with suspicion of OSA and volunteers.	**Ares™ unicorder** (6: nasal pressure, reflectance oximetry, snoring, actigraphy, head position and pulse rate) **Reference PSG a) AASM 2007 recommended OR b) hypopnoea 50% flow reduction or any flow reduction + 4% desat and/or arousal** Denominator: MT based on actigraphy and algorithm. Comparison: Simultaneous device – PSG	**AHI≥5** a) 98/84 **AHI≥10** a) 97/85 b) 91/87 **AHI≥15** a) 92/95 b) 95/94	ICC AHI: a) 0.96 b) 0.93			Additional manual scoring did not produce large changes in the overall SDB indices.

		measurement. Automatic + manual scoring a) hypopnoea: 50% flow reduction + 4% desat b) hypopnoea: 50% flow reduction + 1% desat or surrogate arousal					
Aurora et al. (2018) [37]	53 (47%) Age: 59.0 ± 12.9 Adults with heart failure.	**Apnealink plus** (Resmed) (3: nasal pressure, pulse oximetry, respiratory effort) **Reference PSG a) AASM 2012 Recommended OR b) hypopnoea 30% flow reduction + 4% desat and/or arousal** Denominator: TRT Comparison: Simultaneous device – PSG measurement. Manual scoring. a) hypopnoea: 30% flow	**AHI≥5** a) 96/80	ICC AHI: a) 0.94 b) 0.96 Difference: a) ΔAHI: -3.6 (CI: -6.0 - -1.2) b) ΔAHI: -4.2	**AHI≥5** a) 98	**AHI≥5** a) 67	High degree of agreement also for metrics of both obstructive and central sleep apnoea.

		reduction + 3% desat b) hypopnoea: 30% flow reduction + 4% desat.					
Aurora et al. (2015) [38]	a) 100 (35%) b) 100 (33%) Age: a) 55.8 b) 60.8 a) Adult community based cohort. b) Patients from a cardiology clinic.	**1) Apnealink plus** (3: nasal pressure, pulse oximetry, respiratory effort) **2) Embletta** (5: nasal pressure, thermistor, pulse oximetry, respiratory effort, body position) **Reference: 30% flow reduction with either 3% desat or 4% desat** Denominator: TRT		CC AHI: 1) a) 0.97 b) 0.98 2) a) 0.64 b) 0.76 Difference: 1) a) ΔAHI: 6.1 (CI: 4.9-7.3) b) ΔAHI: 4.6 (CI: 3.5-5.6)			Input by a specialist in reviewing home sleep study results may help to improve diagnostic accuracy and classification of OSA.

		Comparison: No comparison with PSG. Automatic PM scoring vs. Manual PSG a) hypopnoea: 30% flow reduction + 3% desat b) hypopnoea: 30% flow reduction + 4% desat		2) a) ΔAHI: 5.3 (CI: 3.2-7.3) b) ΔAHI: 8.4 (CI: 7.2-9.6)			
Boyd et al. (2016) [18]	28 (25%) Age: 51.4 ± 10.8 Patients with severe OSA treated with CPAP.	**Watchpat-200** (Itamar, Israel) (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy) **Reference PSG: AASM 2007 (14% recommended; 86% alternative)** Denominator: TRT Comparison: Simultaneous device – PSG measurement. Automatic scoring		CC effective AHI: r=0.871 (CI: 0.73-0.94) ΔAHI: 2.1 ± 8.2			Watchpat-200 provides a reasonable accurate measure of the effective AHI.

Bridevaux et al. (2007) [165]	11 (0%) Age: 54 ± 14 Patients with suspected OSA.	**Embletta pds** (Resmed, Iceland) (5: nasal pressure, respiratory effort, pulse oximetry, pulse rate, body position) **Reference: manual scoring (hypopnoea: 50% flow reduction or less flow reduction + 3% desat)** Denominator: TRT Comparison: No comparison with PSG. Automatic scoring and agreement among observers		ICC AHI between observers: 0.73 Automatic vs. manual: ΔAHI: -1.2			Inter-observer agreement on AHI derived from PM is limited in a clinical setting. For patients with AHI score between 10 and 20 a second review of the tracing could be recommended.
Ito et al. (2018) [45]	28 (25%) Age: 59 (50-68) Patients with suspected OSA.	**Smart Watch PMP-300E** (Fuji-Respironics, Japan) (5: nasal pressure, respiratory effort, pulse oximetry, pulse rate, body position)		CC AHI: 0.92 CC AI: 0.95 CC HI: 0.43			Strong correlation between AHI from PSG and type III device. Poor correlation for the HI component of

		Reference: manual scoring (hypopnoea: 50% flow reduction + 3% desat or arousal) Denominator: TRT Comparison: Device – PSG measurement on different nights. Manual scoring.		ΔAHI: -8.52 (-11.27, -4.93)			the AHI.
Aielo et al. (2019) [4]	300 (55%) Age: 48 ± 8 Employees of university (frequency of OSA: 27.3%).	**Embletta Gold** (Natus Medical, Canada) (5: nasal pressure, respiratory effort, pulse oximetry, snoring, body position) **Reference: AASM 2012 with TRT based on actigraphy (Actiwatch model 2, Philips Respironics)**		a) CC AHI: r=0.996 Agreement AHI: Kappa=0.95 b) CC AHI:			Excellent agreement and correlation after three different scoring strategies.

		Denominator: MT based on event button or diary Comparison. No comparison with PSG. Manual scoring AASM 2012. a) TRT based on event button (lights on-off) b) TRT based on diary		r=0.993 Agreement AHI: Kappa=0.96			
Anitua et al. (2019) [49]	99 (44%) Age: 56 ± 14 Patients with suspected OSA.	**APNiA** (BTI, Spain) (5: nasal pressure, pulse oximetry, pulse rate, snoring, body position) **Reference: AASM 2012 (hypopnoea 30% flow reduction + 3% desat), first night home monitoring** Denominator: TRT Comparison:		ICC average AHI: 0.954 ICC single measures AHI: 0.874 SEM AHI: 4.64/h Variability OSA severity (Kappa): Night 1-2: 0.442			Considerable individual variability in AHI. Use of standard error of measurement and AHI of a single night could be helpful in predicting the most frequent OSA severity among different nights.

		No comparison with PSG. Three consecutive nights of monitoring. Evaluation of night-to-night variability.		Night 1-3: 0.554 Night 2-3: 0.536			
Pang et al. (2007) [23]	37 (68%) Age: 50 ± 12 Patients with suspected OSA.	**WatchPat** (Itamar, Israel) (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy) **Reference: PSG (hypopnoea 30% flow reduction + 4% desat and/or arousal)** Denominator: MT based on actigraphy. Comparison: Simultaneous device – PSG measurement. Automatic scoring.	**AHI≥5** a) 94/80 **AHI≥15** a) 96/79	CC AHI: 0.93			High correlation for AHI between WatchPat and PSG in this study.

Jiang et al. (2018) [25]	33 (31%) Age: 49 ± 13 Patients with suspected OSA.	**PRM (mode 4)** (Peking University People's Hospital) (5: thermistor, actigraphy, chest movement, position, pulse oximetry) **Reference: PSG AASM 2012 Recommended** Denominator: MT based on actigraphy. Comparison: Simultaneous device – PSG measurement. Automatic scoring:	**AHI≥5** 97/100 **AHI≥15** 100/89 **AHI≥30** 95/92	Kappa (AHI for AHI ≥ 5): 0.85	**AHI≥5** 100 **AHI≥15** 96 **AHI≥30** 95	**AHI≥5** 67 **AHI≥15** 100 **AHI≥30** 92	Using accelerometry to evaluate sleep time could improve the accuracy of AHI evaluation.

		– estimated sleep time based on actigraphy					
Hedner et al. (2011) [7]	38 normal subjects and 189 patients with OSA (NA) Age: 49 ± 14	**Watchpat-100** (Itamar, Israel) (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy) **Reference: PSG AASM 1999** Denominator: MT based on actigraphy. Comparison: Simultaneous device – PSG measurement. Automatic scoring.		ICC RDI: 0.87 Agreement REM sleep: 88.7% Agreement Light/Deep sleep:88.6% Overall sleep stage agreement: 66%			Moderate accuracy when using PAT and actigraphy signals to detect different sleep stages.

Garcia -Diaz et al. (2007) [5]	62 (13%) Age: 54 ± 10 Patients with suspected OSA.	**Apnoescreen II** (Erich Jaeger GMBH & CoKg, Germany) (8: thermistor, pulse oximetry, pulse rate, respiratory effort, snoring ECG, body position, actigraphy) **Reference: PSG AASM 2007 adapted (hypopnoea: 50% flow reduction + 4% desat or arousal)** Denominator: TRT or MT based on actigraphy. Comparison: Simultaneous device – PSG measurement (as well as separate night measurement). Manual scoring. a) Total recording time b) Sleep estimated time	**AHI≥10** a) 92-95/92-96 b) 95-95/88-96 **AHI≥15** a) 97/97 b) 100/97 **AHI≥30** a) 92/95 b) 96/95				Wrist actigraphy contributed little to improving the efficacy of RP.

		(actigraphy)					
Kinoshita et al. (2018) [166]	61 (21%) Age: 57 ± 14 Patients with suspected OSA.	**Watchpat-200** (Itamar, Israel) (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy) **Reference PSG: AASM 2007 Alternative** Denominator: MT based on actigraphy. Comparison: Device – PSG measurement on different nights	**AHI≥30** 79/80	CC AHI: r=0.69 For PWV<1500 CC AHI: r=0.78 For PWV>1500 CC AHI: r=0.40			Arterial stiffness may affect the respiratory event index measured by the WatchPAT.

		Automatic scoring					
Tiihonen et al. (2009) [42]	10 (50%) Age: 47 ± 13 Patients with suspected OSA.	**APV2** (Remote Analysis Oy, Finland) (7: flow, pulse oximetry, pulse rate, respiratory effort, snoring, body position, actigraphy) **Reference PSG: AASM 2007 Recommended (hypopnoea 30% flow reduction + 4% desat)** Denominator: MT based on actigraphy. Comparison: Simultaneous device – PSG measurement. Manual scoring.		CC AHI: r=0.997			This device was found clinically applicable, technically reliable and sensitive for the diagnostics of OSA.

Masa et al. (2011) [51]	348 (24%) Age: 49 ± 12 Patients with suspected OSA.	**Breas SC20** (Breas Medical AB, Sweden) (7: flow, pulse oximetry, respiratory effort, body position) **Reference PSG: AASM 2012 Recommended (hypopnoea 30% flow reduction + 3% desat or arousal)** Denominator: TRT Comparison: Simultaneous device – PSG measurement (as well as separate night measurement). Manual scoring.	**AHI≥5** 96/57 **AHI≥10** 97/39 **AHI≥15** 94/60	**AUC AHI≥5:** 0.917 **AUC AHI≥10:** 0.883 **AUC AHI≥15:** 0.891			HRP produced lower AHI's than PSG. HRP can exclude or confirm the diagnosis of OSA; and for equal diagnostic efficacy, the cost of HRP is half or less than that of PSG.
Masa et al. (2013) [14]	342 (25%) Age: 49 ± 12	**Breas SC20** (Breas Medical AB, Sweden) (7: flow, pulse oximetry,	**To recommend treatment:**	**AUC AHI≥5:** a) 0.916 b) 0.909			Incorporating a surrogate arousal criterion into the definition of hypopnea in PM did not

	Patients with suspected OSA.	respiratory effort, body position) **Reference PSG: AASM 2012 Recommended (hypopnoea 30% flow reduction + 3% desat or arousal)** Denominator: MT based on valid signal quality. Comparison: Simultaneous device – PSG measurement (as well as separate night measurement). Manual scoring. a) Hypopnoeas (30% flow reduction + 3% desat) b) Hypopnoeas (30% flow reduction + 3% desat or surrogate arousal (resolution of airflow or band reduction by a sudden increase in amplitude and frequency ≥ 2 breaths)	a) 74/79 b) 81/76	**AUC AHI≥10:** a) 0.889 b) 0.885 **AUC AHI≥15:** a) 0.896 b) 0.894 **Agreement level** **Total sample:** a) 76 b) 80 **AHI ≥30:** a) 0.91 b) 0.93			substantially increase agreement with PSG.

Lachapelle et al., 2019 [13]	178 (50%) Age: 49+/- 13 Patients with moderate-high pre-test probability of OSA + inconclusive Type 3 study	**Device: Embletta or Embletta Gold recorders** (Natus, Embla, Mississauga Ont) (5: Nasal pressure, pulse oximetry, thoracic and abdominal respiratory belts, body position) **Reference: PSG - AASM 2007 except for respiratory events (AASM 1999).** Dominator: TRT Comparison: Device – PSG measurement on different nights. Manual scoring a) Hypopnoeas: Either 50% flow reduction or Discernable flow reduction (but <50%) + >3% desat b) Hypopnoeas: Either 50% flow reduction or Discernable flow reduction	**AHI≥5** a) 42/89 b) 77/57 **AHI≥10** a) 14/94 b) 49/76 **AHI≥15** a) 7/99 b) 17/94 **AHI≥30** a) --/-- b) --/--	**PSG AHI – PM AHI** a) ΔAHI: 11.2 (95%CI 33.6, −11.1) b)ΔAHI: 7.1 (95%CI 29.6, −15.4) **AUC AHI≥5** a) 0.66 b) 0.61 **AUC AHI≥10** a) 0.54 b) 0.62 **AUC AHI≥15** a) 0.50 b) 0.54	**AHI≥5** a) 95 b) 91 **AHI≥10** a) 80 b) 77 **AHI≥15** a) 83 b) 68 **AHI≥30** a) --/-- b) --/-	**AHI≥5** a) 22 b) 32 **AHI≥10** a) 40 b) 47 **AHI≥15** a) 60 b) 62 **AHI≥30** a) --/-- b) --/-	Study reported diagnostic accuracy was improved with surrogate arousal scoring as reflected by an increased area under the receiver-operating characteristic curve for AHI thresholds of 10 and 15 events/h. However AUC for threshold of 5 was not improved, and study was conducted using Chicago criteria.

		(but <50%) + >3% desat or microarousal (used increase in pulse oximetry-derived heart rate ≥ 6 beats/min as a surrogate marker for arousal)		**AUC AHI≥30** a) --- b) --			
Miller et al., 2018 [30]	83 (48%) Age: 54.5±15 Adults scheduled for OSA consultation at sleep clinic.	**Device: ApneaLink Air** (ResMed, San Diego, CA) (3:Nasal pressure, thoracic band, oximetry) **Reference: PSG. AASM 2012 Recommended** Dominator: Not described. Comparison: Device – PSG measurement on different nights Automatic PM vs. Manual PSG. Scoring criteria not described	**AHI≥5** a) 82/57 **AHI≥10** a) --/-- **AHI≥15** a) 79/86	**r = 0.750, p = 0.01**	**AHI≥5** a) 89 **AHI≥10** a) -- **AHI≥15** a) 86	**AHI≥5** a) 42 **AHI≥10** a) -- **AHI≥15** a) 80	Study compared measures used in OSA screening (Berlin, Epworth Sleepiness Scale (ESS), STOP Bang) and a PM to AHI and levels from PSG PM and PSG conducted on separate nights. Reported that PM consistently predicted the presence of OSA but there was low sensitivity at AHI levels >=30, indicating some patients with severe OSA would have a

		for PM (automatic).	**AHI≥30** a) 60/87		**AHI≥30** a) 63	**AHI≥30** a) 86	missed diagnosis
Nagubadi, et al. 2016 [41]	71 (41%) Age: 52±10 Inpatients with suspected SDB	**Device: Alice PDx device** (Philips Respironics, Murrysville, PA) (6:Flow (Nasal pressure and thermistor), RIP, pulse oximetry, Actigraphy) **Reference: AASM 2012 Acceptable** Dominator: MT based on actigraphy. Comparison: Device – PSG measurement on different nights. Manual Scoring.	**AHI≥5** a) --/-- **AHI≥10** a) --/-- **AHI≥15** a) 69/87 **AHI≥30**	**ICC = 0.7** **CC = 0.68** **AUC AHI≥15:** 0.8 **AUC AHI≥30:** 0.82 Mean±SD difference between AHI-PM and AHI-PSG: 2±29	**AHI≥15** a) 92 **AHI≥30** a) 76	**AHI≥15** a) 57 **AHI≥30** a) 81	The median time between the two studies was 97 days (IQR 25–75: 24–109) Used acceptable not recommended criteria. Reported that PM was accurate in detecting moderate and severe SDB in in-patients but was better in patients that did not have significant CSA.

		PM criteria: Apnoea: Cessation of airflow during sleep for more than 10 seconds. Hypopnoea: At least a 10 second discernible reduction in flow + 4% desaturation	a) 87/66				
Masa, 2013 (ERJ) [52]	348 (24%) Age: 49 ± 12 Patients with suspected OSA.	**Breas SC20** (Breast Medical AB, Sweden) (5: flow, pulse oximetry, respiratory effort, body position) **Reference: PSG. Equivalent to AASM 2012 Recommended (hypopnoea: 30% flow reduction + 3% desat or arousal)** Denominator: MT based on valid signal quality. Comparison: Simultaneous device – PSG measurement (as well as separate night measurement).		**AUC AHI≥15:** **a)** 0.901 (95%CI 0.867-0.936) **b)** 0.850 (95%CI 0.806-0.893)			Manual PM scoring had better agreement with manual PSG scoring than automatic PM scoring In addition to AUC at cut-point of 15/h this study also reported similar differences at cut-points of 5 and 10/h (i.e. AUC difference = 0.06) but did not report actual AUC values. PM and PSG conducted on separate nights.

		PSG manual vs. (a) PM manual scoring and (b) PM automatic. PM criteria: Apnoeas: Absence of airflow (≥90% reduction) for ≥10 s Hypopnoeas: 30% flow reduction ≥3% desat					
Miyata et al., 2007 [28]	18 (0%) Age: 51+/-10.8 OSAS patients	**LT-200** (6:Thermistor, thoracoabdominal movement (air-bag pressure sensor), pulse oximetry), snoring, and body position) **Reference: PSG. Custom criteria; Hypopnoea: Airflow reduction + 4% desat or arousal** Dominator: Not described. Comparison: Device – PSG measurement on		r=0.94, p<0.0001; mean difference [PSG-PM](range): 4.3/h (–4.4/h to 13.1/h)			Reported that AHI and lowest SpO2 values obtained using PM showed a high level of agreement with those obtained by PSG. No sensitivity/specificity analysis. PM and PSG conducted on separate nights.

		different nights Probably manual scoring but did not specify. PM criteria: Not overtly stated but hypopnoea criteria likely airflow reduction + 4% desat.					
Norman et al, 2017 [6]	n=80 (45%) Age: 55 +/- 15* *Development set Patients with suspected SDB	**Type 1 or Type 2 device with channels removed to simulate type 3 device** (5:Nasal flow, thorax, abdomen, snoring, and oximetry) **Reference: PSG. AASM 2012 Recommended.** Dominator: TRT or calculated TST based on combination of movement, quiescent time and respiratory signals to estimate sleep. Comparison: PSG with channels removed vs. PSG.	**AHI≥15** a) 73/100 b) 97/100	a) mean difference 9.1 events/h (range −1.1 to 61.6); ICC = 0.924; R [2] = 0.919, p < 0.0001 b) mean difference 1.6 events/h (range −20.7 to 27.7); ICC = 0.990; R [2] = 0.962, p < 0.0001			Used a novel method to estimate TST and reduce denominator of AHI. Method involved movement and respiratory signals to estimate sleep. Compared to PSG (either home or lab) Minimizing the difference between denominators improved agreement with PSG.

		Manual scoring. PSG vs. (a) Type 3 (TRT), and (b) Type 3 (Estimated TST) PM Criteria: AASM 2012. Body movement from respiratory bands used as surrogate for arousal					
Oliveira et al., 2012 [33]	n=26 (50%) Age: 62.8+/-8.5 COPD patients with suspected SDB	**Stardust II** (5: Nasal pressure, respiratory effort (piezoelectric sensor), body position and SpO2 + heart rate) **Reference: PSG. ASSM 1999** Dominator: TRT Comparison: Simultaneous device – PSG measurement (as well as separate night measurement). Manual Scoring. In lab PSG vs. (a)	--	a) ICC: 0.61 (95%CI 0.28-0.8); Mean difference (PSG-simultaneous PM): -6.1 (+/-1.96SD: -34.1, 21.9) b) ICC: 0.47(0.11-0.72); Mean difference (PSG-home PM): -0.6 (+/-1.96SD: -30.7, 29.5)	--	--	Purported better agreement in simultaneous vs. home PM compared to PSG. Also reported better agreement in severe (AHI > 30 vs. 5-30/h) using kappa analysis however there was a large failure rate. Overall, PM overestimated AHI, with greater tendency to overestimate mild cases and underestimate severe cases There was no

		Simultaneous PM, and (b) Home PM. PM Criteria: Modified Chicago (No arousal requirement when hypopnoea flow reduction discernible but <50%)					sensitivity/specificity analysis
Pinna et al., 2014 [46]	n=67 (7%) Age: 59+/-8 Clinically stable, optimally treated, moderate-to-severe heart failure patients	**Embla Titanium** (Simulated Type 3). (6: Thoraco-abdominal movements, nasal airflow, oronasal airflow, oxygen saturation and body position) **Reference: Type 2 PSG. AASM 2007 Alternative.** Dominator: MT based on movement and standing position. Comparison: PSG with channels removed	**AHI≥5** a) 98.2/91.7 **AHI≥10** a) --/-- **AHI≥15** a) 88.2/100	Median (IQR) difference (Type3-Type2): -0.8 (-2.9, 0.4)	**AHI≥5** a) 98.2 **AHI≥15** a) 100	**AHI≥5** a) 91.7 **AHI≥15** a) 89.2	Reported good agreement between Type 3 and Type 2 PSG. Reduction in AHI caused by monitoring time denominator was offset by an increase in central events (not scored in PSG due to wakefulness). Authors noted very few arousal associated hypopnoeas were scored.

		vs. PSG Manual Scoring. PM Criteria: Modified AASM 2007 (no arousal criteria)					
Planes et al., 2010 [27]	n=45(1%) Age: 63.4+/-11.6 Patients undergoing percutaneous coronary intervention following an acute coronary event	**CID102L8** (Simulated Type 3) (6: nasal flow, breath sounds, thoracic and abdominal movements, O2 saturation, body position, and actimetry) **Reference: Type 2 PSG. Custom criteria (noticeable change in airflow with associated 3% desat or arousal).** Dominator: Described as "Valid Recording Time" Comparison: Simultaneous device – PSG measurement. Automated PM vs. Manual	**AHI≥5** a) 95/67 **AHI≥15** a) 71/93 **AHI≥30** a) 75/97	Mean difference (automated PM-PSG):−3.4± 7.5	**AHI≥5** a) 97 **AHI≥15** a) 96 **AHI≥30** a) 90	**AHI≥5** a) 50 **AHI≥15** a) 59 **AHI≥30** a) 91	Reported good agreement between Type 3 automatic scoring and Type 2 manual scoring. PM slightly underestimated PSG. Adding flow limitation events to PM index (based on flattening +breath sounds) resulted in overestimation vs. PSG AHI.

		PSG. PM criteria: Apnoea: Absence of breath sounds for ≥ 10-s duration; Hypopnoea: ≥ 30% decrease in nasal flow or heavy snoring on each breath +≥ 3% decrease in SaO2					
Facco et al., 2019 [50]	n=30 Obese pregnant women with a singleton pregnancy - before 21 weeks' gestation and then at 28-32 weeks' gestation	**ApneaLink** (type III) (3:respiration (nasal pressure transducer), a thoracic inductance plethysmography band, finger pulse oximetry.) **Reference PSG: Modified AASM 2012 Recommended (no arousal)** Dominator: TRT Comparison: Automated scoring and manual scoring Device – PSG measurement on		ICC: Device autoscore vs. PSG: 0.76 Device manual scoring vs. PSG: 0.7 Device autoscore vs. device manual scoring: 0.78 Categorical agreement: Device auto scoring vs. PSG: 25/30			Of 24 negative PSGs (AHI<5), 23 (95.8%) were also negative by automated device scoring, while 19 (79.2%) were also negative by manual scoring. Of the 6 positive PSGs, 2 were classified as positive by automated scoring, 4/6 were identified as positive by manual scoring. 2 studies were discordant by both manual and

		different nights Apnoea: ≥ 90%reduction in airflow for a minimum of 10 seconds Hypopnoea: ≥ 30% reduction in airflow for a minimum of 10 seconds, associated with ≥3% reduction in oxyhemoglobin saturation		(83.3%) Device manual scoring vs. PSG: 23/30 (76.7%) Device auto scoring vs. manual scoring: 24/30 (80.0%)			automated scoring. 50% of cases of mild OSA misclassified by device scoring (auto or tech) had AHI values of > 4 but < 5
Saletu et al. (2018) [53]	33 (42%) Age: 63 ± 5 Subacute adult stroke patients.	**SOMNOmedics** (SOMNOmedics GmbH, Germany) (6: flow (thermistor and nasal cannula), pulse oximetry, respiratory effort, ECG, actigraphy and body position) **Reference PSG: AASM 2012 Recommended (hypopnoea 30% flow reduction + 3% desat or arousal)** Denominator: MT -: total recording time minus artefacts and awake time (actigraphy)		CC AHI: 0.97 ΔAHI: -1.40 (+4.8, -7.61)			REI and AHI detected in the same PSG night demonstrated no significant differences.

		Comparison: Device – PSG measurement on different nights Manual scoring.					
Polese et al. (2013) [34]	43 (56%) Age: 70 ± 5 Elderly patients with suspected OSA.	**Stardust II** (PhilipsRespironics, USA) (5: flow (nasal cannula), pulse oximetry, respiratory effort, pulse rate and body position) **Reference: AASM 1999 (hypopnoea 50% flow reduction or any flow reduction + 3% desat or arousal)** Dominator: TRT. Comparison: Simultaneous device – PSG measurement (as well as separate night measurement). Manual scoring.	**AHI≥5** 100/0 **AHI≥15** 100/70 **AHI≥30** 90/68	CC AHI: 0.84 ΔAHI: -2.2 (+26.2, -30.5)	**AHI≥5** 100 **AHI≥15** 90 **AHI≥30** 71	**AHI≥5** 0 **AHI≥15** 0 **AHI≥30** 0.1	Effective and good diagnostic agreement with attended PSG in elderly population.

Weimin et al. (2013) [24]	28 (29%) Age: 48 ± 14 Patients with suspected OSA.	**Watchpat-200** (Itamar, Israel) (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy) **Reference: AASM 2007 (Recommended or Alternative not specified)** Dominator: MT based on actigraphy. Comparison: Simultaneous device – PSG measurement. Automatic scoring.		CC AHI: r=0.92 ΔAHI: 3.0 (+19.5, -13.6) AUC AHI≥5: 0.969 AUC AHI≥15: 0.930 AUC AHI≥30: 0.973			Reasonably accurate estimation of sleep and wakefulness in OSA patients.
Choi et al. (2010) [19]	n=25 (4 females) mean age 40.9 ± 11.2 years (range 21–59)	**Watch-PAT 100** (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy)	AHI≥5 100/83 AHI≥15 81/77	PSG vs. device 1) AHI r = 0.94, p<0.001	AHI≥5 95 AHI≥15 87	AHI≥5 100 AHI≥15 70	Different nights

	Adult patients suspected OSA	**Reference: AASM 2007 Recommended** Dominator: MT based on actigraphy. Comparison: Device – PSG measurement on different nights. Scoring: automatic computerized algorithm of the Watch-PAT 100 systems. Respiratory events: - termination of respiratory disturbances lead to the surge of sympathetic activity that influenced digital arterial vasoconstriction. - Vasoconstriction of the digital vascular bed by mediated alpha-receptors results in attenuation of PAT signal. - PAT signal attenuation,	AHI≥30 92/92	2) lowest saturation r = 0.90, p<0.001 severity of AHI Kendall tau-b = 0.897, p < 0.001	AHI≥30 92	AHI≥30 92	

		increased heart rate, and oxygen desaturation are analysed by the automatic computerised algorithm.					
Gan et al. (2017) [20]	N=20 (2 women) mean age 39 ± 16 years, range (18–70)	**Watch-PAT 200.** (5: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy, body position) **Reference AASM 2012 Recommended** Dominator: Unreported but likely MT based on actigraphy. Comparison: Simultaneous device – PSG measurement. Automatic scoring Respiratory event – one of the	AHI>5 100/75.0 AHI>15 84.6/100 AHI>30 80.0/100	AHI Spearman's coefficient 0.94 Total sleep time Spearman's coefficient 0.6228 (P<0.0034)			Bland–Altman plot: an AHI mean difference of about 4.23 with a slight tendency for the Watch-Pat200 to overscore the AHI at the mild range of OSA and to underscore the range at the severe end of OSA

		three are met: - 30% or greater reduction in PAT amplitude together with a pulse rate acceleration of 10%, - 30% or greater reduction in PAT amplitude together with a 3% oxyhemoglobin desaturation, - 4% oxyhemoglobin desaturation.					
Ng et al. (2010) [26]	N=80 (17 female)	**Embletta PDS** (3: Airflow (nasal cannula connected to a pressure transducer), effort sensor; built in sensor for body position.) **Reference: AASM 2007 Recommended** Dominator: TRT.	**AHI ≥ 5** 92.4/85.7 **AHI ≥ 10** 90.0/86.7 **AHI ≥ 15** 87.8/94.9 **AHI ≥ 20**	Pearson coefficient (*p<0.05) Overall r=0.979* **AHI<5** r=0.46 **AHI >5**	**AHI ≥ 5** 96.8 **AHI ≥ 10** 91.8 **AHI ≥ 15** 94.7 **AHI ≥ 20**	**AHI ≥ 5** 70.6 **AHI ≥ 10** 83.9 **AHI ≥ 15** 88.1 **AHI ≥ 20**	high specificity and negative predictive value at AHI of >20/h Small but statistically significant bias towards slightly lower scores at AHI >5 and >10 cut-offs.

		Comparison: Simultaneous device – PSG measurement. Automated scoring Device default settings for respiratory events: - apnoea - decrease in airflow by 80% of baseline for at least 10 s (max 80 s), - decrease in airflow by 50% of baseline for at least 10 s (max 100 s).	85.3/95.7	r=0.976* **AHI>10** r=0.969* **AHI>15** r=0.964 AHI>20 r=0.961	93.5	89.8	
O'Brien et al. (2012) [22]	N=31 (100% female) Pregnant female (third trimester)	**WatchPat-200** (4: peripheral arterial tonometry (PAT) signal, heart rate, oxyhemoglobin saturation, actigraphy) **Reference: AASM 2007 Alternative** Dominator: MT based on	**AHI ≥5** 88/87 **AHI≥10** 100/81	AHI r = 0.73, p < 0.001 total sleep time r = 0.76, p < 0.001 RDI r = 0.68, p < 0.001;	**AHI ≥5** 70 **AHI≥10** 50	**AHI ≥5** 95 **AHI≥10** 100	No moderate or strong correlations between sleep stages defined as stages 1 and 2 sleep (light sleep in the Watch-PAT), stage 3 sleep (deep sleep in the Watch-PAT), and REM sleep (light sleep r = 0.10, p = 0.60; deep sleep r = 0.32, p = 0.08; REM sleep r = 0.29, p = 0.12).

		actigraphy. Comparison: Simultaneous device – PSG measurement. Automated scoring. (1) PAT amplitude reduction occurred with acceleration in the pulse rate or increase in wrist activity; (2) PAT amplitude reduction occurred with ≥ 3% oxyhemoglobin desaturation; or (3) ≥ 4% oxyhemoglobin desaturation occurred		mean SpO2 r = 0.94, p < 0.001; minimum SpO2 r = 0.88, p < 0.001			The Watch-PAT had a tendency to classify apnoea severity as slightly worse than that reflected on the PSG, especially for RDI measurement.
Santos-Silva et al. (2009) [47]	N=80 (43% female) 3 recordings: only device, device + PSG, only PSG	**Stardust II** (Respironics, Inc., USA) (5: SpO2 (via finger probe), pulse rate (from the oximeter probe), airflow (pressure based airflow through a nasal cannula), respiratory effort (piezoelectric sensor in a belt placed mid-thorax), and body position (mercury switch built into the	**AHI ≥ 5** 98/62 **AHI ≥ 15** 97/74 **AHI ≥ 30**	r = 0.892; P < 0.0001; 95% CI = 0.83 to 0.93 AUC = 0.97, 0.98, and 0.98, respectively	**AHI ≥ 5** 87 **AHI ≥ 15** 78 **AHI ≥ 30**	**AHI ≥ 5** 93 **AHI ≥ 15** 96 **AHI ≥ 30**	Correlation between AHIs from both the PSG lab and PSG+device lab was high (r = 0.89; P < 0.005)

		STD unit and worn mid-sternum).) **Reference: AASM 1999** Dominator: MT based on event marker. Comparison: Simultaneous device – PSG measurement (as well as separate night measurement). Manual scoring. Hypopnoea: 50% or discernable decrement in airflow lasting ≥ 10 sec with a 3% reduction in SpO2. Apnoea: cessation of airflow ≥ 10 sec (whether central, obstructive, or mixed).	96/93		87	98	

To et al. (2009) [11]	N= 175 (25% female) Predominant OSAS	**ARES** (Advanced Brain Monitoring, Carlsbad, CA, USA) (7: Worn on the forehead. Measures blood oxygen saturation, pulse rate, airflow and respiratory effort, snoring levels, head movement and head position) **Reference: AASM 2007 Alternative** Dominator: TRT. Comparison: Simultaneous device – PSG measurement. Apnoea: cessation of airflow for > 10 s; 1% desaturation was included if the obstructive event was accompanied by changes in pulse rate, head positions or snoring sounds, which implied arousals	Desaturation ≥4% as scoring criteria 84 (95% confidence interval (CI): 77–90)/ Desaturation ≥3% as scoring criteria 89 (95% CI: 89–94)/ Desaturation ≥1% + surrogate arousal criteria 97 (95% CI: 94–99)/ 63 (95% CI: 55–71)	Desaturation ≥ 4% as scoring criteria K coefficient (AHI) 0.24, P < 0.01 Desaturation ≥ 3% as scoring criteria K coefficient (AHI) 0.3, P < 0.01 Bland–Altman plot showed that the ARES AHI was lower than the corresponding PSG AHI. ROC Desaturation ≥4% as scoring	Desaturation ≥ 4% as scoring criteria 100 Desaturation ≥3% as scoring criteria 100 Desaturation ≥1% + surrogate arousal criteria 98	Desaturation ≥ 4% as scoring criteria 27 Desaturation ≥3% as scoring criteria 35 Desaturation ≥1% + surrogate arousal criteria 56	Desaturation ≥ 4% as scoring criteria False negative AHI – 73% No false positive Desaturation ≥ 3% as scoring criteria False negative AHI 62.5% No false positive Desaturation 1% as scoring criteria False negative 44% False positive 37.5%

		Hypopnoea: 50% reduction in airflow with either 4% or 3% desaturation from the baseline Note: The ARES provider adopted a different severity grading for OSAS in which an ARES AHI of 0–5/h was labelled as normal, 6–20/h as mild OSAS, 21–40/h as moderate OSAS, 41–60/h as severe OSAS and over 61/h as very severe OSAS. The conventional grading of severity was adopted for reporting OSAS in the PSG: AHI 5–15/h = mild, 15–30/h = moderate and > 30/ h = severe.		criteria 0.96 Desaturation ≥3% + obstruction events 0.97 Desaturation ≥1% + surrogate arousal criteria 0.98			
Yagi et al. (2009) [48]	N=22 (5 female)	Apnoemonitor 51 (Chest Co., Tokyo, Japan) (6: oronasal thermistor, pulse oximeter, chest and abdominal belts, a microphone to monitor tracheal sounds, a position detector and the integrative unit) **Reference PSG: AASM 2012**		AHI r = 0.96 apnoea index r = 0.99 hypopnoea index			The apnoea index (AI) and hypopnoea index (HI) were significantly different between the two methods. The AI was greater with the PSG than Apnomonitor: 26.4 ± 23.0 (events per hour) for the PSG and 23.7 ± 22.1 for the Apnomonitor.

		Recommended Denominator: TRT Comparison: Simultaneous device – PSG measurement.		r = 0.86 obstructive ratio r = 0.94			There were no differences in any other sleep parameters (AHI, ODI, lowest SpO2) between the two methods.
Yuceege et al. (2013) [16]	N=85 (0 female) Highway bus drivers – shift workers	**Watch-PAT 200** (Itamar Medical, Caesarea, Israel) (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy) **Reference PSG: AASM 2007 Alternative** Dominator: TRT. Comparison: Simultaneous device – PSG measurement for 5 hours during daytime after full night of driving	**RDI>5** 96 (91-100/ 10 (0-28) **RDI>10** 89 (82-97)/ 53 (34-73) **RDI>15** 89 (80-98)/ 76 (63-90) Age <45 yrs 71/75 Age > 45	RDI r: 0.909 p < 0.0001 ODI r: 0.923 p < 0.0001 <90% desaturation r: 1,000 p < 0.0001 mean SpO2 r: 1,000 p < 0.0001	**RDI>5** 88 (82-95) **RDI>10** 81 (72-90) **RDI>15** 82 (71-92)	**RDI>5** 25 (0-67) **RDI>10** 70 (49-90) **RDI>15** 85 (74-97	Age was relevant for the correct performance of Watch PAT. Watch-PAT device is helpful in detecting SDB with RDI > 15 in highway bus drivers, especially in drivers older than 45 years, but has limited value in drivers younger than 45 years old who have less risk for OSA

			yrs 92/78				
Cho et al., 2017 [29]	n=149 (22 female) Suspected OSA patients	**ApneaLink plus** (Resmed) (3: nasal pressure, pulse oximetry, respiratory effort) **Reference: PSG - AASM 2012 Recommended** Dominator: TRT. Comparison: Simultaneous device – PSG measurement (also had home comparison) In lab PSG vs. (a) Automatic scoring, and (b) Modified automatic scoring, (c) Manual	**AHI≥5** a) 99.2/9.5 b) 93/85.7 c) 95.3/81 **AHI≥15** a) 96.6/37.7 b) 80.7/86.9 c) 92.0/93.4 **AHI≥30** a) 98.0/66.0 b) 77.6/99.0 c) 89.8/100	**ICC AHI:** a) 0.771 (95%CI 0.698-0.829) b) 0855 (95%CI 0.806-0.893) c) 0.939 (95%CI 0.917-0.956) **AUC:** **AHI≥5** a) 0.876 b) 0.919 c) 0.978 **AHI≥15**			Validation of ApneaLink Plus using automatic scoring, automatic scoring adjusted for AASM 2012 criteria, and manual scoring compared to simultaneous PSG. Reported that manual scoring was superior, followed by adjusted automatic scoring, then unadjusted automatic scoring. AHI difference was negatively associated with sleep efficiency and positively associated with arousal index, for

		Scoring		a) 0.931 b) 0.924 c) 0.973			manual and adjusted automatic scoring.
Massie et al., 2018 [8]	n=101 (44% female) Suspected OSAS patients	**NightOwl** (Ectosense, Leuven, Belgium) (4: peripheral arterial tone, pulse rate, pulse oximetry, actigraphy) **Reference PSG: AASM 2012 Recommended** Dominator: TST (derived from device). Comparison: Simultaneous device – PSG measurement	**AHI≥5** 98/80 **AHI≥15** 97/83 **AHI≥30** 90/97	**AHI:** **ICC = 0.86** **CC = 0.87 p < 0.001** **TST :** **ICC = 0.78** **CC = 0.78 p < 0.001**	**AHI≥5** 98 **AHI≥15** 89 **AHI≥30** 94	**AHI≥5** 80 **AHI≥15** 94 **AHI≥30** 94	Close agreement with PSG for estimation of REI and TST.

Device/Comparison column includes: Device name (Origin), Number and type of channels of device, reference scoring criteria that the device was compared to; the study time used in the denominator to calculate PM event frequency (e.g. TRT); type of comparison.

Table e25 Type III devices in comparison to PSG for investigating sleep-disordered breathing

Study	n (M/F)	Device (Channel n)	Sensitivity/ specificity (%)	Level of Device to PSG AHI Agreement	Positive Predictive Value (%)	Negative Predictive Value (%)	Comments
Westenberg et al. 2021 [167]	Neuromuscular disease patients Canada 38 (23:15) Age: 36.1 yrs (IQR 20.4-55.5) BMI 24.7 (IQR 21.1-30.2)	Alice PDX (4)	Overall 95/78 Home vs PSG AHI ≥5 79/100 AHI ≥ 15 50/88 AHI ≥ 30 38/94 Simultaneous AHI ≥ 5 70/100 AHI ≥ 10 64/100 AHI ≥ 30 31/100	Overall 90%	Home vs PSG AHI ≥5 100 AHI ≥15 92 AHI ≥30 67 Simultaneous AHI ≥5 100 AHI ≥10 100 AHI ≥30 100	Home vs PSG AHI ≥5 25 AHI ≥15 39 AHI ≥30 70 Simultaneous AHI ≥5 25 AHI ≥10 50 AHI ≥30 65	Failure rate of home Type II = 21% Patients given questionnaire for qualitative assessment Device overestimated events at lower cut-off values
Li et al. 2021 [168]	Chronic heart failure; Chinese and American populations 84 (73:11) Age:	Nox T3 (5)	Home vs. PSG AHI ≥ 5 87.6/76.5 AHI ≥ 10 95.6/83.6 AHI ≥ 30 71/97.8 Simultaneous	Overall Home vs. PSG 64.9% (κ = 0.54) Simultaneous 72% (κ = 0.62)	Home vs. PSG AHI ≥ 5 92.9 AHI ≥ 10 84.3 AHI ≥ 30 95.7 Simultaneous AHI ≥ 5 94.5 AHI ≥ 10 91.3	Home vs. PSG AHI ≥ 5 61.9 AHI ≥ 10 92.3 AHI ≥ 30 83.3 Simultaneous AHI ≥ 5 75 AHI ≥ 10 93.1	Manual editing of automatic traces 10.8% failure rate at home Two very different populations ethnically; data combined Device over and underestimated compared

	58.7±16.3 yrs BMI 29.4±13 Left ventricular ejection fraction 40.3±11.5 %		AHI ≥ 5 91/83 AHI ≥ 10 95.4/87.1 AHI ≥ 30 83.3/97.8		AHI ≥ 30 96.2	AHI ≥ 30 89.8	to manual scoring at low and high values
Driver et al. 2011 [44]	73 (30/43) Mean age ± SD: 53 ± 12 Mean BMI ± SD: 32.2 ± 6.8	MediByte PM (4)	**AHI=5** Sensitivity: 97 Specificity: 67 **AHI=10** Sensitivity: 84 Specificity: 91 **AHI=15** Sensitivity: 80 Specificity: 97 **AHI=20** Sensitivity: 80 Specificity: 95 **AHI=30** Sensitivity: 70 Specificity: 100	**Pearson correlation** r= 0.92	**AHI≥15** 97	**AHI≥15** 76	The mean difference between the RDI and the AHI showed an under-reporting with the device by −5.9±11.2 events/h. Failure rate 6.25 – 9%

Saletu et al. (2018) [53]	33 (19:14) Age: 62.7±5.3 yrs BMI: 28±3 4±3 months post-stroke Rehabilitation ward Austrian	SOMNOmedics (7)	Sensitivity of 94%	r = 0.97 (p< 0.001)	Not reported	Not reported	Part of HOPES study in stroke Type II study vs Type III study in-hospital 24% failure rate 92% of studies acceptable Direct comparison cited in 25 patients only Class severity of OSA changed in 43% of cases
To et al. 2009 [11]	175 (132/43) Mean age ± SD: Male: 47.8 ± 9.8 Female: 52.3 ± 12.2 Mean BMI ± SD: Male: 28.5 ± 4.9 Female: 29.2 ± 6.0 American population	ARES (10)	**Oxygen desaturations of 1-4% were enlisted as cut-off points:** **4% desat** Sensitivity: 84 (77–90) Specificity: 100 **3% desat** Sensitivity: 89 (84–94) Specificity: 100 **1% desat** Sensitivity: 0.97 (0.94–0.99)	**Kappa Coefficient:** **4% desat** 0.24, p<0.01 **3% desat** 0.3, p<0.01 **1% desat** 0.55, p<0.01	**4% desat** 100 **3% desat** 100 **1% desat** 98	**4% desat** 27 **3% desat** 35 **1% desat** 56	Device consistently underscored AHI compared to PSG. Some side-effects of wearing device reported by patients but not systematically examined

			Specificity: 0.63 (0.55–0.71)				
Nilius et al. 2017 [169]	85 (68/17) Mean age ± SD: 49.1 ± 13.5 Mean BMI ± SD: 29.7 ± 6.9 Mean ESS ± SD: 10 ± 5.1 Two centres Finnish population	Alice PDx (9)	**In-lab:** **AHI≥5** Sensitivity: 95 Specificity: 96 **AHI≥15** Sensitivity: 90 Specificity: 91 **AHI≥30** Sensitivity: 100 Specificity: 100 **At home:** **AHI≥5** Sensitivity: 77 Specificity: 76 **AHI≥15** Sensitivity: 87 Specificity: 85 **AHI≥30** Sensitivity: 90 Specificity: 91	**Correlation coefficient**: **In-lab device vs. PSG** ICC 0.95, p<0.001 **At home device vs. PSG** ICC 0.79, p<0.001	**In-lab:** **AHI≥5** 98 **AHI≥15** 85 **AHI≥30** 100 **At home:** **AHI≥5** 88 **AHI≥15** 77 **AHI≥30** 75	**In-lab:** **AHI≥5** 89 **AHI≥15** 94 **AHI≥30** 100 **At home:** **AHI≥5** 58 **AHI≥15** 92 **AHI≥30** 97	Device overestimated AHI value at lower AHIs and underestimated AHI value at higher AHIs. When performed in the lab and at home.

Pereira et al. 2013 [170]	128 (84/44) Mean age ± SD: 50 ± 12.3 Mean BMI ± SD: 31 ± 6.6 Canadian	MediByte (6)	**AHI≥5** Sensitivity: 87 Specificity: 67 **AHI≥10** Sensitivity: 79 Specificity: 86 **AHI≥15** Sensitivity: 77 Specificity: 95 **AHI≥30** Sensitivity: 50 Specificity: 93	**Positive correlation between device RDI and PSG AHI:** r = 0.69, r^2 = 48%	**AHI≥5** 96.2 **AHI≥10** 95.1 **AHI≥15** 97.1 **AHI≥30** 84.8	**AHI≥5** 34.8 **AHI≥10** 53.3 **AHI≥15** 65.5 **AHI≥30** 70.5	Device under-reported RDI compared to PSG AHI Scorer related concordance reported 9.5% device failure rate
Cairns et al. 2014 [39]	32 (18/14) Mean age ± SD: 46.8 ± 12.3 Mean BMI ± SD: 32.8 ± 6.8 American population	Nox T3 (8)	**AHI≥5** 100/70 **AHI≥15** 92/85	**Correlation between device and PSG AHI:** r(29) = .93, p < .001	**AHI≥5** 88 **AHI≥15** 79	**AHI≥5** 100 **AHI≥15** 94	Device non-significantly overestimated AHI compared to PSG. Scorer concordance reported.
Xu et al. 2017 [43]	80 (62/18) Mean age ± SD: 47.6 ± 14.0 Mean BMI ± SD: 27.5 ± 5.4 Mean ESS	Nox T3 (8)	**In lab:** **AHI≥5** 97/75 **AHI≥10** 96/100 **AHI≥15** 100/94 **AHI≥30** 97/98 **At home:**	**Squared correlation coefficient for AHI:** **PSG vs HSAT Identity Plot** r^2= 0.79 **PSG vs in-lab PM** r^2= 0.96	**In Lab:** **AHI≥5** 95 **AHI≥10** 100 **AHI≥15** 95 **AHI≥30** 97 **At home:** **AHI≥5**	**In lab:** **AHI≥5** 82 **AHI≥10** 92 **AHI≥15** 100 **AHI≥30** 98 **At home:** **AHI≥5**	Device underestimates AHI compared to PSG, with increasing disparity as AHI value increases. AHI in automatic scoring of the portable monitor recording had good agreement with manually edited scoring.

	± SD: 10.1 ± 4.9 Chinese population		**AHI≥5** 95/69 **AHI≥10** 92/79 **AHI≥15** 93/85 **AHI≥30** 63/93		94 **AHI≥10** 91 **AHI≥15** **89** **AHI≥30** 86	75 **AHI≥10** 83 **AHI≥15** 91 **AHI≥30** 80	Failure rate – 6.3%
Santos-Silva et al. 2009 [47]	80 (57/43) Mean age ± SD: 47 ± 14 Mean BMI ± SD: 28 ± 5 Brazil	Stardust II (5)	**AHI≥5** Sensitivity: 92 Specificity: 48 **AHI≥15** Sensitivity: 94 Specificity: 71 **AHI≥30** Sensitivity: 86 Specificity: 79	**PSG vs. In-lab device agreement:** 75%	**AHI≥5** 82 **AHI≥15** 76 **AHI≥30** 63	**AHI≥5** 71 **AHI≥15** 93 **AHI≥30** 93	In-lab device recording demonstrated 10% overestimation of AHI and 5% AHI underestimation compared to PSG. Scorer concordance reported 12% failure rate
Aurora, Patil and Punjabi 2018 [37]	53 (28/25) Mean age ± SD: 59.0 ± 12.9 Mean BMI ± SD: 37.1 ± 18.8 Hospitalised heart failure patients American	ApneaLink Plus (4)	**Overall** **AHI≥5** Sensitivity 95.8 Specificity 80.0 **Central** **AHI≥5** Sensitivity 90.9 Specificity 100 **Obstructive** **AHI≥5** Sensitivity	**Correlation coefficient for AHI:** Overall 0.94 Central 0.98 Obstructive 0.91 **Classification agreement:** Overall 88.7% Central 94.3% Obstructive 77.3%	**AHI≥5** Overall 97.9 Central 100 Obstructive 92.9	**AHI≥5** **Overall 66.7** **Central 93.9** **Obstructive 90.9**	Respiratory polygraphy underestimated overall AHI by 3.6 events/hr. High degree of agreement of both obstructive and central sleep apnoea. 3.5% failure rate

	population		97.5 Specificity 76.9				
Ayappa et al. 2008 [12]	97 (69/28) Mean age: Patients: 46 Volunteers: 36 Mean BMI: Patients: 30 Volunteers: 24 North American population	ARES Unicorder (5)	**In lab:** **AHI≥5** 98/84 **AHI≥10** 97/85 **AHI≥15** 92/95 **At home:** **AHI≥5** 90/79 **AHI≥10** 86/82 **AHI≥15** 74/88	**PSG vs ARES In-lab** **ICC = 0.95** **Lab PSG vs Home ARES** **ICC 0.76** **Lab ARES vs Home ARES** **AHI 4% in the lab higher than in the home, p=0.05** **AHI 1% in the lab higher than in the home, p=0.001**	**Not reported**	**Not reported**	"Large Increase" in disease probability based on a RDI of greater than 15 per hour and a "Large Reduction" in disease probability based on a RDI of less than 15 per hour. Home ARES failure rate 12%, PSG failure rate 1%
O'Brien et al. 2012 [22]	31 pregnant (0/31) Mean age ± SD: 30.2 ± 7.1 yrs Mean gestational age ± SD:	Watch-PAT 200 (4)	**AHI≥5** 88/87 **AHI 6.1 Threshold** 88/91 **AHI 4.9 Threshold** 88/91 **RDI 9.3**	**AHI Correlation** **r=0.76, p<0.001** **RDI Correlation** **r=0.68, p<0.001** **AHI≥5:** **k statistic =0.71, p<0.001** **Categories of AHI severity:** **k statistic =0.32,**	**AHI≥5** 70 **RDI≥10** 50	**AHI≥5** 95 **RDI≥10** 100	Autoscoring for WatchPAT Watch-PAT had a tendency to classify apnoea severity as slightly worse than that reflected on the PSG especially for RDI measurement.

	33.4±3.0 weeks Mean BMI ± SD: 31.9 ± 8.1 American		**Threshold** 100/82 **RDI≥10 Threshold** 100/81	**p=0.002** **Categorical agreement of RDI≥10: k statistic =0.42, p=0.013** **Categories of RDI severity:** **k statistic =0.08, p=0.37**			10% failure rate
Gan, Lim and Chong 2017 [20]	20 (18/2) Mean age ± SD: 39 ± 16 Mean BMI ± SD: 27.2 ± 5.5 Mean ESS ± SD: 9.55 ± 4.8 Asian population	Watch-Pat 200 (4)	**AHI>5** Sensitivity 100 Specificity 75.0 **AHI>15** Sensitivity 84.6 Specificity 100 **AHI>30** Sensitivity 80.0 Specificity 100	**ρ = 0.94** **p<0.001**	**Not Reported**	**Not Reported**	Tendency for the WatchPAT 200 to overscore the AHI at the mild range and to underscore the range at the severe end of OSA.

Cheliout-Heraut et al. 2011 [67]	90 (60/30) Mean age ± SD: 55.4 ± 8.7 Mean BMI ± SD: Mild OSA 26.7 ± 7.3 Moderate OSA 28.9 ± 5.3 Severe OSA 29.7 ± 4.1 French population	Somnolter (5)	**AHI Overall** Sensitivity 83.6 Specificity 81.8	**r=0.95** **p<0.001**	Not reported	Not reported	Underestimation of the AHI by the Somnolter compared to PSG. 13.5% failure rate of device
Polese et al. 2013 [34]	43 (19/24) Mean age ± SD: 70.0 ± 5.0 Mean BMI ± SD: 30.3 ± 6.0 Mean ESS ± SD: 9.1 ± 6.1 Elderly population Brazil	Stardust II (5)	**In Lab:** **AHI≥5** 100/0 **AHI≥15** 100/70 **AHI≥30** 90/68 **At home:** **AHI≥5** 90/30 **AHI≥15** 80/60 **AHI≥30** 80/80	**Home vs PSG** **r=0.67, p<0.001** **Simultaneous** **r=0.84, p<0.001**	**In Lab:** **AHI≥5** 1 **AHI≥15** 90 **AHI≥30** 71 **At Home:** **AHI≥5** 90 **AHI≥15** 88 **AHI≥30** 70	**In Lab:** **AHI≥5** 0 **AHI≥15** 0 **AHI≥30** 0.1 **At Home:** **AHI≥5** 60 **AHI≥15** 45 **AHI≥30** 15	Correlation, accuracy and agreement were greater when the recordings were made simultaneously. 10.5% loss of data Good specificity only for AHI values greater than 15 NPVs were 0 or very low at all threshold levels Partial loss of data in 44% of home studies with device and 31% in PSG

Pang et al. 2007 [23]	37 (12/25) Mean age ± SD: 50.1 ± 12.2 Mean BMI ± SD: 34.6 ± 5.2	WatchPat (4)	**AHI>5** 94/80 **AHI>15** 96/79 **AHI>35** 83/72	**r=0.9288** **p<0.0001**	**Not reported**	**Not reported**	
Oliveira et al. 2012 [33]	26 (13/13) Mean age ± SD: 62.8 ± 8.5 Mean BMI ± SD: 31.0 ± 5.6 Mean ESS ± SD: 10.5 ± 4.1 Chronic Pulmonary Obstructive disease Brazil	Stardust-STD (5)	**Not reported**	**AHI PSG vs AHI lab** **r = 0.61** **p <0.0001** **AHI PSG vs AHI Home** **r = 0.47** **p<0.007** **PM Lab vs PM home** **r = 0.47** **p=0.006**	**Not reported**	**Not reported**	41 volunteers had poor recording quality and were excluded from the analysis. 61% loss rate. Tendency to overestimate mild cases and underestimate severe cases due to SpO_2 and flow recording failures Study participants asked to provide feedback
Garg et al. 2014 [171]	75 (18/57) Age 44.7 ± 10.6 ys BMI: Not reported Urban	WatchPat 200 (4)	**At home:** **AHI≥5** Sensitivity 96 Specificity 43 **AHI≥10** Sensitivity 90 Specificity 69 **AHI≥15** Sensitivity 92 Specificity 77	**ICC for AHI:** **PSG and AHI PM 0.73** **AHI PSG and AHI LAB 0.79** **AHI PM and AHI LAB = 0.75** **Correlation between PSG vs home r=0.37, p=0.02.**	**AHI>5** 79 **AHI>10** 83 **AHI>15** 83	**AHI>5** 82 **AHI>10** 82 **AHI>15** 88	82% participants preferred home over in-laboratory testing. Failure rate at 5.3% Visual analogue score and questionnaire regarding sleep quality undertaken

	African-American Population; underserved population						
Masa et al. 2011 [172]	348 (263/85) Mean age ± SD: 48.7 ± 11.8 Mean BMI ± SD: 31.0 ± 6.6 Multicentre study Spain	BreastSC20 and Breast Medical AB (5)	**AHI≥5 from PSG** **AHI≥5** 96/57 **AHI≥10** 87/86 **AHI≥10 from PSG** **AHI≥5** 97/39 **AHI≥20** 71/90 **AHI≥15 from PSG** **AHI≥20** 94/60 **AHI≥25** 67/92				**All AUCs were statistical significant, p<0.001, expressing a high level of diagnostic accuracy.** **Home RPs produced lower AHIs than PSG.** **14% of home RPs repeated compared to 25% of PSGs** **5% failure rate overall**

Oliveira et al. 2015 [173]	32 (18/14) Mean age ± SD: 49.2 ± 10.9 Mean BMI ± SD: 40.8 ± 5.2 Morbidly obese patients (BMI≥ 35kg/m^2)	Stardust 2 (5) Airflow, respiratory effort, body position, heart rate, SpO_2	**In Lab:** **AHI 5-30** Sensitivity 40 Specificity 77 **AHI≥30** Sensitivity 89 Specificity 100 **At Home:** **AHI 5-30** Sensitivity 40 Specificity 81 **AHI≥30** Sensitivity 67 Specificity 100	**AHI_PSG and AHI_LAB** **r = 0.92, p=0.0001** **AHI_PSG and AHI_Home** **r = 0.84, p = 0.0001** **Diagnostic agreement:** **PSG vs Lab 87%** **PSG vs Home 65%**	**In lab:** **AHI 5-30** 25 **AHI≥30** 100 **At Home:** **AHI 5-30** 29 **AHI>30** 100	**In lab:** **AHI 5-30** 87 **AHI ≥30** 87 **At Home:** **AHI 5-30** 87 **AHI>30** 68	45% data loss Patient feedback requested and recorded **Underestimation:** PSG vs Lab 10% **PSG vs Home 32%** **Overestimation:** PSG vs Lab 3% **PSG vs Home 3%**
Smith et al. 2007 [174]	Chronic heart failure 20 (14/6) Mean age ± SD: 61 ± 10 Mean BMI ± SD: 29 ± 6 White Scottish population	Embletta (7+)	**Not reported**	Lab kappa coefficient 0.63, p <0.01 Home Kappa coefficient 0.27, p = 0.06 Correlation between mean A + H per hour in bed Lab vs PSG r = 0.92, p <0.01 and LE A +H per hour in bed r =0.94, p <0.01. Correlation between mean A + H home r = =0.54, p <0.01.	Home 83	Home 57	Technical and situational factors may be more marked in this patient population with poor sleep efficiency and should be taken into consideration in screening or diagnostic tools for sleep-disordered breathing in patients with CHF. **A+H per hour in bed used to report results** 20% problems with equipment but all data used

Jiang et al. 2018 [25]	35 (23/12) Mean age ± SD: 49 ± 13 Mean BMI ± SD: 26.6 ± 3.3 Chinese population	Airflow plus SpO_2 plus accelerometer (5)	**AHI≥5** 96.5/100 **AHI≥15** 100/88.9 **AHI≥30** 94.7/91.7	**5≤ AHI <15** **0.49** **15≤ AHI < 30** **0.53** **30≤ AHI** **0.74** **15≤ AHI** **0.79** **5≤ AHI** **0.85**	**AHI≥5** 100 **AHI≥15** 95.7 **AHI≥30** 94.7	**AHI≥5** 66.7 **AHI≥15** 100 **AHI≥30** 91.7	
Weimin et al. 2013 [24]	28 (21/7) Mean age ± SD: 47.45 ± 13.46 Mean BMI ± SD: 29.99 *± 5.74 Chinese and American population	WatchPat 200 (6)	**AHI threshold 5** 95.8/100 **AHI threshold 15** 93.7/91.7 **AHI threshold 30** 85.7/100	**AHI agreement: R=0.92, p<0.001** Overall agreement for PSG and WatchPAT **56.4** ± 9.0%			The WatchPat overestimated events in the lower range but underestimated events when AHI was high. 6.7% failure rate of WatchPat compared to 17.65% for PSG. WatchPAT cannot differentiate hypopnoeas from apnoeas accurately.

Cho and Kim 2017 [29]	149 (127/22) Age 43.0 ± 12.3 yrs BMI 26.0 ± 3.6 South Korean population	Apnea Link Plus (4)	**For Manual scoring only** Home vs PSG AHI ≥5 93/62 AHI ≥ 15 75/86.9 AHI ≥ 30 69.3/96 Simultaneous AHI ≥ 5 95/81 AHI ≥ 15 92/93 AHI ≥ 30 90/100	**PSG vs device** **ICC = 0.94** **P<0.001** **Home device vs PSG** **ICC = 0.082** **P<0.001** **Home device vs. lab device** **ICC = 0.84** **P< 0.001**	**Not reported**	**Not reported**	Manual scoring was superior to autoscoring; autoscoring not specific for respiratory events and sensitivity very high esp. at an AHI≥5 level. Failure rate at home = 6%
Gjevre et al. 2011 [175]	47 Canadian women Age 52.0 ± 11 yrs BMI 34.9 ± 9 68% post-menopausal	Embletta PDS (7)	Home vs PSG AHI ≥5 91/60 AHI ≥ 10 75/87 AHI ≥ 20 47/100 AHI ≥ 30 28/100	**Kappa statistic** AHI ≥5 0.54 AHI ≥ 10 0.56 AHI ≥ 20 0.36 AHI ≥ 30 0.2	AHI ≥5 83 AHI ≥ 10 92.3 AHI ≥ 20 100 AHI ≥ 30 100	AHI ≥5 75 AHI ≥ 10 62 AHI ≥ 20 47 AHI ≥ 30 40	

Ioachimescu et al. 2020 [176]	500 (80/20) 72% African American BMI mean – 31.6 Age not stated as mean	WatchPAT200 (4)	**For a diagnosis of no OSA 35/97** **Moderate-severe OSA 91/61**	**Concordance of accuracy overall = 53.4%** **Accuracy using 3% desats. = 68.9%** **Accuracy using 4% desats = 71%**	**For a diagnosis of no OSA 66** **For a diagnosis of moderate to severe OSA 76**	**For a diagnosis of no OSA 89** **For a diagnosis of moderate to severe OSA 83**	High rates of diagnostic misclassification in a very large sample using WatchPAT 5% failure rate Missestimation of sleep time significant. Pulse oximetry artefacts 3% desats overestimate OSA 4% desats underestimate OSA Suggested using 4% desats in the autoscoring programme to increase negative predictive value and specificity
Chang et al. 2019 [40]	Patients with COPD Chinese and American populations combined data 90 (80:10) Age 66.5 ± 7.8 yrs BMI 27.5 ± 5.8	NoxT3 (8)	Home vs PSG AHI ≥5 95/78 AHI ≥ 30 58/98 Simultaneous AHI ≥ 5 96/84 AHI ≥ 30 79/98	**AHI 4%** **Home vs PSG 67.8%** **K = 0.57** **Simultaneous 86.1%** **K = 0.80** **AHI 3%** **Home vs PSG 57.5%** **K = 0.43** **Simultaneous 79.1%** **K = 0.7**	Home vs PSG AHI ≥5 88 AHI ≥ 30 93 Simultaneous AHI ≥ 5 91 AHI ≥ 30 96	Home vs PSG AHI ≥5 89 AHI ≥ 30 86 Simultaneous AHI ≥ 5 93 AHI ≥ 30 98	More severe COPD in Chinese patients (p = 0.022); BMI significantly lower in Chinese compared to American patients (p = 0.009) 5.6% failure rate Post-study questionnaire to patients about experience Close agreement of automatic vs. manual scoring Lower mean SpO2 and percentage spent below SpO2 of 90% on home

							study
Ng et al. 2010 [26]	80 (63/17) Age 51.4 ± 11.9 yrs BMI 27.1 ± 4.2 Hong Kong Chinese	Embletta PDS (5)	AHI ≥5 92/86 AHI ≥ 10 90/87 AHI ≥ 15 88/95 AHI ≥ 20 85/96	**AHI overall** **r = 0.98** **p<0.05**	AHI ≥5 97 AHI ≥ 10 92 AHI ≥ 15 95 AHI ≥ 20 94	AHI ≥5 71 AHI ≥ 10 84 AHI ≥ 15 88 AHI ≥ 20 90	Autoscoring with manual scoring of PSG 11% failure rate of home device
Yuceege et al. 2013 [16]	Highway bus drivers in Turkey 85 Male only 23 were aged <45yrs 62 aged> 45 years No other demographic or anthropometric data available	WatchPAT200 (4)	RDI ≥5 96/100 RDI ≥ 10 89/65 RDI ≥ 15 89/76	**RDI overall** **r = 0.9** **p< 0.0001**	RDI ≥5 88 RDI ≥ 10 81 RDI ≥ 15 82	RDI ≥5 23 RDI ≥ 10 70 RDI ≥ 15 85	Manual PSG scoring vs WatchPAT autoscoring Excluded patients with vascular issues etc. Limited information **RDI – not defined as also used AHI**

Topor et al. 2020 [177]	27 (19/8) Age 54.7±7 yrs No other demographic or anthropometric data Canada	MATRx plus (4)	**Using 4% desats to mark events** **84/96.5**	**Moving average using 4% desats r = 0.95** **Moving average using 3% desats r = 0.93**	**84.4**	**95.5**	No comment on autoscoring override Conflicts of interest Limited information on participants 15.6% failure rate for device set-up at home
Jen et al. 2020 [178]	33 patients with COPD 33 (20/13) Canada	WatchPAT200 (4)	AHI ≥5 96/56 AHI ≥ 15 92/65 AHI ≥ 30 89/96	**r = 0.85** **p<0.001**	**Not reported**	**Not reported**	Automated scoring on device; manual scoring of PSG No patient outcomes discussed
Gupta et al. 2021 [179]	North Indian patients 35 (22/13) Age 48.6±10.7 yrs BMI for 35 patients not available separately	Stardust II Sleep Recorder (5)	AHI ≥5 94/25 AHI ≥ 20 94/77 AHI ≥ 30 93/91	**Home vs PSG** AHI ≥5 P = 0.4 p = 0.6 AHI ≥ 30 P = 0.89 p <0.0001	AHI ≥5 91 AHI ≥ 20 81 AHI ≥ 30 57	AHI ≥5 33 AHI ≥ 20 92 AHI ≥ 30 95	Failure rate 12% Patient questionnaire 30% of patients had concerns over device at home citing ease of use, diagnostic accuracy, and safety issues. (15 patients of 50 cited undertook PSG only)

Kasai et al. 2020 [180]	120 (102/18) Age 58±11.9 yrs BMI 26.4±5.4 Patients classified as with and without cardiovascular disease (CVD)	WatchPAT200 (4)	**Overall** **AHI ≥30** **88.5/74.6** **Patients with CVD** **AHI ≥30** **76.7/80** **Patients without CVD** **AHI ≥30** **87.1/82.4**	**Overall** **r = 0.896** **p <0.001** **Patients with CVD** **N = 55** **r = 0.849** **p <0.001** **Patients without CVD** **N= 65** **r = 0.927** **p<0.001**	**Not reported**	**Not reported**	Autoscoring of device; manual scoring of PSG Significant underscoring of SpO2 by device in patients with and without cardiovascular disease (p<0.001) as well as nadir SpO2 (p= 0.008).
Kinoshita et al. 2018 [21]	61 (48/13) Age 57.1±13.5 yrs BMI 26.5±4.4	WatchPAT (4)	**AHI ≥30** **79/80**	**r = 0.69** **p <0.0001**	**Not reported**	**Not reported**	Device underestimated AHI Arterial stiffness associated with aging and other co-morbidities resulted in underestimation of signal

Table e26. Technical differences across currently available type III devices that have been studied in paediatric populations.

Type III device	Nasal pressure cannula	Oronasal thermistor	Thoracic-abdominal inductance plethysmography belts	Pulse oximetry	Snoring by microphone	Body position	Actigraphy	Study
Alice PDx (Philips Respironics)		Yes	No	Yes		Yes		Chiner (2020) [181]
Alice 3, 3.5 or 6 (Philips Respironics)	Yes		Yes	Yes				Tabone (2019); Dudoignon (2017) [182, 183]
ApneaLink (Resmed, Germany)	Yes			Yes				Taddei (2015); Kasapkara (2014); Massicotte (2014) [184-186]
ApneaLink Air (Resmed)	Yes		Yes	Yes	Yes			Stöberl (2019) [187]
ApneaLink Plus (Resmed)	Yes		Yes	Yes				Modesti-Vedolin (2018); Lesser (2012) [188, 189]

Apnoescreen Pro (Erich Jaeger GmbH & CoKg)		Yes	Yes	Yes	Yes		Yes	Luna-Paredes (2012) [190]
Cidelec polysomnography (Angers, France)		Yes	Yes	Yes	Yes	Yes		Tabone (2019); Caggiano (2017): Dudoignon (2016) [182, 183, 191]
CID102L8 (Cidelec)	Yes		Yes	Yes	Tracheal sound			Giabicani (2019) [192, 193]
Edentec model 3711 (Edentrace II)		Yes	Yes	Yes	Yes	Yes		Alonso-Álvarez (2012) [194]
Embla S4500 System (Natus Medical)	Yes		Yes	Yes	Yes			Abel (2019); Pabary (2019) [195, 196]
Embletta Gold III (Embla)	Yes		Yes	Yes		Yes		Corbelli (2020); Michelet (2020); Brockmann (2013) Plomp

								(2012) Scalzitti (2017) [197-201]
eXim Apnea polygraph (BitMed, SIBEL Group)	Yes	Yes	Yes	Yes	Yes	Yes		Alonso-Alvarez (2015) [202]
Grass Technologies Aura PSG system, Astro-Med Inc.,	Yes	Yes	Yes	Yes	Yes		**Yes**	Luna-Paredes (2012) [190]
MediByte (Braebon Medical Corporation)	Yes		Yes	Yes	Yes	Yes		Masoud (2019) [203]
NoxT3 portable sleep monitor (ResMed)	Yes	Yes	Yes	Yes	Yes (built into box)	Yes (built into box)	Yes (built into box)	Vezina (2020); Gudnadottir (2019); Blanc (2019); Ikizoglu (2019); Orntoft (2019) Stöberl (2019); Jonson (2017) [187, 204-208]

Polysmith (Nihon Kohden America Inc)	Yes	Yes	Yes	Yes				Tan (2014) [209]
Porti 6		Yes (nasal airflow)	Respiratory effort by strain gauges	Yes				Lecka-Ambroziak (2017) [210]
Sleep Monitoring System c510 (Compumedics)		Yes	Yes	Yes				Caggiano (2017) [191]
Sleeptester (Fukuda Lifetch)	Yes		Yes	Yes	Yes	Yes		Hamada and Iida (2012) [211]
Smart Watch PMP-300E (Pacific Medico)	Yes		Yes	Yes	Yes	Yes	Yes	Kitamura (2014); Hamada (2015); Kitamura (2016) [212-214]
SNAP test (SNAP Diagnostics)	Yes		Yes	Yes	Acoustical analysis of snoring sound (from oronasal airflow cannula)			Brietzke (2015) [215]
SOMNOscreen System (SOMNOmedics)	Yes	Yes	Yes	Yes	Yes (pressure sensor)			Joyce (2020); Abel (2019) [195, 216]

SOMNOtouch (Somnomedics)	Yes		Yes	Yes	Yes (nasal pressure with integral snore sensor)	Yes	Yes	Joyce (2020); Kingshott (2019); Joyce (2017); Hill (2016) [166, 216-218]
Unspecified device	Yes	Yes	Yes	Yes	Yes	Yes		Paoloni (2018); Pavone (2015) [219, 220]

Table e27. Summary of studies comparing type III devices (respiratory polygraphy-PG) to full attended polysomnography (PSG).

Study	Number (male/female)	Device (Channel number)	Sensitivity/ Specificity (%)	Level of type III device to PSG AHI Agreement	Positive Predictive Value (%)	Negative Predictive Value (%)	AHI Correlation: Comments
Ikizoglu et al. (2019) [204]	19 (7/12) Median age 11.3 years (6-18 years old); **Down syndrome** **Home PG** 2/19: home PG without airflow and had to be repeated; 12/19 (63.2%) with mild OSAS (AHI 1-5/h); 4/19 (21.1%) moderate/severe OSAS (AHI>5/h) **PSG** 2/19 (11%) with mild OSAS (AHI 1-5/h); 4/19 (21.1%) moderate/ severe OSAS (AHI>5/h)	**PSG** vs. **home PG (2012 AASM scoring rules):** NoxT3 portable sleep monitor; (ResMed); **(channels n=8: nasal flow canula, chest and abdominal wall movements, EKG, oximetry, body position, snoring)**	**AHI≥1/h** Sensitivity: 100% Specificity: 30% **AHI≥3/h** Sensitivity: 100% Specificity: 85% **AHI≥4.3/h** Sensitivity: 83% Specificity: 100%	**PSG vs home PG:** AHI r=0.642; P=0.003; ODI r=0.629; P=0.04	**AHI≥1/h** 38%	**AHI≥1/h** 100%	**ROC analysis:** for PSG AHI ≥1/h the best cut-off in home PG was AHI ≥3/h with sensitivity 100% and specificity 85% Success rate for home PG was 89% in the first night **Home PG overestimated AHI compared to PSG**

Masoud et al. (2019) [203]	70 (38/32) Median age 10.8 (8.6-14.3) yrs Median BMI 25.1 (20.8-31.6) kg/m²	**PSG** vs **in-laboratory PG** performed simultaneously **PSG (2012 AASM scoring rules):** Alice 5 (Philips Respironics, Pennsylvania, USA) **PG** MediByte (Braebon Medical Corporation, Canada) (channels **n= 7: respiratory effort by thoracic-abdominal belts; nasal pressure transducer; pulse oximetry; heart rate; snoring by microphone; body position)**	**Automatic analysis** **AHI ≥1.5/h** Sensitivity: 100% Specificity: 21.7% **AHI ≥5/h** Sensitivity: 95.5% Specificity: 66.7% **AHI≥10/h** Sensitivity: 100% Specificity: 93.4% **Automatic analysis with exclusion of bad data** **AHI ≥1.5/h** Sensitivity: 97.9% Specificity: 21.7% **AHI ≥5/h** Sensitivity: 90.9% Specificity: 70.8% **AHI ≥10/h** Sensitivity:	**Automatic analysis** AHI correlation r=0.932; ODI correlation r= 0.854 **Automatic analysis with exclusion of bad data** AHI correlation r=0.935; ODI correlation r= 0.881 **Manual analysis** AHI correlation r=0.939; ODI correlation r= 0.904 All p≤0.05	**Automatic analysis** **AHI ≥1.5/h** 72.3% **AHI ≥5/h** 56.8% **AHI ≥10/h** 69.2% **Automatic analysis with exclusion of bad data** **AHI ≥1.5/h** 71.9% **AHI ≥5/h** 58.8% **AHI ≥10/h** 69.2% **Manual analysis** **AHI ≥1.5/h** 87% **AHI ≥5/h** 88.2% **AHI≥10/h** 90%	**Automatic analysis** **AHI≥1.5/h** 100% **AHI≥5/h** 97% **AHI≥10/h** 100% **Automatic analysis with exclusion of bad data** **AHI≥1.5/h** 83.3% **AHI≥5/h** 94.4% **AHI≥10/h** 100% **Manual analysis** **AHI ≥1.5/h** 70.8% **AHI ≥5/h** 86.8% **AHI ≥10/h** 100%	**ROC analysis:** PG performed best when diagnosing severe OSAS (AHI ≥10/h) with all 3 scoring methods. Automated scoring resulted in a higher sensitivity when using a lower cut-off for AHI (e.g. AHI ≥1.5 or ≥ 5).

			100% Specificity: 93.4% **Manual analysis** **AHI≥1.5/h** Sensitivity: 85.1% Specificity: 73.9% **AHI≥5/h** Sensitivity: 68.2% Specificity: 95.8% **AHI≥10/h** Sensitivity: 98.4% Specificity: 90%				
Scalzitti et al. (2017) [201]	33 (<18 years) with symptoms and signs of OSAS	**PSG (2007 AASM scoring rules):** Sandman, Natus Medical Inc, USA) with Sentec tcCO2 (Therwil, Switzerland) **PG (2007 AASM scoring**	**AHI ≥1/h** Sensitivity: In-lab 81.5% Home 1 69.2% Home 2 70% Specificity In lab 60% Home 1 42.9% Home 2				For Home 2 sensitivity could be increased to 90% using AHI ≥0.75/h, but specificity decreased to 63%. Age <5 years had a greater error for AHI; male patients had less error in AHI.

		rules): Embletta Gold (n=6: nasal pressure, thermistor, thoracic and abdominal effort, oximetry, ECG). Patients underwent in-laboratory PSG and polygraphy and one or two nights polygraphy studies.	83.3%				
Alonso-Alvarez et al. (2015) [202]	50 (27/23) Age 5.3 ± 2.5 yrs OSAS definition: PSG-ORDI ≥3/h <u>PSG</u> ORDI ≥3/h: 66% ORDI ≥5/h: 54% OAHI ≥3/h: 52% OAHI≥5/h: 44%	**PSG: (2007 AASM scoring rules):** DeltaMed Coherence 3NT PSG (Diagniscan S.A.U. Group Werfen) **Home and in-laboratory PG (2007 AASM scoring rules)**: eXim Apnea polygraph	For **a PSG ORDI ≥3/h** the best cut-off value <u>in lab PG</u> was **ORDI ≥4.6/h** with sensitivity 100% and specificity 88.4% For a **PSG ORDI ≥3/h** the best cut-off value <u>in home PG</u> was **ORDI ≥5.6/h** with sensitivity	Intra-class correlation coefficient: <u>Lab PG</u> RDI ≥3/h: 96 (91.8-97.9) ORDI ≥3/h: 96.5 (92.3-98.3) OAHI ≥3/h: 95.8 (92.6-97.6) <u>Home PG</u> RDI ≥3/h: 85.9 (75.2-92) ORDI ≥3/h: 86.7 (76.5-92.5) OAHI ≥3/h: 84.3 (72.5-91.1)	For **a PSG ORDI ≥3/h** the best cut-off value <u>in lab PG</u> was **ORDI ≥4.6/h** with PPV 94.3% For a **PSG ORDI ≥3/h** the best cut-off value <u>in home PG</u> was **ORDI ≥5.6/h** with PPV 96.8% For **a PSG OAHI ≥3/h** the	For **a PSG ORDI ≥3/h** the best cut-off value <u>in lab PG</u> was **ORDI ≥4.6/h** with NPV 100% For a **PSG ORDI ≥3/h** the best cut-off value <u>in home PG</u> was **ORDI ≥5.6/h** with PPV 84.2% For **a PSG OAHI ≥3/h** the	The agreement for lab PG and home PG was >80%. Higher for lab PG than home PG. Home PG valid alternative for PSG For **a PSG ORDI ≥3/h** the best cut-off value <u>in lab PG</u> was **ORDI ≥4.6/h** For a **PSG ORDI ≥3/h** the best cut-off value <u>in home PG</u> was **ORDI ≥5.6/h** For **a PSG OAHI ≥1/h** the best cut-off value in home

		(BitMed, SIBEL group)– automated and manual scoring **(n=7: oronasal airflow by pressure transducer and nasal pressure, thoracic and abdominal inductive plethysmo-graphy belts, body position, microphone for snoring, pulse oximetry)** Flow limitation events were also scored. AHI, OAHI and RDI were calculated. RDI=central apnoeas+ obstructive apnoeas+ hypopnoeas+RERAs+ flow-limitation events/h. In PG RDI did not include RERAs. ORDI did not include central apnoeas	90.9% and specificity 94.1% For **a PSG OAHI ≥3/h** the best cut-off value in lab PG was **OAHI ≥4.6/h** with sensitivity 96.1% and specificity 83.3% For a **PSG OAHI ≥3/h** the best cut-off value in home PG was **OAHI ≥4/h** with sensitivity 92.3% and specificity 83.3% For **a PSG OAHI ≥1/h** the best cut-off value in home PG was **OAHI ≥3/h** with sensitivity 72.5% and specificity 90% For a **PSG OAHI ≥5/h** the best cut-off		best cut-off value in lab PG was **OAHI ≥4.6/h** with PPV 86.2% For a **PSG OAHI ≥3/h** the best cut-off value in home PG was **OAHI ≥4/h** with PPV 85.7% For **a PSG OAHI ≥1/h** the best cut-off value in home PG was **OAHI ≥3/h** with PPV 96.7% For a **PSG OAHI ≥5/h** the best cut-off value in home PG was **OAHI ≥6.7/h** with PPV 90%	best cut-off value in lab PG was **OAHI ≥4.6/h** with NPV 95.2% For a **PSG OAHI ≥3/h** the best cut-off value in home PG was **OAHI ≥4/h** with NPV 90.9% For **a PSG OAHI ≥1/h** the best cut-off value home PG was **OAHI ≥3/h** with NPV 45% For a **PSG OAHI ≥5/h** the best cut-off value in home PG was **OAHI ≥6.7/h** with NPV 86.7%	PG was **OAHI ≥3/h** For a **PSG OAHI ≥5/h** the best cut-off value in home PG was **OAHI ≥6.7/h** **Differences between PSG and home PG are greater for higher RDI**

			value in home PG was **OAHI ≥6.7/h** with sensitivity 81.8% and specificity 92.9%				
Tan et al. (2014) [209]	100 (59/41) Age 2-16 years No comorbidities	**PSG system used with** concealment of EEG channels for PG **(2007 AASM scoring rules)** Polysmith (Nihon Kohden America Inc, CA, USA)	**AHI≥1/h** Sensitivity: 82.5% (95% CI: 72.4-90.1) Specificity: 90% (95% CI: 68.2-98.9) **AHI≥3/h** Sensitivity: 70.4% (95% CI: 56.4-82) Specificity: 100% (95% CI: 92.3-100) **AHI≥5/h** Sensitivity: 62.5% (95% CI: 45.8-77.3%), Specificity: 100% (95% CI: 94-100%)	AHI correlation r=0.91 **Significant underestimation of total hypopnoea index if AHI≥1/h**	**AHI≥1/h** 97.1% (95% CI: 89.8-99.6) **AHI≥3/h** 100% (95% CI: 90.7-100) **AHI≥5/h** 100% (95% CI: 86.3-100%)	**AHI≥1/h** 56.3% (95% CI: 37.7-72.6) **AHI≥3/h** 74.2% (95% CI: 61.5-84.5%) **AHI≥5/h** 80% (95% CI: 69.2-88.4%)	**PG significantly underestimated AHI due to missed hypopnoeas accompanied by arousals without desaturation if AHI≥1** AHI≥1/h AUC 0.86 (95% CI: 0.78-0.94) AHI≥3/h AUC 0.85 (95% CI: 0.79-0.91) AHI≥5/h AUC 0.81 (95% CI: 0.74-0.89). **Change in classification of severity in 28% of patients.** **AHI <1/h:** PG no difference **AHI >=1 and <5/h:** PG 32.5% AHI<1 **AHI >=5 and <10/h:**

							PG 5% AHI<1/h and 70% AHI >=1 and <5/h **AHI ≥10/h:** PG 35% AHI >=5/h and <10/h Change of clinical decisions in 23% of patients.
Massicotte et al. (2014) [186]	35 (22/13) Median age – 11 years Comorbidities: obesity (10), sickle cell anaemia (13); CNS tumour (3), Chiari (2)	**PSG:** XLTEK, (Natus Medical Inc, USA) – **2007 AASM** scoring rules **PG:** ApneaLink (ResMed, USA) – automated scoring and manual scoring: apnoea <=20% baseline airflow; hypopnoea <=50% baseline airflow +3% desaturation **(n=2: nasal airflow by pressure transducer, pulse**	**AHI≥1.5/h (manual PG) & AHI≥1.5/h (PSG)** Sensitivity: 94% Specificity: 16% **AHI≥5/h (manual PG) & AHI≥5/h (PSG)** Sensitivity: 100% Specificity: 40% **AHI≥5/h (manual PG) & AHI≥1.5/h (PSG)** Sensitivity: 94% Specificity: 61%	**AHI correlation** **PSG vs manual PG:** Pearson correlation r=0.89; P<0.001; intraclass correlation r=0.81 **manual PG vs automated PG** Pearson correlation r=0.59; P=0.002; intraclass correlation r=0.57 **PSG vs automated PG** Pearson correlation r=0.36; P=0.03	**AHI≥1.5/h (manual PG) & AHI≥1.5/h (PSG)** 51% **AHI≥5/h (manual PG) & AHI≥5/h (PSG)** 22% **AHI≥5/h (manual PG) & AHI≥1.5/h (PSG)** 70%	**AHI≥1.5/h (manual PG) & AHI≥1.5/h (PSG)** 75% **AHI≥5/h (manual PG) & AHI≥5/h (PSG)** 100% **AHI≥5/h (manual PG) & AHI≥1.5/h (PSG)** 92%	The device cannot distinguish central/obstructive apnoeas (no thoracic/abdominal belts) **Manually scored PG overestimates AHI and overdiagnoses OSAS and its severity** (false positives due to wakefulness, mouth-breathing) Best PPV and NPV for PSG AHI >= 1.5/h with manual PG and AHI >=5/h

		oximetry)					

Lesser et al. (2012) [189]	25 (15/10) Mean age 13.6 ± 3.0 yrs; Obese 7/25 (28%) with moderate/severe OSAS (OAHI>5/h)	**PSG (2007 AASM scoring rules):** Sandman Elite Sleep Diagnostic Software System (Natus, USA) **PG (simultaneously with PSG:** Apnea Link Plus (ResMed, USA) – automated and manual scoring **(n=4: nasal airflow by pressure transducer, thoracic belt for chest movements and EEG, pulse oximetry)** <u>Definitions-automatic scoring</u> obstructive apnoea: 80-100% reduction in airflow with respiratory effort for >=10	**OAHI>1.5/h** (manual/automatic) Sensitivity: 100%/100% Specificity: 46.2%/46.2% **OAHI>5/h** (manual/automatic) Sensitivity: 85.7%/85.7% Specificity: 83.3%/83.3% **OAHI>10/h** (manual/automatic) Sensitivity: 80%/100% Specificity: 100%/90%	PSG vs. automated PG Spearman Rho = 0.886 ($p<0.001$)	**OAHI>1.5/h** (manual/automatic) 63%/63% **OAHI>5/h** (manual/automatic) 67%/67% **OAHI>10/h** (manual/automatic) 100%/71%	**OAHI>1.5/h** (manual/automatic) 100%/100% **OAHI>5/h** (manual/automatic) 94%/94% **OAHI>10/h** (manual/automatic) 95%/100%	Apnea Link Plus may overestimate OAHI (includes events during periods of wakefulness, events not associated with arousal or desaturation, post-arousal decreases in airflow) Manual scoring (paediatric criteria) compared to automatic scoring (adult criteria) did not improve sensitivity/specificity for OSAS diagnosis. 35% ± 25% of total Apnea Link recording time was not used for sleep scoring because of poor airflow signal

		sec; obstructive hypopnoea: 50-80% reduction in airflow with respiratory effort for >=10 sec <u>Definitions-manual scoring:</u> obstructive apnoea: 80-100% reduction in airflow with respiratory effort for >=2 respiratory cycles; obstructive hypopnoea: 50-80% reduction in airflow with respiratory effort for ≥2 respiratory cycles					

AASM = American Academy of Sleep Medicine; AHI = Apnoea-hypopnoea index; EEG = Electroencephalogram; OAHI = Obstructive Apnoea-Hypopnoea Index; ORDI = Obstructive Respiratory Disturbance Index; OSAS = Obstructive Sleep Apnoea Syndrome; PG = Polygraphy; PSG = Polysomnography; ROC = Receiver Operating Characteristic

Table e28. Summary of studies involving type III devices (respiratory polygraphy or home respiratory polygraphy) in paediatric populations without comparison to full attended polysomnography.

Study	Number (male/female)	Device	Channels	Unattended vs. attended PG	Manual/automated scoring	Scoring criteria	Comments
Chiner et al. (2020) [181]	121 (70/51) Age 7±4 years Children with suspected OSAS	Alice PDx (Philips)	Airflow (nasal cannula); respiratory movement with chest band; pulse oximetry; body position	Unattended	Manual	Not defined.	OSAS was defined as an AHI ≥3 events/h. 76% of patients were diagnosed based on HRP alone, indicating that HRP is usually sufficient for diagnosing OSAS in children.
Corbelli et al (2020) [199]	289 interpretable PG studies were included 6 groups with indication for OSAS included: craniofacial malformation,	Embletta Gold III (Embla)	Not specified	Both attended and unattended	Manual	**2012 AASM scoring rules**	

	neuromuscular disease, obesity, suspected OSAS, prematurity, other						
Joyce et al (2020) [216]	Parents of 80 (50/30) children aged 36 to 71 months (mean = 56.90, SD = 10.19 months) completed the Behaviour Rating Inventory of Executive Function - Preschool Version (BRIEF-P). Of 80 children, 69 were successfully studied overnight with domiciliary cardiorespiratory polygraphy to diagnose OSAS	Somnotouch (Somnomedics)	Abdominal and thoracic effort, nasal air flow, snore sensor, pulse oximetry, actigraphy and body position monitoring	Home and hospital	Manual	**2012 AASM guidelines**	Studies were scored in 30 second epochs for sleep and wake, using a combination of parental sleep log to interpret sleep onset and offset; the in-device X, Y, Z accelerometer; changes in variables such as heart rate and respiratory rate; and what had occurred in surrounding epochs. Epochs with at least 15

							seconds of increased activity were classified as estimated wake.
Michelet et al. (2020) [198]	400 polygraphies (in-hospital or at-home) in 332 (202/130) patients (age 7±4.7 years) with: suspected OSAS, obesity, craniofacial malformation, neuromuscular disease, history of prematurity, achondroplasia, Arnold-Chairi malformation, Prader-Willi syndrome	Embletta Gold III (Embla)	Nasal airflow (nasal pressure transducer); respiratory inductance plethysmography (RIP) chest and abdominal belts; RIP flow; pulse oximetry; snoring; transcutaneous CO_2 partial pressure in subset of patients.	Both attended and unattended	Manual	**2012 AASM scoring rules**	87% of polygraphies (in-hospital or at-home) were interpretable. Main reasons of non-interpretability: poor SpO_2 signal (80%); poor nasal canula signal (41%); poor abdominal belt signal (29%); poor thoracic belt signal (18%).
Vezina et al. (2020) [208]	Canadian Healthy Infant Longitudinal Development Study 562 participants	Nox T3 respiratory polygraphy (Nox Medical Inc. Reykjavik, Iceland)	Airflow (nasal cannula connected to pressure transducer; oronasal thermistor);	Unattended at home	Manual	**2007 AASM scoring rules**	91% of recordings were technically acceptable; 90% of unacceptabl

	(50.4% males/49.6% females) Age 1.1±0.2 yrs Excluded: Infants with major congenital abnormalities or born after pregnancy of less than 35 weeks; infants of multiple births		thoracic and abdominal inductance plethysmography belts for respiratory effort; actigraphy; body position; pulse oximetry.				e studies were due to displaced or disconnected oximetry probes. 99.5% of infants had an obstructive apnoea index <2/h; **Obstructive apnoea index:** median 0/h; 10th percentile 0/h; 90th percentile 0.5/h. **Central apnoea index:** median 2.5/h; 10th percentile 0.6/h; 90th percentile 7.1/h. **AHI:** median 4.2/h; 10th percentile 1.2/h; 90th percentile

							15.8/h. **Oxygen desaturation (≥3%) index:** median 6.7/h; 10th percentile 1.4/h; 90th percentile 10.7/h. **Oxygen saturation of haemoglobin:** median 97%; 10th percentile 95.4%; 90th percentile 97.9%.
Abel et al. (2019) [195]	48 (20/28) children with Prader Willi syndrome, age 2.3 yrs (0.2-14.1); 92 (53/39) controls, female, age 2.2 yrs (0.3-15.1).	Embla S4500 System (Natus Medical Inc., CA, USA) SOMNOscreen™ System (SOMNOmedics GmbH, Germany)	Centre 1 Nasal pressure transducer; respiratory inductance plethysmography; pulse oximetry (Nonin Medical BV; Amsterdam, The Netherlands);	Attended in the hospital	Manual	**2012 AASM scoring rules**	OAHI >1 was used for OSAS diagnosis

			snoring (microphone); electrocardio-gram; video/sound recording. Simultaneous transcutaneous carbon dioxide partial pressure ($PtcCO_2$) tracing was obtained using a TCM TOSCA 500 transcutaneous monitor (Radiometer Medical ApS, Denmark). Centre 2 Airflow (thermistor and nasal pressure transducer); chest and abdominal wall movements (respiratory inductance plethysmogra-phy); pulse oximetry (Nonin Medical				

			BV), snoring (pressure sensor); electrocardio-gram; video recording. Transcutaneous CO_2 tracing was recorded using a TCM40 transcutaneous monitor (Radiometer Medical ApS, Denmark).				
Blanc et al. (2019) [205]	50 (32/18) Age 5.5 ± 2.3 years Polymalformative syndrome (6 pts); Down syndrome (3 pts); Overweight/obesity (3 pts); neurologic disorders (4 pts)	Nox T3 respiratory polygraphy (Nox Medical Inc. Reykjavik, Iceland)	Finger sensor to record heart rate; SpO2 and pulse rate; thoracic and abdominal inductance plethysmography belts to measure respiratory effort; nasal cannula to assess airflow; actigraphy; body position; and snoring sound recording.	Unattended in the hospital, nurse checked every 3 hours	Compared automatic to manual scoring	**2012 AASM scoring rules**	Well-accepted: 98% Average signal quality: 70.8% (86% in pts >3 yrs; 25% in pts <3 yrs) OSAHS was defined by obstructive apnoea index (OAI) > 1/h or obstructive apnoea/

							hypopnoea index (OAHI) > 1.5/h; 32% pts diagnosed with OSAHS based on manual scoring; 100% pts diagnosed with OSAHS based on automatic analysis
Giabicani et al. (2019) [192]	40 (50% male). mean age at first polygraphy was 6.0 years (1.7–15.3) Silver – Russell syndrome 45 polygraphies and 16 PSGs were performed. 21 sleep recordings in 20 patients without rGH therapy and sleep recording	CID102L8 (Cidelec)	Nasal pressure, thoraco-abdominal movement, tracheal sound, pulse oximetry, transcutaneous carbon dioxide pressure (Radiometer TINA). In later studies, EEG was added.	Attended	Not specified	**2012 AASM scoring rules**	SDB was present in 86.4% of patients before rGH therapy and was severe in 13.6%. AHI worsened for 5 of 12 patients with sleep recordings before and after rGH therapy initiation,

	before and during rGH therapy in 12 patients.						reaching mild impairment.
Gudnadottir et al (2019) [207]	60 children Age 4-10 years	Nox T3 respiratory polygraphy (Nox Medical Inc. Reykjavik, Iceland)	Respiratory flow via a nasal cannula pressure transducer, thoracic and abdominal respiratory effort with calibrated RIP measurements, SpO2 via wireless pulse oximetry (Nonin® 3150 Wristox2 Bluetooth Wrist Pulse Oximeter), body position, actigraphy and audio recording.	Unattended at home	Manual scoring	**2015 AASM scoring rules**	The requirement of 3 hours of valid data for an acceptable RP was not fulfilled for nasal airflow in 40% and for SpO2 in 19%, and both parameters were missing in 11%.
Kingshott et al. (2019) [166]	194 research participants with Down syndrome (0.5 to 5.9 years old) and 61 typically	SOMNOtouch device (SOMNO-medics, Germany)	Chest and abdominal respiratory inductance plethysmography; pulse	Unattended at home	Manual scoring	**2012 AASM scoring rules**	74% (95% CI 67% to 79%) of research participants and 82% (95% CI 71%

	developing children with suspected sleep disorders and children with comorbidities (clinical cohort) (0.4 to 19.5 years)		oximetry (Bluepoint); plethysmography; nasal pressure flow with integral snore sensor; body position sensor and actigraphy Clinical cohort had SpO2 (Massimo, USA) and PtcCO2 (SenTec, Switzerland) and subgroup had video-monitoring				to 90%) of clinical cohort had successful home PG at the first attempt.
Ørntoft et al. (2019) [193]	51 (23/28); median age – 13.6 yrs; median BMI – 99.6th percentile. Overweight and obese patients entered in a weight loss program. Patients included during follow-up visit 6-	Nox T3 analysed using manual & Automated scoring	Nasal airflow, thoracic and abdominal inductance plethysmongra-phy, body position, activity (integrated accelerometer), pulse oximetry	Home study (Not specified if attended or not)	Manual & Automated	**AASM scoring rules version 2.1**	Central apnoeas were excluded from the AHI OSA severity: Mild defined as ≥ 2 AHI

	8 months after initial OSAS diagnosis.						
Pabary et al. (2019) [196]	561 children: 104 with neuromuscular disorders; 58 with craniofacial disorders. 35 with T21; 32 with a metabolic condition; 20 with Hb; 157 patients with uncommon conditions. Median age 7 yrs (2-19 yrs)	Embla S4500 (Stowood Scientific)	Nasal pressure transducer, thoracic and abdominal excursion via respiratory inductance plethysmography, 2-lead EKG, pulse oximetry transcutaneous pCO_2 and infrared video recording with microphone.	Attended	Manual	**AASM 2012**	
Stöberl et al. (2019) [187]	24 (8/16) Children with Ehlers-Danlos syndrome and median age 14.2 (10.7–15.3) years; 24 (8/16) control children with median age 13.9 (10.7–15.9) years.	ApneaLink Air (ResMed, Germany) Or Nox T3 respiratory polygraphy (Nox Medical Inc. Reykjavik, Iceland)	Not reported.	Unattended at home	Manual scoring	**2012 AASM scoring rules**	OSAS was more prevalent in children with Ehlers-Danlos syndrome than in controls: 42% vs. 13% and OR of 4.5 (95% CI = 0.97–20.83, p = 0.054).

Tabone et al. (2019) [182]	7 (2/5) Age: median 1.8 years (range 0.3–17.4 years) Children with Mucolipidosis	Cidelec (St. Gemme sur Loire, France) Alice 6 (Philips Respironics, St. Priest, France)	Nasal flow (nasal pressure transducer); pulse oximetry; thoracic and abdominal respiratory inductance plethysmography; synchronized infrared video monitoring; transcutaneous carbon dioxide pressure	Attended in the hospital	Manual	**2012 AASM scoring rules**	OSAS defined as OAHI ≥2/h. Moderate OSAS in one patient (AHI 5/h) and severe OSAS in five patients (AHI ranging from 23 to 52 events/h)
Modesti-Vedolin et al. (2018) [188]	18 (10/8) Age: 8.3 ± 2.3 years Patients with a clinical history of snoring on the waiting list for tonsillectomy.	ApneaLink Plus (ResMed, Germany)	Nasal respiratory pressure signal; pulse oximetry	Unattended at home	Manual	**2012 AASM scoring rules**	A portable EMG device (BiteStrip) was used in combination with the ApneaLink
Paoloni et al. (2018) [219]	20 children with Marfan syndrome (11/9)	Not specified	Oronasal flow by oronasal thermistor nasal cannula to assess	Unattended at home	Not specified	**2007 AASM scoring rules**	80% of Marfan children had an AHI>1/h and were

			pressure; chest and abdominal movements by impedance plethysmography; body position; snoring by microphone; pulse oximetry				diagnosed with OSAS
Lee et al. (2018) [221]	Meta-analysis: 18 studies published between 1997 and 2016; two PG studies, one from 2003 and one from 2016 (the latter included in this table).					Not defined	
Lecka-Ambroziak et al. (2017) [210]	36 pts with Prader Willi syndrome	Porti 6	Nasal flow; respiratory effort by thoracic and abdominal strain gauges; pulse oximetry.	Not specified	Manual scoring	**AASM rules (no reference, to year)**	Normal value for AHI <1/h
Caggiano et al. (2017) [191]	5 (1/4) infants (mean age: 11.0 ± 9.8 months) with	Sleep monitor system C510 (Compumedics)	Nasal flow, respiratory inductance plethysmograp	Attended	Not specified	Not defined	Patients evaluated with PG and PSG; results

	congenital myasthenic syndrome; patients evaluated with PG and PSG; results are not specified based on the method.	Cidelec polysomnography (Angers, France)	hy, tracheal sound, body position, pulse oximetry, videotape recording				are not specified based on the method.
Dudoignon (2017) [183]	57 Mean age 6.2 ± 5.9 years Down syndrome 58% - associated disorders	Cidelec (St. Gemme sur Loire, France) Alice 6 (Philips Respironics, St. Priest, France) SenTec (Thurnwill, Switzerland) – for PtcCO2	Nasal flow (nasal pressure transducer); pulse oximetry; thoracic and abdominal respiratory inductance plethysmography; synchronized infrared video monitoring; transcutaneous carbon dioxide pressure	Attended		**2012 AASM scoring rules**	
Griffon et al. (2017) [222]	13 (9/4) Age 6.8 ± 7.7 months (range 1-24 months). Children	Cidelec (St. Gemme sur Loire, France) SenTec (Thurnwill, Switzerland) –	Nasal pressure (nasal canula) thoraco-abdominal movements (inductive plethysmogra-	Attended (PICU)		**2012 AASM scoring rules** Besides apnoeic	

	hospitalized in PICU	for PtcCO2	phy); respiratory sound; body position; pulse oximetry; PtcCO2			and hypopnoeic events, other events were looked for: • progressive simultaneous decrease in airflow and thoracic and abdominal movements accompanied or not by a change in gas exchange, suggestive of a decrease in central drive or global inspiratory muscle weakness •	

						paradoxical breathing with opposition phase on the thoracic and abdominal belts, suggestive of dia-phragmatic dysfunction or weakness of the intercostal muscles	
Jonson et al. (2017) [206]	20 children with enuresis (age 11.1±1.9 years, M/F: 19/1); 21 controls (age 10.8 ± 1.8 years, M/F: 18/3)	Nox T3 respiratory polygraphy (Nox Medical Inc. Reykjavik, Iceland)	Respiratory effort was measured using thoracic and abdominal strain gauges; nasal airflow via nasal cannula and pressure transducer; pulse oximetry. To evaluate the sleep-awake pattern and arousals	Fitted in the hospital by the trained personnel Study done at home, unattended		**2012 AASM scoring rules**	Acceptable recordings from the nasal cannula were obtained in 60% of the patients and 71% of the controls

			during sleep, two central electrodes were fitted (C3, C4) with contralateral mastoid reference electrodes (M1, M2).				
Joyce et al. (2017) [217]	22 typically developing children (16/6); age 39.03±2.11 (25.8–59.5) months. 22 with Down syndrome (13/9); age 36.57±2.07 (24.38–56.48) months.	SOMNOtouch (SOMNOmedics, Germany)	Abdominal and thoracic respiratory inductance plethysmography (RIP); pulse oximetry (Nonin/Masimo); nasal pressure flow with snore sensor; body position sensor; and actigraphy	Unattended	DOMINO LIGHT software (SOMNOmedics, Germany). Manual scoring	**2012 AASM paediatric scoring criteria**	Half of subjects did not tolerate nasal cannula; when the nasal flow signal was lost, RIP sum was used as the recommended alternative measure of airflow

Hill et al. (2016) [218]	202 (110/92) Age 36 (6 to 71) months Down syndrome	SOMNOtouch (Somnomedics, Germany)	Chest and abdominal respiratory inductance plethysmography; pulse oximetry; nasal pressure flow with integral snore sensor; body position sensor; and actigraphy.	Unattended (194/200) and attended (6/200)	Domino Light software (Somnomedics, Germany) Manual scoring. Every 10th study – re-scored by second technologist	**2012 AASM scoring rules** Parental sleep log to interpret sleep onset and offset, and the in-device actigraphy. Blind inter-rater scoring was undertaken for 17 studies. A reliability coefficient of 0.917 (95% CI 0.791 to 0.969) was achieved for the main outcome variable, OAHI, and 0.988 (95% CI 0.967, 0.995) for	Following pilot testing in healthy volunteer children, the equipment was customized for the purpose of the study with both shortened nasal prongs and connecting leads between the chest and abdominal RIP bands. This minimised risk of accidental disconnection of equipment or entanglement. Where nasal flow signal was lost, assuming that good quality RIP

						estimated total sleep time (i.e. total time analysed, TTA), indicating excellent inter-rater scoring agreement. Taking an OAHI threshold of > 5/h as diagnostic criterion for moderate to severe OSAS there was 100% agreement between the two scorers.	and oximetry signals were present, an 'undefined apnoea' was scored, where RIP sum indicated paradoxical breathing in the presence of a minimum 3% oxyhaemoglobin desaturation for at least 2 breaths. Comparing studies completed in the laboratory and the ones completed in the home setting, total time analysed (TTA) was greater in home studies compared to

							laboratory studies but this difference was not statistically significant (median home 514 versus 468 minutes in laboratory; P =0.170).
Kitamura et al. (2016) [213]	67 (39/28) Age: 4.6 ± 0.9 years Mean BMI: 15.1±1.1 kg/m² Children included through kindergarten	Smart Watch PMP-300E (Pacific Medico, Tokyo, Japan)	Nasal pressure; chest movements; body position; snoring; pulse oximetry; actigraphy	Unattended	Manual	**2007 AASM scoring rules** Sleep/wake time was estimated by actigraphy, in conjunction with light-out and wake-up time from a diary	Paediatric OSA was diagnosed based on ICSD-2 and ICSD-3 criteria. Type 3 recordings were performed on 2 consecutive nights. The recording with the

						written by parents. Movement periods >5 min were excluded from estimated sleep time.	longer precise recording time was used in the analyses.
Hamada et al. (2015) [214]	147 children (102 male) 11 months – 6 years	Sleeptester (Fukuda Lifetech)	Not specified	Home, unattended	Manual	**2007 AASM scoring rules**	Sensors were set by patients' guardians.
Pavone et al. (2015) [220]	88 subjects Median age 5.1 [1.0–14.5] years (range 0.3–44.3 years) 20% - adults Prader-Willi syndrome	Equipment not specified SenTec Digital Monitor for $PtcCO_2$	Nasal pressure (nasal canula); thoraco-abdominal movements; tracheal sound; body position, pulse oximetry $PtcCO_2$ in some subjects	Attended		Multicentre study – different scoring criteria (desaturation 3% or 4%); hypopnoea – decrease of at least 50%. Total sleep time – period between lights off/on.	
Taddei et al. (2015) [184]	30 (12/18) Marfan's syndrome	ApneaLink (ResMed, Germany)	Nasal pressure signal, pulse oximetry	Unattended	Automatic; manual review	Apnoeas - cessation of airflow	

	Control - 30 (12/18) untreated children Age: Before treatment 8.9±0.8 years. After rapid maxillary expansion 10.8 ± 0.5 years. After mandibular advancement 12.2 ± 0.4 years					lasting more than 10 s. Hypopnoeas - reduction in airflow of at least 50 % lasting >10 s, associated with a drop in SpO_2 >4 %	
Brietzke et al. (2015) [215]	22 (10/12) Mean age 5.4 years [range 2.4-8.4 years]	SNAP test (SNAP Diagnostics, USA)	Sound (measured via an oronasal cannula); airflow; pulse oximetry; one respiratory effort channel	Unattended			SNAP test: detailed acoustical analysis of snoring. Snoring index (snoring events per hour), the resistance occurrence percentage (percentage of breathing events with a snoring type

							noise), average snoring loudness above the baseline noise (dB), maximal snoring loudness above the baseline (dB), average snoring frequency (Hz). Based on the frequency of the snoring sound, the SNAP test can determine whether it is predominantly palatal or non-palatal in origin.
Amaddeo et al. (2015) [223]	26 (20/6) Age: 7.8 ± 5.1 3 overweight, 3 obese Children and	SOMNOscreen (SOMNOscreen plus PSG+, SOMNOmedics GmbH, Germany); or CID 102* (Cidelec,	Airflow (pneumotacho-graph); airway pressure in the CPAP line; body position; body movements; thoracic and	Attended		According to the SomnoNIV group consensus opinion	Transcutaneous carbon dioxide tension was measured simultaneous

	infants undergoing CPAP treatment	Angers, France)	abdominal movements assessed with piezoelectric belts (somnoSCREEN) or inductance belts (CID 102*); pulse oximetry; photoplethy-smographic pulse-wave amplitude				ly with PG. Nocturnal gas exchange cannot predict PG results and thus a systematic sleep study is necessary for the routine follow-up of children treated with CPAP.
Kasapkara et al. (2014) [185]	19 (12/ 7) Age: 7.97 ± 4.90 years Patients with mucopolysaccha-ridoses (MPS)	ApneaLink (Resmed, Poway, CA)	Nasal respiratory pressure signal; pulse oximetry	Unattended	Manual	**Apnoea -** cessation of airflow lasting more than 10 s. **Hypopnoea -** amplitude of nasal airflow fell to 50 % of the average amplitude of the two preceding breaths **Oxygen desaturation** ≥3%	The prevalence of OSAS was 94.7 % (18/19) in patients with MPS

						decrease	
Brockman et al. (2013) [197]	101 (47/54) Median age 2.8 (0-15.4) years Children with habitual snoring or suspicion of having apnoeas	Embletta Gold III (Embla, Broomfield, Colorado, USA)	Nasal flow (pressure transducer cannula); thoracic and abdominal movements; pulse oximetry; heart rate (ECG); position sensor	Unattended (75/101); attended (26/101)	Manual	**2007 AASM scoring rules** Artefact free recording time was used to define respiratory events. Artefacts were defined as the loss of signals in nasal pressure, thoracic and abdominal movements, pulse oximetry, and heart rate.	OSAS was defined as a MOAHI ≥1/h. Upper airway resistance syndrome was defined as an RDI ≥ 1/h and AHI < 1/h. Primary snorers had an RDI and AHI < 1/h. A portable home recording method for OSAS is possible and technically feasible in children regardless of their age or gender. Unattended home recordings seem to perform as well as supervised in-hospital measurements.

Kitamura et al. (2014) [212]	170 (85/85) Age: 7.01 ± 0.69 years Mean BMI: 15.4 ±1.9 kg/m²	Smart Watch PMP-300E (Pacific Medico, Tokyo, Japan)	Nasal pressure; chest movements; body position; snoring; pulse oximetry; actigraphy	Unattended	Manual	**2007 AASM scoring rules** Sleep/wake time was estimated by actigraphy. Movements periods >5 minutes were excluded from estimated sleep time.	3 diagnostic criteria for paediatric OSAS were compared: AHI≥5; oAHI≥1 and ICSD II criteria.
Alonso-Álvarez et al. (2012) [194]	100 (68/32) Age: 4.17± 2.06 years Mean BMI: 16.05 ± 2.53 kg/m^2 Children with snoring or respiratory pauses and indication for adenotonsillectomy	Edentec Polygraph Monitoring System; model 3711 (Edentrace II, Minnesota, USA)	Oronasal flow (thermistor); chest movements (impedance plethysmo-graphy); body position (sensor position); snoring (microphone); pulse oximetry	Attended	Manual	**2007 AASM scoring rules** **Apnoea** - interruption of oronasal flow for at least two respiratory cycles, with maintenance of thoracoabdominal effort	OSAHS was defined as AHI ≥4.6 events/h Success rate of adeno-tonsillectomy was 88.4%

						(**obstructive apnoea**) or without thoracoabdominal effort (**central apnoea**) **Hypopnoea** - a decrease of at least 50% in the amplitude of oronasal flow for the duration of two respiratory cycles, with maintenance of respiratory effort associated with a drop in oxygen saturation of at least 3% Respiratory events divided by	

						total study time	
Hamada and Iida (2012) [211]	48 (34/14) Median age: 5 (2-11) years	The Sleeptester (Fukuda Lifetech Inc., Tokyo, Japan)	Oronasal airflow; thoracoabdominal effort; snoring; body position; and oximetry	Unattended	Manual using SAS-100 software	2007 AASM scoring rules	Attended home monitoring by guardians using a portable device can be useful in the perioperative assessment of paediatric OSAS.
Luna-Paredes et al. (2012) [190]	44 (20/24) Age: 5 years (8 months – 14 years) Patients with major craniofacial abnormalities	Apnoescreen Pro (Erich Jaeger GmbH & CoKg, Wuerzburg, Germany); and Grass Technologies Aura PSG system (Astro-Med Inc., Richmond, VA, USA)	Pulse oximetry; chest and abdominal respiratory movements (plethysmo-graphic bands); wrist actigraphy; nasal airflow (nasal cannula) and oronasal airflow (thermistor)	Attended	Manual	**Obstructive apnoea:** complete cessation of airflow (measured both by the nasal cannula and the oronasal thermistor) for more than 2 breaths; **hypopnoea:** ≥50% reduction in	Children with craniofacial anomalies have high frequency of symptoms of airway obstruction and obstructive sleep apnoea

						respiratory airflow accompanied by a decrease of ≥3% in SpO_2; **central apnoea:** cessation of nasal, abdominal and thoracic airflow with a duration of at least two breaths if associated with desaturation.	
Plomp et al. (2012) [200]	13 (5/8) children with Treacher Collins Syndrome	Embletta Portable Diagnostic System; analysed using somnologica (Medcare Flaga, Reykjavik, Iceland)	Nasal airflow, Chest & abdomen wall motion, Snoring, SpO_2 and pulse waveform.	Unattended (Home study)	Not specified	**In keeping with AASM but not specified.**	Central apnoeas were not counted towards AHI. OSAS was defined as OAHI>1/h.
Heimann et al.	18 (13/5)	Alice 3 and 3.5	Heart and respiratory rate	Attended	Manual	**Periodic breathing**	6-h cycle in three

(2010) [224]	Premature infants Gestational age: 28 (24-32) weeks Chronological age 36 days (range 7-64 days).	(Respironics Carlsbad, Calif., USA; Healthdyne Technologies, Marietta; GA, USA)	and pattern with ECG monitoring and respirogram; pulse oximetry			(occurrence of 3 or more successive central apnoeic pauses of 4 s or longer, each separated by less than 20 s of breathing) **Apnoeas** - short (≥ 3 but < 5 s), medium (≥ 5 but < 10s), long (≥10s)	subsequent measurement series of 120 min per position (prone, supine, skin to skin care position)

Table e29. Definition of obstructive sleep apnoea syndrome (OSAS) in paediatric studies involving type III devices (respiratory polygraphy-PG) without comparison to full attended polysomnography (PSG).

Study	**AHI**	**Obstructive AHI**	**RDI**	**Obstructive RDI**	**Obstructive apnoea index**	**Frequency of**	**Central apnoea**

						respiratory events during CPAP treatment	**index (abnormal)**
Chiner et al. (2020) [181]	AHI<3/h: normal AHI 3-5/h: mild OSAS AHI 5-10/h: moderate OSAS AHI >10/h: severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Corbelli et al (2020) [199]	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Joyce A et al. (2020) [216]	Not defined	Obstructive AHI≥1/h: OSAS	Not defined	Not defined	Not defined	Not defined	Not defined
Michelet et al. (2020) [198]	AHI ≥OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Vezina et al. (2020) [208]	Not defined	Not defined	Not defined	Not defined	Obstructive apnoea index <2/h: normal Obstructive apnoea index 2-4.9/h: mild OSAS Obstructive apnoea index 5-9.9/h: moderate	Not defined	Not defined

					OSAS Obstructive apnoea index ≥10/h: severe OSAS		
Abel et al. (2019) [195]	Not defined	OAHI 1–5/h: mild OSAS OAHI >5/h: moderate-to-severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined
Blanc et al. (2019) [205]	Not defined	OSAS was defined by obstructive apnoea index (OAI) > 1/h or obstructive apnoea/hypopnoea index (OAHI) > 1.5/h.	Not defined	Not defined	Not defined	Not defined	Central apnoea syndrome was defined as central apnoea index > 1/h
Giabicani et al (2019) [192]	AHI 1.5-5/h: mild OSAS AHI 5-10/h: moderate OSAS AHI >10/h: severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Central AHI >0.45/h: abnormal in children between 3-5 years Central AHI >0.85/h: abnormal in children older than 6 years.

Kingshott et al. (2019) [166]	Not defined	OAHI 1/h to <5/h: mild to moderate OSAS OAHI >5/h: moderate to severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined
Masoud et al. (2019) [203]	AHI 1.5 to <5/h: mild OSAS AHI 5 to < 10/h: moderate OSAS AHI ≥ 10/h: severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Orntoft et al (2019) [193]	AHI < 2/h: normal 2 ≤AHI <5/h: mild OSAS 5 ≤ AHI <10/h: moderate OSAS AHI ≥ 10/h: severe OSAS.	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Pabary et al (2019) [196]	AHI ≥5/h (or AHI ≥1/h) was indicative of SDB.	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Stöberl et al. (2019) [187]	Not defined	OAHI ≥1/h	Not defined	Not defined	Not defined	Not defined	Not defined
Tabone et al. (2019) [182]	Not defined	OAHI ≥2/h	Not defined	Not defined	Not defined	Not defined	CAI ≥5/h
Modesti-Vedolin et al. (2018) [188]	Not defined	Not defined	RDI ≥ 1.5/h: OSAS	Not defined	Not defined	Not defined	Not defined
Paoloni et al. (2018) [219]	AHI >1/h: OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined

Caggiano et al. (2017) [191]	AI<1 or AHI <1.5: normal AHI 1.5-5/h: mild OSAS AHI 5-10/H: moderate OSAS AHI>10/h: severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Dudoignon (2017) [183]	Not defined	OAHI <2/h: normal	Not defined	Not defined	Not defined	Not defined	CAI <5/h: normal
Griffon et al. (2017) [222]	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Jonson et al. (2017) [206]	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Joyce et al. (2017) [217]	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Lecka-Ambroziak et al. (2017) [210]	AHI<1/h normal AHI 1-5/h: mild OSAS AHI 5-10: moderate OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Scalzitt et al (2017) [201]	AHI >1/h: OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Hill et al. (2016) [218]	Not defined	OAHI 1/h to <5/h: mild to moderate OSAS OAHI >5/h: moderate to severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined

Kitamura et al. (2016) [213]	Not defined	OSAS: OAHI≥1/h in combination with habitual snoring and/or laboured/obstructed breathing, and at least 1 additional OSAS related symptom	Not defined	Not defined	Not defined	Not defined	Not defined
Amaddeo et al. (2015) [223]	Not defined	Not defined	Not defined	Not defined	Not defined	Total respiratory events on PG <1.5/h: normal PG under CPAP 1.5 – 4.9/h: mild frequency of respiratory events under CPAP 5 – 10/h: moderate frequency of respiratory events under CPAP >10/h: high frequency of respiratory events under CPAP	Not defined
Brietzke et al. (2015)	AHI>1: OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined

[215]							
Hamada M et al. (2015) [214]	AHI ≥5/h: OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Pavone et al. (2015) [220]	AHI ≥1.5/h: sleep disordered breathing for paediatric patients AHI ≥5/h: sleep disordered breathing for adult patients	MOAHI (mixed obstructive apnoea – hypopnoea index): <1.5/h: normal	Not defined	Not defined	Not defined	Not defined	Central apnoea index <1/h: normal
Taddei et al. (2015) [184]	AHI >5/h AHI >10/h Severity not defined	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Kasapkara et al. (2014) [185]	AHI <1/h: normal ≥1 – 5/h: mild OSAS 5 – 10/h: moderate OSAS >10/h: severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Kitamura et al. (2014) [212]	AHI ≥1-5/h: OSAS (various cut-off values used)	OAHI≥1-5/h: OSAS (various cut-off values used)	Not defined	Not defined	Not defined	Not defined	Not defined
Brockmann et al. (2013) [197]	Not defined	MOAHI ≥1/h: OSAS	Upper airway resistance syndrome was defined as an RDI ≥1/h and AHI <1/h. Primary snorers had an RDI and AHI < 1/h.	Not defined	Not defined	Not defined	Not defined

Alonso-Álvarez et al (2012) [194]	AHI ≥4.6/h: OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Hamada et al (2012) [211]	AHI >5/h: OSAS	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined
Luna-Paredes et al (2012) [190]	Not defined	OAHI < 3/h: normal 3 – 5/h: mild OSAS 6 – 10: moderate OSAS > 10/h: severe OSAS	Not defined	Not defined	Not defined	Not defined	Not defined
Plomp et al (2012) [200]	Not defined	OAHI <1/h: normal 1-5 /h: mild 5-24/h: moderate >24/h: severe	Not defined	Not defined	Not defined	Not defined	Central events not assessed
Heimann et al (2010) [224]	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined	Not defined

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