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Performance of Non-invasive Devices in Evaluation of Periodic Limb Movements and Sleep-disordered Breathing

■

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to my family

Abstract

Polysomnography is considered as the gold standard in diagnosing sleep disturbances. These studies are quite expensive as they are performed in sleep laboratories with continuous attendance by a technician or nurse. Recordings with large amounts of cables and measurement devices can be inconvenient for the patients. Therefore there is a constant need to develop ambulatory, unmolested, inexpensive but reliable methods for clinical sleep investigations.

The aim of this thesis was to study the suitability of mattress type movement sensors for the detection of periodic limb movements and in the characterization of sleep-disordered breathing. Also the performance of a new method, compressed tracheal sound analysis, in the evaluation of sleep-disordered breathing is presented in the thesis.

Patients with restless leg syndrome have very often periodic limb movements in their sleep recordings. This finding is considered as a supportive criterion in restless legs diagnostics. The gold standard in recording of periodic limb movements is anterior tibialis electromyography. The studies in this thesis showed that the static-charge-sensitive bed (SCSB) and Emfit sensors detect periodic limb movements reliably and the periodic movement indexes were quite comparable with the gold standard.

A special signal feature, a spiking phenomenon is seen with both the SCSB and Emfit sensors. Spiking can also reliably be detected with an automatic method that does not need any recording-specific tuning before the analysis. In sleep-disordered breathing prolonged spiking was found to be associated with an increase in transcutaneously measured carbon dioxide ($TcCO_2$). During apnea and hypopnea episodes no significant change in $TcCO_2$ level was observed. The non-round

inspiratory flow shapes, which are related to flow limitation, were most constant phenomenon during prolonged spiking episodes.

Snoring is very often associated with sleep-disordered breathing and an important reason for referrals to sleep studies. Analyses of snoring and tracheal sound can be used in the evaluation of sleep related breathing disorders. In the compressed tracheal sound analysis the signal curve can visually be divided into three distinct patterns. The characteristics of the patterns differed significantly from each other. Based on the diverse appearances of the patterns, breathing with apneas/hypopneas, flow limitation and normal breathing could be distinguished.

The studied non-invasive methods, the SCSB and Emfit seem to be suitable for detecting periodic limb movements. Emfit and compressed tracheal sound analysis can help in the evaluation of sleep-disordered breathing. They all can be used as parts of larger recording systems or even as stand-alone devices. The recordings are inexpensive, easy to perform and cause minimal if any disturbance to a patient. Thus they are very suitable for ambulatory sleep recording systems. Additionally, they all have special features that can detect and characterize the still poorly understood prolonged flow limitation.

Tiivistelmä

Polysomnografiaa pidetään vertailustandardina unihäiriöiden diagnostiikassa. Sillä tehdyt tutkimukset ovat melko kalliita, koska ne tehdään unilaboratorioissa hoitajan valvonnassa. Rekisteröintijärjestelmässä on paljon johtoja ja mittalaitteita, jotka voivat häiritä potilasta. Siksi tarvitaan jatkuvaa kehitystyötä, jotta kliinisiin uni- ja yörekisteröinteihin saadaan ambulatorisia, nukkujaa häiritsemättömiä, edullisia, mutta luotettavia metodeja.

Tämän väitöskirjan aihe oli tutkia patjatyyppisten liikeantureiden soveltuvuutta jaksoittaisten raajaliikkeiden havaitsemiseen ja unenaikaisen hengityshäiriön selvittelyyn. Väitöskirjassa esitellään myös uuden metodin, henkitorven päältä rekisteröitävän tiivistetyn äänianalyysin käytettävyyttä unenaikaisten hengityshäiriöiden tutkimuksessa.

Levottomat jalat -oireyhtymästä kärsivillä potilailla on unirekisteröinneissä hyvin usein raajojen jaksoittaisia liikkeitä. Tätä löydöstä pidetään levottomat jalat -diagnoosia tukevana löydöksenä. Jaksoittaisten raajaliikkeiden mittauksen standardimenetelmänä käytetään tibialis anterior -lihaksen päältä rekisteröityä lihassähkötoimintaa. Väitöskirjan osajulkaisuissa todettiin, että sekä static-charge-sensitive bed (SCSB)- ja Emfit-anturit havaitsevat hyvin jaksoittaiset raajaliikkeet ja että jaksoittaisten liikkeiden määrä tuntia kohden oli samaa luokkaa kuin standardimenetelmällä saadut tulokset.

Sekä SCSB- että Emfit-antureilla on mahdollista todeta erityinen signaalin ominaisuus, ns. piikikkyys-ilmiö. Kyseinen ilmiö voidaan erottaa myös automaattisella menetelmällä, joka toimii ilman edeltävää signaalin muokkausta. Unenaikaisen hengityshäiriön yhteydessä pitkäkestoisen piikikkyyden havaittiin liittyvän ihon läpi mitatun hiilidioksidiosapaineen nousuun. Hengityskatkoja (apnea) tai jaksoittaista vajaahengitystä (hypopnea) sisältävien jaksojen aikana

hiilidioksidiosapaineen nousua ei havaittu. Virtausrajoitukseen liittyvien, ei-pyöreiden sisäänhengitysten virtausprofiilien määrä oli suurin pitkittyneissä piikikkyysjaksoissa.

Kuorsaus liittyy hyvin usein unenaikaisiin hengityshäiriöihin, ja se on tärkeä unitutkimuksiin lähettämisen syy. Kuorsauksen ja henkitorven päältä rekisteröitävän äänen analyyseja voidaan käyttää unenaikaisten hengityshäiriöiden tutkimiseen. Henkitorven päältä rekisteröitävä tiivistetty äänisignaali voidaan visuaalisesti jakaa kolmeen eri toisistaan erottuvaan muotoon. Niiden ominaisuuksien todettiin eroavan merkittävästi toisistaan. Näiden ominaisuuserojen perusteella apneoita/hypopneoita, virtausrajoitusta ja normaalia hengitystä sisältävät jaksot voitiin erottaa toisistaan.

Tutkitut ei-invasiiviset tutkimusmenetelmät, SCSB ja Emfit näyttävät sopivan jaksoittaisten raajaliikkeiden havaitsemiseen. Emfit ja henkitorven päältä rekisteröitävä tiivistetty äänianalyysi voivat auttaa unenaikaisten hengityshäiriöiden selvittelyssä. Kaikkia näitä antureita voidaan käyttää osina laajempia järjestelmiä tai erillisinä rekisteröintilaitteina. Niillä tehtävät rekisteröinnit ovat edullisia, suoritukseltaan helppoja ja häiritsevät vain vähän jos lainkaan potilasta. Niinpä ne ovat erittäin sopivia liikuteltaviin unirekisteröintijärjestelmiin. Lisäksi niillä kaikilla on erityisominaisuuksia, joiden avulla voidaan havaita ja luonnehtia vielä huonosti tunnettua pitkittynyttä virtausrajoitusta.

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Abbreviations

AHI	apnea-hypopnea index
BMI	body mass index
CO_2	carbon dioxide
Emfit	Electromechanical film
EMG	electromyography
ICSD	International Classification of Sleep Disorders
IRR	increased respiratory resistance
NREM sleep	non-rapid eye movement sleep
OSA	obstructive sleep apnea
OSAS	obstructive sleep apnea syndrome
OSAHS	obstructive sleep apnea-hypopnea syndrome
PLMs	periodic limb movements
PSG	polysomnography
REM sleep	rapid eye movement sleep
RERA	respiratory effort-related arousal
RLS	restless legs syndrome
SCSB	static-charge-sensitive bed
SDB	sleep-disordered breathing
$TcCO_2$	transcutaneous carbon dioxide
UARS	upper airway resistance syndrome

List of Original Publications

The dissertation is based on the following five original publications, referred to in the text by their Roman numerals I-V. In addition, some unpublished data are presented. The original publications are reprinted with the permission of the copyright holders.

I   Rauhala E, Erkinjuntti M and Polo O (1996): Detection of periodic leg movements with a static-charge-sensitive bed. J Sleep Res 5:246-250.

II   Rauhala E, Virkkala J and Himanen SL (2009): Periodic limb movement screening as an additional feature of Emfit sensor in sleep-disordered breathing studies. J Neurosci Methods 178:157-161.

III   Alametsa J, Rauhala E, Huupponen E, Saastamoinen A, Varri A, Joutsen A, Hasan J and Himanen SL (2006): Automatic detection of spiking events in EMFi sheet during sleep. Med Eng Phys 28:267-275.

IV   Rauhala E, Himanen SL, Saastamoinen A and Polo O (2007): Prolonged spiking in the Emfit sensor in patients with sleep-disordered breathing is characterized by increase in transcutaneous carbon dioxide. Physiol Meas 28:1163-1173.

V   Rauhala E, Hasan J, Kulkas A, Saastamoinen A, Huupponen E, Cameron F and Himanen SL (2008): Compressed tracheal sound analysis in screening of sleep-disordered breathing. Clin Neurophysiol 119:2037-2043.

1. Introduction

The mysteries of sleep have exercised the minds of poets, artists and philosophers for thousands of years. Also scientists have always been intrigued about the function of sleep. Despite remarkable development in understanding the mechanisms of sleep and neurobiology of sleep, we still do not know the exact function of sleep. Several theories of the function of sleep have been proposed, but no comprehensive clarification has been presented. The importance of sleep disorders has been recognized by physicians only for the last 40 years, and no more than 30 years objective diagnostic procedures have been routinely performed for their diagnosis.

The oldest method in human sleep research is undoubtedly the direct observation of the sleeping person. Continuous direct observation to characterize sleep has been used even in modern times in conjunction with other methods (Gastaut and Broughton, 1965, Muzet et al., 1974, Gardner and Grossman, 1975). Movement activity is always associated with normal sleep. Before the development and systematic use of electrophysiological techniques for studying sleep, movements were one of the few available physiological indices of sleep. The systematic quantification and timing of movements has depended on technology. Szymanski introduced in 1914 a method which has later been called the "sensitive bed" principle of movement registration (Szymanski, 1914). Movements of the subject cause displacement of the bed which is then recorded to a recording device. The modern sleep research uses a multiplicity of techniques and equipments to study and characterize sleep but simple movement sensors still have to offer for sleep research.

Snoring is also a very well-known and frequent phenomenon during sleep. Despite this long-standing awareness of snoring and the havoc it creates in many bedrooms, it was firmly believed in the 19th and even in the first part of the 20th century that snoring was nothing but a social nuisance, without any adverse health

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consequences to the snorer. From the latter part of last decade to today snoring has been treated with various surgical methods with variable results. After 1980 there has been a growing understanding about possible adverse health effects of snoring. Thus the serious scientific research of snoring, snoring sound and not infrequently associated flow limitation has been increased substantially during the last thirty years.

2. Review of the literature

2.1 Classification of sleep disorders

The earliest, primarily symptom-based, sleep disorder classification systems go back to the 19th century, and they formed the basis for the modern classifications. In 1885, Henry Lyman, a professor of neurology in Chicago, classified insomnias into two groups: those resulting from either abnormal internal or physical functions; or from external, environmental influences (Lyman, 1885). Roger added other sleep disorders to the classification and classified them into three main groups, insomnias, hypersomnias and parasomnias (Roger, 1932); a classification very similar to the system used today. Kleitman developed the system further by also adding disorders of excessive sleepiness with main forms of narcolepsy and encephalitis lethargica to the classification (Kleitman, 1939).

The Diagnostic Classification of Sleep and Arousal Disorders was released in 1979 (Association of Sleep Disorders Centers, 1979). The second edition of the International Classification of Sleep Disorders (ICSD-2) was published in 2005 (American Academy of Sleep Medicine, 2005). This system classifies sleep disorders into eight major categories: (1) insomnia, (2) sleep related breathing disorders, (3) hypersomnias of central origin, (4) circadian rhythm sleep disorders, (5) parasomnias, (6) sleep related movement disorders, (7) isolated symptoms and normal variants, (8) other sleep disorders. ICSD-2 includes over 70 specific diagnoses within these eight major categories, as well as two appendices for classification of sleep disorders associated with medical or psychiatric disorders. For example periodic limb movements and restless legs syndrome are classified into the sleep related movement category and snoring into category of isolated symptoms, normal variants and unsolved issues.

2.1.1 Obstructive sleep-related breathing disorders

2.1.1.1 *Obstructive sleep apnea syndrome*

The obstructive sleep apnea syndrome (OSAS) is a condition characterized by repeated episodes of upper airway closure during sleep. Total closures are designated as obstructive apneas and partial closures as obstructive hypopneas. OSAS is associated with a variety of symptoms and objective findings. The most common presenting complaints are loud, disruptive, interrupted snoring, associated with unrefreshing sleep and excessive daytime sleepiness or fatigue. OSAS is common and has many kinds of adverse systemic consequences (Peppard et al., 2000, Parish and Somers, 2004). Severe OSAS significantly increases the risk of fatal and non-fatal cardiovascular events and continuous positive airway pressure (CPAP) treatment reduces this risk (Marin et al., 2005). Arterial hypertension and brain infarctions have been connected with OSAS and snoring (Partinen, 1995). Executive functions are the most defected cognitive domain in OSAS (Saunamaki and Jehkonen, 2007).

No satisfactory surveys have been conducted to determine the true prevalence of OSAS in general population. One problem is that while some prevalence studies of OSAS use polysomnographic diagnostics, some other studies are done with questionnaires. This has led to a wide variation of reported prevalences. Most surveys have been cohort studies which have estimated the prevalence OSAS to be between 0.9 – 4% (Lavie, 1983, Telakivi et al., 1987, Gislason et al., 1988, Young et al., 1993).

A typical OSAS patient is an obese, middle-aged or older man with a large neck circumference and possibly short mandible. In other words OSAS is associated in high body mass, male gender and some anatomical features.

2.1.1.2   Upper airway resistance syndrome

The upper airway resistance syndrome (UARS) is a breathing disorder without periodic apneas and hypopneas. UARS was introduced by Guilleminault (Guilleminault et al., 1993) and it represents a form of sleep-disordered breathing in which repetitive increases in resistance to airflow within the upper airway lead instead of apneas to flow limitation with following brief arousals and consequential daytime somnolence (Guilleminault et al., 1993, Exar and Collop, 1999, Bao and Guilleminault, 2004).   Patients with UARS display progressive increases in respiratory effort, indicated by progressively more negative esophageal pressure, terminated by a sudden change in pressure to a less negative level and an arousal. These respiratory effort-related arousal (RERA) events should be scored if they last ≥ 10 seconds (1999).

There are very little epidemiological data on UARS and flow limitation. The prevalence has been estimated at 8.4% of all patients referred for sleep-disordered breathing (Velamuri, 2006). UARS is found in younger age group than in OSAS and the male predominance present is OSAS is not noted in UARS; a more equal distribution between sexes is seen. UARS patients are usually slim or their body mass index is normal whereas OSAS patients are most often obese. Chronic insomnia tends to be more common in patients with UARS than with OSAS and UARS patients are more likely to complain of fatigue than sleepiness. Patients with UARS also have more somatic complaints such as fibromyalgia, chronic pain and headaches (Bao and Guilleminault, 2004).

2.1.1.3   Snoring

Snoring is a noise produced when vibration occurs at several levels of upper airway. It can be associated with various degrees of upper airway resistance. It may be heard after complete airway obstruction, with a significant hypopnea, or with hypoventilation, leading to a cohort of symptoms (Robinson and Guilleminault, 1999). Snoring is known to be an important symptom of obstructive sleep apnea

syndrome (Bliwise et al., 1991, Guilleminault et al., 1993). It has also been shown that systemic hypertension, cardiac dysfunction, angina pectoris, and cerebral infarction are more frequent among habitual snorers (Koskenvuo et al., 1985, Norton and Dunn, 1985, Jennum et al., 1992, Nieto et al., 2000, Lindberg et al., 2007). However, Marin and co-workers did not find any cardiovascular complications in patients with simple snoring (Marin et al., 2005).

Snoring is a very common symptom among adult population. The prevalence of snoring has found to be higher among men than women and to increase with age. Surveys have found that 20-50% of subjects are habitual or frequent snorers (Lugaresi et al., 1980, Fitzpatrick et al., 1993, Koskenvuo et al., 1994). Epidemiologically snoring has been associated with increased body mass index, smoking, physical inactivity, hostility and morning tiredness (Koskenvuo et al., 1994).

2.1.1.4   *Inspiratory flow limitation*

Flow limitation occurs when increased negative esophageal pressure is not accompanied by a flow increase. Flow limitation depends on the interaction between the negative pleural pressure, which tends to collapse the upper airway, and upper airway muscle activity, which helps to keep the airway open.

Flow limitation is characterized by a constant, or even decreased, inspiratory flow, occurring simultaneously with an increased, i.e. more negative, intraluminal pressure. Sleep-related inspiratory flow limitation has been reported in healthy, nonobese, nonsnoring subjects (Hudgel et al., 1988), healthy snorers (Skatrud and Dempsey, 1985, Liistro et al., 1991), in UARS (Rees et al., 2000) and in OSA patients (Liistro et al., 1991). It is possible that in OSAS upper airway narrowing may progress to airway occlusion, so that inspiratory flow limitation is followed by an apnea (Lugaresi et al., 1983).

Flow limitation is reliably identified with nasal cannula as the shape of the inspiratory flow signal is distorted (Hosselet et al., 1998). Recent studies presented below suggest that flow limitation is a clinically significant finding. Flow limitation with arousals represents the physiological basis of the UARS which appears as respiratory effort-related arousals in PSG (Bao and Guilleminault, 2004). Cracowski and co-workers have found that RERA events represented only 5.3% of all obstructive nonapneic events in patients with moderate obstructive sleep apnea/hypopnea (Cracowski et al., 2001). However, 14.8% of the events were classified as indeterminate owing to a 30-50% airflow decrease without arousal or desaturation or an airflow decrease less than 30% without arousal. RERA is defined to last at least 10 seconds but also a short-term (even merely one inspiration) non-round flow limitation pattern in nasal cannula has been associated to increased breathing effort and arousals (Johnson et al., 2005). However, flow limitation is also found in a prolonged pattern. Hernandez and co-workers have found that this prolonged flow limitation is quite common; 73-83% of the subjects had prolonged flow limitation in some degree (14-26% of TST) (Hernandez et al., 2001). The authors stated that the pathophysiologic significance of these periods of prolonged flow limitation has not been established, and further investigation is required.

2.1.2 Movement disorders during sleep

Movement disorders during sleep can be classified into physiologic (normal) and pathological (abnormal) types (Chokroverty, 2000). Physiologic motor activity during sleep includes postural shifts, body and limb movements, physiologic fragmentary myoclonus consisting of phasic muscle bursts seen typically in REM sleep, and hypnic jerks consisting of sudden muscle jerks involving either the whole body or part of it at sleep onset during the transition between wakefulness and sleep.

The most common pathological movements that occur during sleep consist of restless legs syndrome or periodic movements in sleep and body movements seen in patients with sleep apnea syndrome. Also motor parasomnias, nocturnal seizures, diurnal involuntary movements persisting during sleep, drug-induced nocturnal

dyskinesias, jerks and excessive fragmentary myoclonus seen in many sleep disorders are included in this group. Additionally hyperekplexia or exaggerated startle syndrome, sleep-related panic attacks, dissociated disorders, fatal familial insomnia, post-traumatic stress disorder, and sleep paralysis as seen in patients with narcolepsy-cataplexy syndrome are classified in the group of abnormal movements during sleep.

The new International Classification of Sleep Disorders (American Academy of Sleep Medicine, 2005) has separated sleep-related movement disorders into simple repetitive movement disorders (such as PLMS, leg cramps (Butler et al., 2002), bruxism (Sjoholm et al., 1995), and rhythmic movement disorder (Dyken et al., 1997) and parasomnias.

2.1.2.1 Restless legs syndrome

RLS is characterized by intense disagreeable feelings in the legs with an urge to move the legs to get relief (Hening et al., 2000). The symptoms appear mostly or exclusively when the patient is at rest, particularly in the evening and nightfall (Walters, 1995). The symptoms occur at sleep onset and may also occur on awakenings in the middle of the night. Idiopathic RLS is often a lifelong condition.

The severity of the symptoms varies widely, ranging from occurring only occasionally in a stressful situation to nightly and severe, with almost total disruption of normal sleep patterns. Patients describe abnormal sensations as a creeping, heating, prickling, aching, numbing, burning and sometimes painful feeling.

The International RLS Study Group (IRLSSG) has developed an RLS severity rating scale to evaluate this wide range of symptom severity (Allen et al., 2003). The IRLSSG determined that all four essential criteria are required to make the diagnosis of RLS. Three clinical features may support the diagnosis in uncertain clinical cases,

and three additional features of the disorder deserve consideration when evaluating the patient with a potential diagnosis of RLS (Table 1).

The pathophysiologic cause of RLS is not well clarified, although dopamine and iron abnormalities have been implicated. A major genetic contribution is suggested by the high percent of patients with affected first-degree relatives (often more than 50% in some studies), with many pedigrees suggesting a dominant inheritance (Chokroverty et al., 2003). Secondary RLS may be associated with a variety of causes. The most important associations with pathologic contitions are with iron deficiency (with or without frank anemia), uremia, rheumatoid arthritis, and diabetes (Hening et al., 2000).

Table 1. Diagnostic criteria for RLS

Essential criteria
  (1) An urge to move the legs, usually accompanied or caused by uncomfortable and unpleasant sensations in the legs.
  (2) The urge to move or unpleasant sensations begin or worsen during periods of rest or inactivity such as lying or sitting.
  (3) The urge to move or unpleasant sensations are partially or totally relieved by movement, such as walking or stretching, at least as long as the activity continues.
  (4) The urge to move or unpleasant sensations are worse in the evening or night than during the day or only occur in the evening or night.

Supportive criteria
  a) Positive family history of RLS.
  b) Positive response to dopaminergic therapy.
  c) Periodic limb movements during wakefulness or sleep.

Associated clinical features
  * Natural clinical course of the disorder: Can begin at any age, but most patients middle-aged or older. Most often a progressive but sometimes static clinical course. Remissions possible.
  * Sleep disorders a frequent but unspecific symptom of the RLS.
  * Medical evaluation/physical examination: The neurological examination usually normal.
  * Probable causes for secondary RLS should be excluded. A low serum ferritine (<45-50 μg/l) may be found in RLS patients.

Prevalence estimates for restless legs syndrome have generally ranged from 1% to 15%. Two early studies in general medical populations using clinical diagnosis led to estimates between 3.2% and 5% (Ekbom, 1945, Strang, 1967). A Canadian study found a prevalence of 15% which increased with age and was higher among women (Lavigne and Montplaisir, 1994). The studies that have employed the

IRLSSG essential criteria report most frequently prevalences of RLS between 6% and 12% (Berger and Kurth, 2007).

2.1.2.2   Periodic limb movements (PLMs) in sleep

PLMs are characterized by periodically recurring movements, particularly dorsiflexion of the ankles and sometimes flexion of the knees and hips occurring periodically at an average interval of 20 to 40 seconds during NREM sleep. Movements generally affect both legs but may predominantly affect one leg or alternate between the two legs. Occasionally, PLMs may be noted in the upper limbs (Yokota et al., 1995, Chabli et al., 2000). PLMs are found in at least 80% of cases of RLS (Montplaisir et al., 1997), and RLS is noted approximately 30% of cases of PLMs. PLMs are quite common also among snorers and OSAS patients (Coleman et al., 1980, Chervin, 2001, Haba-Rubio et al., 2005) and during continuous positive airway pressure treatment (Fry et al., 1989). PLMs may occur in normal individuals and can be found in polysomnograms without other findings (Bixler et al., 1982, Polo-Kantola et al., 2001, Carrier et al., 2005). PLMs are common particularly in subjects older than 65 years in whom there is an incidence of approximately 44% (Coleman et al., 1983).

2.2   Polysomnography in diagnosing sleep disorders

Internationally the standard approach to diagnosing sleep disorders is in-laboratory, technician-attended polysomnography. It includes simultaneous recording of various physiologic characteristics, which allow assessment of sleep stages and wakefulness, respiration, cardiocirculatory functions, and body movements. Sleep staging is based on electroencephalogram (EEG), electro-oculogram (EOG), and electromyogram (EMG) of chin muscles. PSG study also records airflow, respiratory effort, snoring, electrocardiogram (ECG), oxygen saturation, and limb muscle activity and body positions. Some laboratories also record intraesophageal pressure, end-tidal or transcutaneous carbon dioxide and general movement activity.

Simultaneous video monitoring of patient behaviour is important in diagnosing parasomnias or nocturnal seizures.

2.2.1 Sleep stage scoring

The forty year old sleep stage scoring system according to the committee led by Allan Rechtschaffen and Anthony Kales has recently been renewed. However, in many European countries old guidelines are still used. According to Rechtschaffen and Kales technique standard EEG electrode derivations are C3-A2 or C4-A1 (Rechtschaffen and Kales, 1968). The recommended derivations of the very new American Academy of Sleep Medicine (AASM) manual are F4-M1, C4-M1 and O2-M1 (Iber et al., 2007).

Electro-oculogram (EOG) records corneoretinal (a relative positivity at the cornea and a relative negativity at the retina) potential difference. Any movement of the eyes changes the orientation of the dipole. In AASM manual the recommended EOG derivations are E1-M2 (E1 is placed 1 cm below the left outer canthus) and E2-M2 (E2 is placed 1 cm above the right outer canthus).

Chin electromyogram (EMG) activity is monitored to record submental muscle tone, which is significantly decreased during REM sleep. According to AASM manual three electrodes should be placed on the chin, one at the midline and one each side below the inferior edge of the mandible. In patients with bruxism (tooth grinding), an additional electrode can be placed over the masseter muscle.

Sleep stages are classified with these EEG, EOG and EMG parameters as NREM and REM sleep. Depending on the scoring system NREM sleep is divided into either four or three classes (S1-4 or N1-3).

2.2.2 Standard monitoring of nocturnal breathing

The gold standard for assessment of airflow is pneumotachography. However, it is not suitable for routine polysomnography because it requires the use of full face mask in the measurement. Therefore several alternative user-friendly devices for the measurement of airflow have been developed.

2.2.2.1 Thermistor

A thermistor or a thermocouple device between the nose and mouth have commonly been used to monitor airflow to detect changes in temperature (e.g. cool air during inspiration and warm air during expiration). A thermistor consisting of wires registers changes in electrical resistance, and thermocouples consisting of dissimilar metals register changes in voltage that result from temperature variations. It is now recognized that while thermistry may be a fairly reliable method to detect complete airflow cessation (i.e., apneas), these sensors do not provide quantitative measures of airflow for detection of hypopneas or nonapneic flow limitation events (Berg et al., 1997, Farre et al., 1998, Redline et al., 2007).

2.2.2.2 Respiratory inductance plethysmography

Also respiratory inductance plethysmography (RIP) has been used for breath amplitude/airflow evaluation. It derives breath volume from changes in self-inductance of wire coils excited by an oscillator circuit placed around the rib cage and abdomen. RIP calibrated to a known volume measure has been considered appropriate for obtaining both qualitative and quantitative indices of breath volume, including identification of the time components of the respiratory cycle (Cantineau et al., 1992). Calibrated RIP provides semi-quantitative information on ventilation and provides hypopnea detection that agrees well with gold standard assessments (Carry et al., 1997, Berg et al., 1997). However, the number of studies that have addressed the validity of RIP is small, and there may be lower levels of accuracy for

uncalibrated RIP, which is the more commonly used technique (Heitman et al., 2002).

2.2.2.3   Nasal pressure transducers

The deficiencies of thermistors have posed a need for alternative physiological sensors that allow a more accurate assessment of airflow limitation. An alternative is the use of nasal pressure transducers, adapted to detect flow limitation. The validation studies of nasal transducers have included a comparison with pneumotachographs; esophageal, gastric, or pharyngeal catheters attached to manometers; and thermistors (Berg et al., 1997, Farre et al., 1998, Akre et al., 2000, Watanabe et al., 2000). Two high-class studies have reported excellent levels of agreement between the total number of apneas and hypopneas scored using the gold standard compared to the nasal pressure signal (Thurnheer et al., 2001, Heitman et al., 2002). Neither study showed large differences when the nasal pressure signal was transformed to produce a more linear signal. The event detection by nasal cannula pressure transducer has also shown to be good compared to esophageal manometry which is considered a gold standard sensor for detecting elevated effort associated with increased upper airway resistance. The nasal pressure signal has not only been used to identify changes in the size of each breath (flow amplitude), but also to provide information on airflow limitation, defined by changes in the pressure signal contour. Development of a plateau in the nasal pressure signal has been shown to correlate with changes on esophageal or supraglottic catheter recordings, suggesting that changes in contour of the nasal pressure signal may be used to detect increased airway resistance (Hosselet et al., 1998). Also the area index derived from the measurements of pressure at the nares has been found to be a sensitive measure of flow rate and flow limitation (Clark et al., 1998).

These studies suggest that nasal transducers may improve hypopnea detection, and often provide data that agree with the gold standard pneumotachograph. All the studies have shown that more events are identified by nasal transducers than are detected by either thermistors or inductance plethysmography. Additionally nasal

prongs provide information of snoring and short and prolonged periods of flow limitation (Hernandez et al., 2001). The short type is related to UARS and RERA but the prolonged type is insufficiently studied and its significance is poorly understood.

2.2.2.4   *Esophageal pressure measurement*

The reference standard for measuring respiratory effort is continuous overnight monitoring of esophageal pressure (Guilleminault et al., 1993, Exar and Collop, 1999). Esophageal manometry has been found to be a reliable tool for excluding OSA and identification of severe OSA (Reda et al., 2001). It has been found suitable for ambulatory use alone or with additional measurements (Tvinnereim et al., 1995, Overland et al., 2005). Subjective tolerance of the esophageal pressure recording with new micro-tip type catheters have been high (93-96%) (Berg et al., 1995, Overland et al., 2005). Studies on potential effects on sleep quality using a naso-esophageal catheter have found minimal changes in sleep architecture or sleep quality in patients with sleep-related breathing disorders (Skatvedt et al., 1996, Chervin and Aldrich, 1997).

2.2.2.5   *Evaluation of respiratory gas balance during sleep*

2.2.2.5.1   Peripheral oxygen saturation

Finger or earlobe pulse oximetry is routinely used to monitor arterial oxygen saturation (SaO_2) or arterial oxyhemoglobin saturation, which reflects the percentage of haemoglobin that is oxygenated. The difference in light absorption between oxyhemoglobin and deoxyhemoglobin determines oxygen saturation. Continuous monitoring of SaO_2 is crucial because it provides important information about the severity of respiratory dysfunction.

2.2.2.5.2   Transcutaneous and end-tidal CO_2 monitoring

The gold standard technique for the measurement of pressure of arterial carbon dioxide ($PaCO_2$) is represented by direct analysis of arterial blood gases, but it is invasive, intermittent and may be unpleasant. The measurement of transcutaneous carbon dioxide ($TcCO_2$) and end-tidal carbon dioxide ($ETCO_2$) represent alternative methods for non-invasive estimation of $PaCO_2$ during sleep (Sanders et al., 1994).

The measurement of CO_2 on human skin surfaces was first described in 1960 by Severinghaus (Severinghaus, 1960). The first commercially available $TcCO_2$ sensors were introduced in 1980, and since then they have been continuously improved but are still using warming to arterialize the cutaneous tissue. Transcutaneous measurement of CO_2 is based on the fact that CO_2 gas diffuses through body tissue and skin and can be detected by a sensor at the skin surface. By warming the sensor, a local hyperemia is induced, which increases the supply of arterial blood to the dermal capillary bed below the sensor. In general, this value correlates well with the corresponding $PaCO_2$ value and gives a more accurate estimation of arterial CO_2 partial pressure than does $ETCO_2$ monitoring (Casati et al., 2006). $TcCO_2$ monitors changes accurately in $PaCO_2$ also during sleep (Senn et al., 2005, Kirk et al., 2006). It has also been shown that with an electrode temperature of 43°C, 8-h continuous monitoring of $TcCO_2$ was well tolerated, and lowering the electrode temperature did not decrease performance for CO_2 monitoring (Janssens et al., 2001). Situations in which $TcCO_2$ monitoring may not be interpretable include patients with perfusion problems, hypothermia, skin diseases, edema, or hypovolemia.

2.2.2.6   *Scoring of respiratory events*

Three types of sleep apnea have been described based on the analyses of breathing patterns: upper airway obstructive, central, and mixed apneas. Cessation of airflow without respiratory effort defines central apnea, during which both diaphragmatic and intercostal muscle activities, as well as air exchange through the nose or mouth,

are absent. During obstructive sleep apnea, a cessation of airflow through nose or mouth with persistence of diaphragmatic and intercostals muscle activities occur. An initial cessation of airflow with no respiratory effort (central apnea), followed by a period of upper airway OSA constitutes mixed apnea.

Central hypopnea during sleep is defined as a decrease of airflow at the nose and mouth along with decreased respiratory effort, causing a reduction of tidal volume and amplitude of the oronasal thermistor or nasal pressure signal. Airflow is decreased also during obstructive hypopnea even if the respiratory effort is maintained or increased (Iber et al., 2007).

The AASM has recently defined the respiratory rules in the AASM manual for the scoring of sleep and associated events (Iber et al., 2007). According to the AASM respiratory rules absence of airflow during apneas has to be detected with oronasal thermal sensor. Instead hypopneas should be recorded with a nasal air pressure transducer with or without square root transformation. For detection of respiratory effort either esophageal manometry or inductance plethysmography are recommended.

In the AASM manual an apnea is defined with a drop in peak thermal sensor excursion by ≥ 90% of baseline in an event lasting at least 10 seconds and at least 90% of the event's duration must meet the amplitude reduction criteria for apnea. For an hypopnea two alternative definitions are presented: a) the nasal pressure signal excursions drop by ≥ 30% of baseline in an event lasting at least 10 seconds with a ≥ 4% desaturation from pre-event baseline; or b) the nasal pressure signal excursions drop by ≥ 50% of baseline in an event lasting at least 10 seconds with a ≥ 3% desaturation from pre-event baseline or the event is associated with arousal. As for apnea at least 90% of the event's duration must meet the amplitude reduction criteria for hypopnea. Respiratory-effort related arousal is defined as a sequence of breaths lasting at least 10 seconds characterized by increasing respiratory effort or flattening of the nasal pressure waveform leading to an arousal from sleep when the sequence of breaths does not meet criteria for an apnea or hypopnea.

The criteria for OSAS in adults defined by the AASM in ICSD-2 are presented in Table 2.

Table 2. The criteria for OSAS according to ICSD-2

A, B and D or C and D fulfil the criteria

A. At least one of the following:
   1) Complaints of unintentional sleep episodes during wakefulness, daytime sleepiness, unrefreshing sleep, fatigue, or insomnia.
   2) Wakings with breath holding, gasping, or chocking.
   3) The bed partner reports loud snoring, breathing interruptions, or both.

B. Findings in PSG:
   1) ≥ 5 scoreable respiratory events (i.e., apneas, hypopneas, or RERAs) per hour of sleep.
   2) Evidence of respiratory effort during all or a portion of each respiratory event (In RERA, the use of esophageal manometry is preferred).

OR alternatively

C. Findings in PSG:
   1) ≥ 15 scoreable respiratory events (i.e., apneas, hypopneas, or RERAs) per hour of sleep.
   2) Evidence of respiratory effort during all or a portion of each respiratory event (In RERA, the use of esophageal manometry is preferred).

D. The disorder is not better explained by another current sleep disorder, medical or neurological disorder, medication use, or substance use disorder.

2.2.3   Evaluating nocturnal movements

2.2.3.1   Anterior tibialis EMG

The gold standard for the quantification of leg movements (LMs) and PLMs is the recording of bilateral surface electromyogram of anterior tibialis muscles performed in the sleep laboratory. In 1982 Coleman originally developed the standard method for recording and scoring PLMs (Coleman, 1982), and a task force of the American Sleep Disorders Association later revised them (The ASDA Atlas Task Force, 1993). A PLM index (number of PLMs per hour of sleep) greater than 5 for the entire night was considered pathologic, although data supporting this feature was very limited. Periodic limb movement disorder (PLMD) was defined as a PLM index of five or greater that is associated with an otherwise unexplained sleep-wake complaint.

In 2005, the new International Classification of Sleep Disorders (ICSD-2) adopted the ASDA definition with 2 changes (American Academy of Sleep Medicine, 2005). The new ICSD recommended a cut-off of PLM index >15 instead of 5 to determine clinical significance. It has been shown that even a high PLM index with associated arousals had little relationship to symptomatic sleep disturbances (e.g., insomnia or hypersomnia). Therefore the ICSD-2 recommended that to use the term periodic limb movement disorder, a relationship must be established between the PLMs and the insomnia and hypersomnia, with no other disorder accounting for the PLMs. Otherwise, the presence of PLMs should simply be noted. For children, the recommendation for clinical significance is a PLM index of >5 with symptomatic sleep disruption.

The instructions for measuring PLM in ICSD are qualitative, as an EMG amplitude increase of at least 25% compared to the EMG biocalibration signal was required to define a movement event (The ASDA Atlas Task Force, 1993, American Academy of Sleep Medicine, 2005). The PLM Task Force of the IRLSSG recommended in 2006 that the baseline EMG values be obtained from the resting tibialis anterior EMG and not from the EMG signal obtained from dorsiflexion of the great toe during biocalibration (Zucconi et al., 2006). Another important change was that the maximum duration of the leg movement was defined as 10 seconds instead of 5 seconds. The new AASM scoring manual (Iber et al., 2007) has adopted these scoring rules for LM and PLM as such (Table 3).

Table 3. The AASM scoring rules for periodic leg movements

A. The definition for a significant leg movement (LM) event:

1) The minimum duration of an LM event is 0.5 seconds.
2) The maximum duration of a LM event is 10 seconds.
3) The minimum amplitude of a LM event is an 8µV-increase in EMG voltage above resting EMG.
4) The timing of the onset of a LM event is defined as the point at which there is an 8 µV-increase in EMG voltage above resting EMG.

B. The definition of a PLM series:

1) The minimum number of consecutive LM events needed to define a PLM series is 4 LMs.
2) The minimum period length between LMs (defined as the time between onsets of consecutive LMs) to include them as part of a PLM series is 5 seconds.
3) The maximum period length between LMs (defined as the time between onsets of consecutive LMs) to include them as part of a PLM series is 90 seconds.
4) Leg movements on 2 different legs separated by less than 5 seconds between movement onsets are counted as a single leg movement.

C. Additional rules:

1) An LM should not be scored if it occurs during a period from 0.5 seconds preceding an apnea or hypopnea to 0.5 seconds following an apnea or hypopnea.
2) An arousal and a PLM should be considered associated with each other when there is <0.5 seconds between the end of one event and the onset of the other event regardless of which is first.

Studies examining night-to-night variability in PLM have all found a significant night-to-night variation in PLM indices within individual subjects, but not in the group level. Hence, multiple nights of recording is recommended for within-subject evaluations (Bliwise et al., 1988, Mosko et al., 1988, Sforza et al., 2005, Hornyak et al., 2005).

2.2.3.2   Actigraphy

Actometry and actigraphy are used as synonyms for recording methods based on accelerometric sensors. The actigraph is an activity monitor, (Sadeh et al., 1995) or a motion detector, designed to record activities during sleep and waking. It is a small device, slightly larger than a wrist watch and is generally attached to the limbs and can be used for recordings lasting for days or even weeks. The actigraph stores the activity data in epoch-by-epoch samples in its internal memory to be downloaded to a computer to process the data graphically and to generate a report of

the sleep-wake pattern. According to the practice parameters of American Academy of Sleep Medicine (Morgenthaler et al., 2007) actigraphy provides an acceptable accurate estimate of sleep patterns in normal, healthy adult populations and inpatients suspected of certain sleep disorders. It is indicated to assist in the evaluation of patients with advanced sleep phase syndrome, delayed sleep phase syndrome and shift work disorder. Additionally, there is some evidence to support the use of actigraphy in the evaluation of patients suspected of jet lag disorder and non-24hr sleep/wake syndrome. In patients with insomnia and hypersomnia, actigraphy can be used in the characterization of circadian rhythms and sleep patterns/disturbances.

Actigraphs have also been used for long in RLS and PLMD studies. The first devices showed lower sensitivity compared to PSG (Kazenwadel et al., 1995) but recent developments in the actometer hardware and the scoring algorithms have improved accuracy in detection of leg movements. Sforza and co-workers have shown that actigraphy is highly reliable in detecting periodic leg movements when compared to PSG (Sforza et al., 2005). However, they state that the poorer performance in patients with sleep-disordered breathing and insomnia may preclude the use of the device in these patients.

2.2.4   Non-conventional diagnostic methods

Besides standard diagnostic methods described above, also other, not so well validated methods are widely used in clinical settings.

2.2.4.1   *Static-charge-sensitive bed (SCSB)*

The SCSB is a movement sensor that was introduced in 1979 by Alihanka and Vaahtoranta (Alihanka and Vaahtoranta, 1979). It is a non-invasive, validated method to monitor breathing, heart beats and gross body movements in bed without any electrodes attached to the subject (Alihanka et al., 1981, Polo, 1992). The SCSB

is placed under a normal foam mattress and the sensor charge is modified by static charge layers, moving in conjunction with body movements, resulting in potential difference. The SCSB has been widely used in sleep-disordered breathing studies in Finland but it has not become common in other countries supposedly because of insufficient validation.

Scoring of respiratory events is possible with the SCSB because breathing disorders during sleep induce alterations in the SCSB signal. These changes in signal characteristics have been divided into 5 categories (P-1, OP-1, OP-2, OP-3 and IRR) representing periodic breathing, hypopneas, apneas or increased respiratory resistance (IRR) (Alihanka, 1987, Polo et al., 1988).

The suitability of the SCSB for evaluation of sleep-related breathing disorders has been evaluated in many studies (Salmi et al., 1989, Svanborg et al., 1990, Polo, 1992, Lojander et al., 1998, Anttalainen, 2008). The SCSB has shown to identify the periodic episodes of obstructive apnea with high sensitivity (Polo et al., 1988). It also displays a distinctive pattern, increased respiratory resistance, which can be prolonged (from 1 min up to 30 min) and considered as an abnormal breathing pattern. Experimentally it has been found to correlate with slowly increasing intrathoracic pressure variations (Polo et al., 1992). IRR is often associated with sustained arterial oxyhemoglobin desaturation and terminated by movement arousal (Polo, 1992). The IRR pattern or high frequency spiking in SCSB signal is related to high-drive breathing and is respiratory in origin and is likely to represent fast components of respiratory movements (Kirjavainen et al., 1996). Kirjavainen also presented in his thesis a study where prolonged high frequency spiking during high-effort snoring with flow limitation was shown to be related to an increase in end-tidal carbon dioxide level (Kirjavainen, 1997).

The SCSB has been used most often in studies with breathing disorders but it has also been used in a large variety of other applications. The SCSB has been shown to perform well in sleep staging in adults (Jansen and Shankar, 1993, Kaartinen et al., 1996) and infants (Erkinjuntti and Kero, 1990). Erkinjuntti and Kirjavainen have used it in pediatric sleep studies (Erkinjuntti, 1987, Kirjavainen, 1997). Polo-Kantola and Saaresranta (Polo-Kantola, 1999, Saaresranta, 2000) have utilized the

SCSB in studies with sleep and hormones. Sjöholm has studied the sleep of teethgrinders with the SCSB (Sjöholm, 1995). The SCSB has also been used in anaesthesiology (Tallila et al., 1996, Nikkola et al., 2004) and in sleep studies in the field of internal medicine (Kantola et al., 1991, Pelttari, 1995). Also epidemiological studies have been conducted with the SCSB (Telakivi et al., 1987). Hyyppä and Kronholm have evaluated sleep quality and nocturnal movement activity with the SCSB in various cohorts (Hyyppä and Kronholm, 1987, Kronholm et al., 1993).

2.2.4.2   Electromechanical film (Emfit)

The Emfit (Emfit Ltd, Finland) is also a movement sensor, which has in many sleep laboratories in Finland replaced the SCSB. The Emfit is a small mattress, which consists of thin, flexible, lightweight biaxially oriented plastic film coated with electrically conductive, permanently polarized layers. Changes in the pressure acting on the film generate a charge on its electrically conductive surfaces and this charge can be measured as a current or voltage signal. When used as a sensor it converts mechanical energy into an electrical signal (Paajanen et al., 2000). Strong direct and converse dynamic piezoelectricity with a d_{33} coefficient of 140pC/N at 600 Hz is identified in the Emfit (Neugschwandtner et al., 2000). The piezoelectric d_{33} coefficient exceeds that of the polyvinylene fluoride (PVDF) by a factor of 5 and compares favourably with ferroelectric ceremics. Emfit has been used in ballistocardiographic studies (Koivistoinen et al., 2004, Alametsa et al., 2008), measuring heart sounds (Karki et al., 2007), heart beat interval (Kortelainen and Virkkala, 2007), respiratory rate (Kim et al., 2007) and ultrasonic air applications (Ealo et al., 2008). Comparisons between Emfit and SCSB signals have produced high similarity in amplitude, correlation and spectral properties (Alametsä et al., 2005). An example of Emfit recording with nasal flow and oximetry is presented in Figure 1.
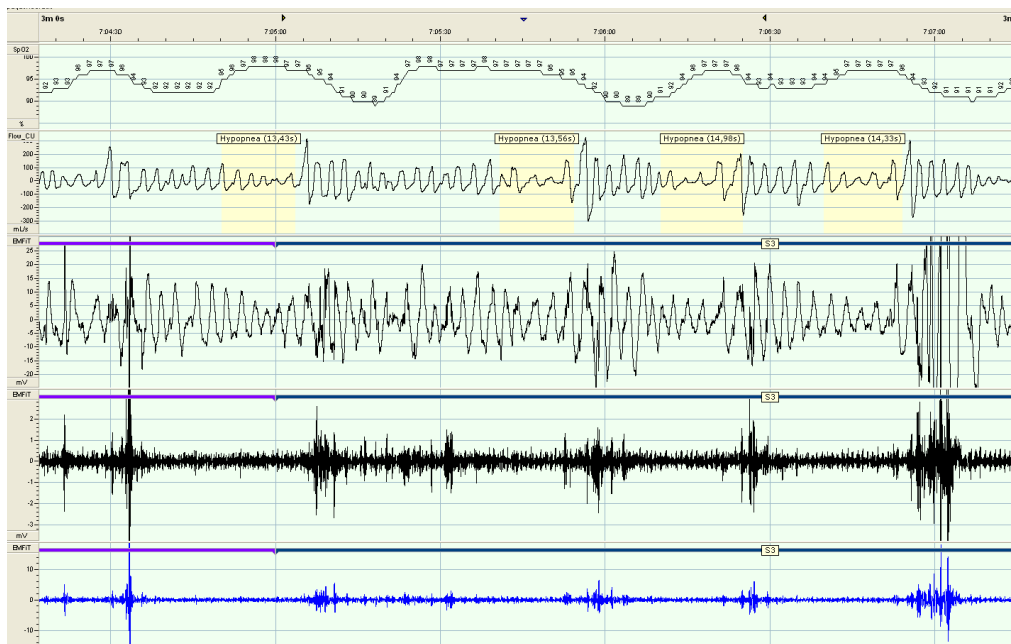
Figure 1. A three-minute episode of a recording presenting hypopneas (trace 2, nasal flow) with desaturations (trace 1, SaO_2) and OP-1 and OP-3 type breathing patterns in the Emfit signals (traces 3,4 and 5).

2.2.4.3   Other movement sensors

Besides Emfit there exist several different measuring principles for pressure sensitive foils: force sensing resistors, capasitive foils and piezoelectric polymer foil PVDF. The PVDF mattress provides a three-channel output for body movements, respiratory movements and ballistocardiogram (Siivola, 1989). Methods with water-filled under-pillow sensor (Chen et al., 2005) and pneumatic mattress (Nino-Murcia et al., 1991) or cushion placed under mattress (Watanabe and Watanabe, 2004) have also been used to measure heart beat, respiration, snoring and body movements. Brink and co-workers have presented a modern version of sensitive bed where four high-resolution force sensors are placed under the bedposts (Brink et al., 2006). Cardiac and respiratory parameters, body position and the subject's movement activity can be calculated from these sensor signals.

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2.2.5 Snoring and tracheal sound

Snoring is a respiratory sound that is generated during sleep when breathing is causing vibration of tissues in the upper airways. Acoustic analysis of snoring sound has been developed to find an accurate, efficient and more objective way to acquire reliable documentation about snoring. Spectral characteristics depend on source properties which in case of snoring can have relation to obstruction level (Beck et al., 1995, Dalmasso and Prota, 1996, Agrawal et al., 2002, Saunders et al., 2004). It is still unclear how the acoustic analysis has to be performed and which parameters have to be investigated before anatomical cause of snoring can be defined (Moerman et al., 2002).

The severity and harmfulness of snoring has mostly been simply associated with its intensity (in dBs) (Lugaresi et al., 1983, Itasaka et al., 1999, Wilson et al., 1999). Analyses of tracheal breathing sounds have long been used in the diagnostics of OSAS. Intensity changes of tracheal breathing sounds have been shown to correlate with apnoea and hypopnoea events (Cummiskey et al., 1982, Peirick and Shepard, Jr., 1983). Apnea events have been shown to be surrounded by significant increase of medium- and high-frequency power, with OSAs showing the largest increase (Kaniusas et al., 2005). Moreover, the tracheal sound analysis have be used to estimate the severity of the OSAS using relative changes of the total tracheal sound power (Nakano et al., 2004). Also intensity levels, (Van Brunt et al., 1997), pitch analysis (Abeyratne et al., 2005), spectral characteristics (Duckitt et al., 2006) and formant frequencies (Ng et al., 2008) of snoring sounds have shown correlation between snoring and OSAS. Snoring density and loudness of snoring have also shown to correlate with increasing OSAS severity (Brietzke and Mair, 2007).

The studies of snoring and snoring sound have mainly concentrated on technical solutions and detection of apneas and hypopneas. However, also flow limitation is at present considered as a significant form of SDB. Very recently the high frequency components of tracheal sound has been shown to be emphasized during prolonged flow limitation (Tenhunen et al., 2009).

3. Aims of the study

The aims of the study were to investigate

1. the suitability of the SCSB for the detection of periodic movements during sleep

2. the usability of the small Emfit sensor in revealing periodic movements

3. the appearance and detection of spiking events and the performance of various algorithms in automatic detection of the spiking phenomenon in the Emfit signal

4. the impact of prolonged spiking and other types of sleep-disordered breathing on respiratory gas exchange

5. the characteristics of various patterns in the compressed tracheal sound analysis and to evaluate their use in sleep-disordered breathing studies

4. Material and methods

4.1 Subjects and recordings

In study I, fifty polysomnographic sleep recordings were carried out in 47 patients (35 men and 12 women) who had been referred to the sleep laboratory of the Department of Physiology in Turku University because of heavy snoring, suspected sleep apnoea or nocturnal restlessness. The mean age of the subject group was 50.3 (SD 11.4) years and BMI 28.5 (SD 5.0) kg/m^2.

In study II, twenty-seven consecutive patients (18 men and 9 women) referred to the sleep laboratory of the Pirkanmaa Hospital District volunteered to participate in this study. The mean age of the subjects was 41.6 (SD 11.2) years and BMI 28.4 (SD 4.7) kg/m^2. The reasons for referrals were a suspicion of restless legs syndrome for five patients, a control of continuous positive airway pressure treatment for four patients and possible sleep-disordered breathing for 18 patients.

In study III, six patients (4 men and 2 women) with nocturnal respiratory disturbances were studied. Their mean age was 46.7 (SD 5.9) years and BMI 30.8 (SD 7.3) kg/m^2. From each recording one episode of frequent Emfit spiking was selected for the analysis.

In study IV, 115 consecutive polysomnograms (from different patients) equipped with TcCO$_2$ measurement, were analyzed retrospectively. Ten recordings had technical shortcomings and were therefore discarded. Twenty-two recordings were children's polysomnograms and were not included in the study. The inclusion

criterion was that the patient had to present at least one over 3 min episode of prolonged spiking with another episode with repetitive hypopneas or obstructive apneas. 32 subjects had no breathing disorders and 32 patients had sleep-disordered breathing not fulfilling the inclusion criteria. Nineteen subjects (14 men and 5 women) passed all the study phases and were accepted into further evaluation. The mean age of the accepted subjects was 48.4 (SD 7.4) years and BMI 30.9 (SD 5.9) kg/m^2.

In study V, thirty-three consecutive patients (27 male, 6 female) referred to the Sleep Unit of Pirkanmaa Hospital District volunteered to participate in this study. The mean age of the subjects was 44.3 (SD 11.4) years and BMI 29.3 (SD 6.4) kg/m^2. One female and one male were referred due to the suspicion of narcolepsy, one male subject had restless legs, and the rest were referred due to the suspicion of sleep-disordered breathing.

4.2 Methods

4.2.1 Recording parameters

4.2.1.1 Polysomnography

Grass Model 79C pen recorder (Grass Instrument Co., Quincy, MA, USA; I) and Embla N7000 with Somnologica software (Medcare, Iceland; II-V) were used as recording systems. The recording montage consisted of one (C4-A1; I) or six (Fp1-A2, Fp2-A1, C3-A2, C4-A1, O1-A2, O2-A1; II-V) EEG derivations, electrooculogram and submental electromyography. The recordings were scored into sleep stages according to the Rechtschaffen and Kales criteria (Rechtschaffen and Kales, 1968). Microarousals were scored according to the criteria of the American Sleep Disorders Association (1992). Electrocardiogram was monitored using standard leads. Nasal airflow was recorded with a nasal pressure transducer and a thermistor (II-V). Thoracic and abdominal belts were used for respiratory

movement detection (II-V). Body position was monitored with an internal three-dimensional actometer (II-V). Blood oxygen saturation and pulse were recorded by finger pulse oximetry (Nonin XPOD®, Nonin Medical Inc, USA). The sampling rate of 1 Hz was used for pulse oximetry and 200 Hz for all other parameters (II-V).

4.2.1.2   Anterior tibialis EMG

Anterior tibialis EMG was recorded with surface electrodes placed at 2-3 cm apart on the belly of the anterior tibialis muscle in both legs. In study I the patients were asked to dorsiflex both of their feet prior to the onset of the recordings to adjust the EMG signal amplitude within the dynamic range of the recorder. In study II the recordings were performed using the World Association of Sleep Medicine (WASM) standards (Zucconi et al., 2006). The sampling rate for EMG in study II was 200 Hz and in study I recordings were performed with an analogous device.

4.2.1.3   Static-charge-sensitive bed

Static-charge-sensitive bed (Biomatt, Biorec, Turku, Finland) was used in study I (Figure 2). The signals were recorded on a Grass Model 79C pen recorder with a paper speed of 1.5 mm/s. The composite movement signal was filtered into two channels: low frequency band (LFB, 0.25-0.9 Hz), representing the respiratory movements and high frequency band (HFB, 6-16 Hz), displaying the ballistocardiogram. Body and limb movements were distinguished as high amplitude deflections displayed on either channel. An 4.5 min epoch was defined as periodic movement activity (PMA) -positive, if there was a series of at least four consecutive movements in the epoch that fulfilled the criteria for duration (0.5 - 5 s) and the movement interval (5 - 90 s).

4.2.1.4   *Electromechanical film*

The Emfit sensor was used in studies II, III and IV. The Emfit sensor with dimensions of 32 x 62 cm was placed under the thoracic area of the sleeping patient below the sheet (Figure 2). The sampling rate for Emfit was 200 Hz. The Emfit signal was amplified and filtered into 6 − 16 Hz band (Kirjavainen et al., 1996, Alametsa et al., 2006). In addition in study III also the 3−16 Hz band was used for comparison purposes.
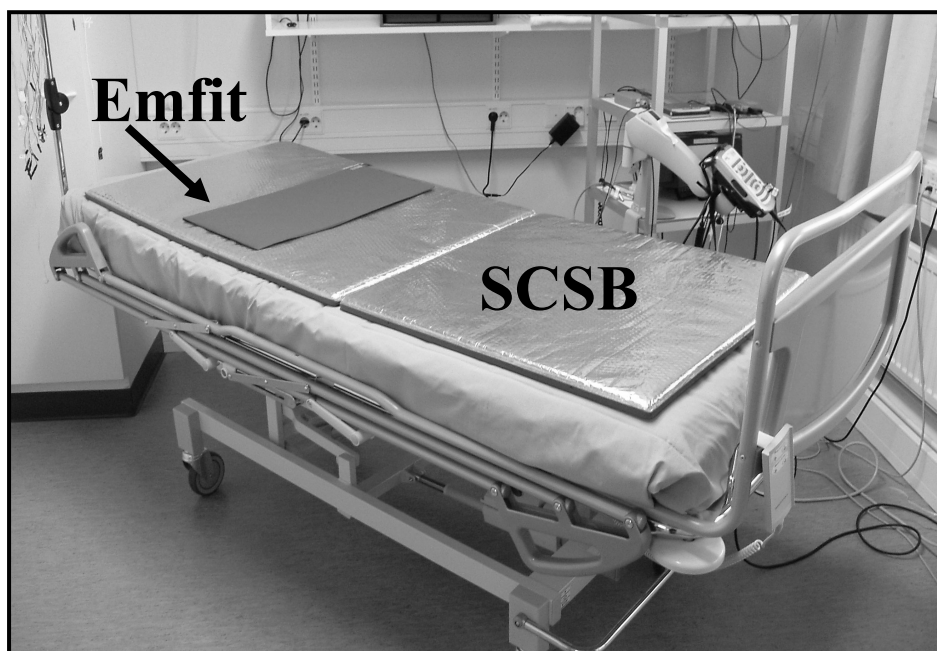
Figure 2. In the schematic picture the Emfit sheet is placed on the SCSB. In reality the SCSB is placed under the mattress and the Emfit below the sheet. Note the difference in size and thickness of the devices.

4.2.1.5   *Monitoring of carbon dioxide*

In study IV continuous transcutaneous measurement of carbon dioxide was performed with Tina TCM4 (Radiometer, Denmark) with the sampling rate of 1 Hz. The electrode was placed on the sternum and the temperature was set on 43 degrees Celsius.

4.2.1.6   Tracheal sound recording

The tracheal sound recordings were performed with a small electret microphone, Panasonic WM-60A (Matsushita Electric Industrial Co., Ltd, Kadoma Osaka, Japan). The microphone has conical air cavity of 25 mm in diameter and 3 mm deep. The sensitivity of the microphone is 10 mV/Pa and the frequency range in the free field is 20 Hz - 20 kHz, ±2 dB (Sovijarvi et al., 1998). The microphone is attached to the suprasternal fossa with an adhesive tape ring and additional taping on the top (Figure 3). The measured breathing sound signal is amplified and high-pass filtered with a preamplifier unit. After the preamplifier unit the signal is fed into an external sound card USB Sound Blaster Audigy 2 NX (Creative Labs, Singapore) for A/D conversion followed by USI-01 USB isolator (MESO, Mittweida, Germany) providing galvanic isolation between the measuring device HeLSA (Helsinki Lung Sound Analyzer) and the PC. The sampling frequency for the sound signal is 11025 Hz. SuperHeLSA software (Pulmer, Helsinki, Finland) provides the raw data from the sounds recorded over the trachea, which is converted into Embla data format.

Figure 3. The microphone is attached to the jugulum in the tracheal sound recording.

4.2.2 Movement detection and protocols in studies I and II

In study I in the first setup 500 samples, ten from each recording, were randomly picked up for the analyses. Each sample contained 4.5 minutes of recording which, according to the PLM definitions, would allow four repetitive movements to occur at three intervals of 90 seconds. The right and the left anterior tibialis EMG channels and the two SCSB channels (the LFB and HFB) were separated from the rest of the recording, and cut apart from each other. The EMG and SCSB extracts were randomly ordered and separately analysed by two scorers (ER, ME). Seven samples were eliminated from the analyses because of defects in the EMG channels. The frequency of PLM was determined from the EMG recordings according to the criteria proposed in the International Classification of Sleep Disorders (Diagnostic Classification Steering Committee, 1990). A 4.5 min epoch was defined as PLM-positive, if it contained a series of at least four consecutive EMG events that fulfilled the criteria for the duration (0.5 - 5 s) and the movement interval (5 - 90 s). Analogously in scoring SCSB a 4.5 min epoch was defined as periodic movement activity (PMA) -positive, if there was a series of at least four consecutive movements in the epoch that fulfilled the same criteria for the duration and movement interval.

In the second setup continuous all-night recordings from seven men and three women with periodic movement activity exceeding 15 % of time in bed were analysed. Two independent scorers (ER, ME) analysed in different sessions the EMG and the SCSB curves which were cut apart. The average length of the recordings was 424 (range 253 – 523) minutes. The analysis epoch length was one page equal 200 seconds at the paper speed of 1.5 mm/s. In addition to epoch-based analyses, the periodic movements were also analysed on a movement-by-movement basis with the same criteria as in first setup.

In study II the scorings of EMG were performed using the WASM standards (Zucconi et al., 2006). Periodic movements in EMG and Emfit channels were scored blindly and independently by two neurophysiologists (ER, SLH) using the time window of 4 min/screen. PLMs were defined as repetitive muscle jerks lasting from

0.5 to 10 seconds, separated by an interval ranging from 5 to 90 seconds, with an EMG amplitude rise of 8 µV above the resting level, organized in series of 4 or more consecutive leg movements. After the independent movement scorings the two neurophysiologists together formed the EMG consensus scorings, which were used in further analyses.

The WASM rules and the leg activity monitoring standards in the WASM guidelines and were also used in the Emfit scorings. The threshold of movement detection was set as double the basic Emfit signal amplitude, which seems to correspond well with the anterior tibialis EMG amplitude increase of ≥ 8 µV. The Emfit consensus scorings analogously with EMG were used in further analyses.

For the comparison between EMG and Emfit the movements in the right and left EMG channels were combined and bilateral movements were counted as one. The movements occurring on the two sides were considered bilateral when they were overlapping or separated by less than 0.5 s (Zucconi et al., 2006). The total number of single and periodic movements were tabulated and the periodic limb movement index (PLMI) for EMG and the periodic movement index (PMI) for Emfit were counted. These indexes were divided into three severity categories according to ICSD (Diagnostic Classification Steering Committee, 1990); mild: $5 \leq$ PLMI / PMI < 25, moderate: $25 \leq$ PLMI / PMI < 50, severe: PLMI / PMI ≥ 50 and results gathered by various methods were compared. Additionally, a software was developed for comparing the selections of single and periodic movements within Emfit and EMG channels.

4.2.3 Scoring of respiratory events

In study I in addition to movement scoring also respiratory abnormalities in each 4.5 min epoch were determined from the SCSB channels. Normal breathing was distinguished from four types of periodic breathing (P-1, OP-1, OP-2, OP-3) or prolonged spiking episodes. This was performed according to previously described respiratory scoring principles (Polo et al., 1988). If the epoch contained one or more

periods of continuous motor activity exceeding 40 seconds, it was scored as motor active wakefulness.

The apnea-hypopnea index (AHI) in studies II-V was calculated from the nasal pressure channel as the number of cessations or diminutions $\geq$ 50% of airflow lasting over 10 s per hour of sleep. In study V for hypopneas new scoring rules were used (Iber et al., 2007), hypopnea rule 4B).

In studies IV and V the inspiratory flow shapes in selected episodes were visually classified on breath-by-breath basis according to principles earlier described (Aittokallio et al., 2001b). The rounded inspiratory flow shapes (without flow limitation) were distinguished from the non-rounded (or flow-limited) ones by two classifiers. Breathing was defined as normal if more than 80% of inspiratory flow shapes were round. The share of round flow shapes was also calculated for prolonged spiking, hypopnea and apnea episodes. The agreements for flow shape scoring were 96% (89-98%, IV) and 95% (87-98%, V).

In study IV to study the long-term transition of $TcCO_2$, SaO_2 and pulse rate signals in collected breathing pattern sections, the episodes were divided into 50 parts. Median values of every part were used to define a trend line and to calculate the change per minute values for $TcCO_2$, SaO_2 and pulse rate. In addition, the initial values for $TcCO_2$, SaO_2 and pulse rate signals in the beginning of the breathing episodes were collected.

In study V breathing sounds from the neck were measured by a microphone using 11025 Hz sampling rate. To reduce the sound signal data only the maximum and minimum sound signal values of each consecutive 15 s epochs were taken. This compressed sound information (four samples per minute) was used for the actual analysis. Simultaneous visualization of the two compressed sound traces through time revealed three dominating patterns: a plain signal curve close to zero, periods with a thin signal curve deviating clearly from zero and periods with a thick signal curve (see Figure 14). Additionally, the fourth pattern, where the line was at zero, was present in two recordings. Further examination of these recordings revealed that

the microphones had become detached during the night. Therefore these recordings were excluded from the data set.

All the compressed tracheal sound traces during the sleep period were manually analysed by two different scorers (ER, SLH) and divided into plain, thin and thick periods. Altogether 780 different pattern periods were scored in the recordings. The agreement between the classifiers was 86.3%. The two individual scorers together classified consensus scorings based on the individual scorings. The consensus scorings were used in the analyses.

After that one representative episode of plain, thin and thick tracheal sound periods (later pure plain, pure thin and pure thick) lasting 10 minutes were selected from each subject, whenever present. AHIs during these pure periods were calculated.

To examine the connection between AHI and the total nocturnal amount of different sound patterns, percentages of plain, thin and thick periods during the night for each subject were computed.

4.2.4 Spiking detection

In study III one episode of prolonged spiking from each of the six recordings was selected for the analysis. These episodes lasted 632 – 1575 s. The total duration of the episodes was 5993 s. From these episodes individual spikes were visually selected by a clinical neurophysiologist (ER). The total number of visually scored spiking events was 1503.

The developed feature quantifies relative individual spike amplitude changes using one-minute history window as baseline data. Features were calculated with 1 s time resolution.

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Two separate fixed-length windowing approaches were applied. In the first approach, non-overlapping 1 s Hanning windowed BCG signal segments were collected. In the second approach, shorter Hanning windows, in which five partly overlapping data segments (length of 0.5 s of each) centered at 0.1, 0.3, 0.5, 0.7 and 0.9 s were extracted from a 1.3 s signal segment. To compute feature values, amplitude spectra of these windows were extracted. Altogether 10 feature versions were developed using different variations of change features and frequency ranges of 3 - 16 and 6 - 16 Hz.

In the performance evaluation a true positive finding was calculated if the detected spiking event occurred simultaneously with a visually scored spiking event. If a spiking event was detected and there was no visual scoring of spiking, a false positive finding was counted. In the second approach no false positive findings were calculated one second prior and after each event. This was done because a scored spiking event near second border could be detected also in adjacent second. In addition after one false finding there was 1 s latency time during which no additional false positive findings were counted.

4.2.5   Statistical analysis

In studies I, II and III the sensitivities, specificities and the positive and negative predictive values were computed using standard formulas.

In studies II-V non-parametric tests were used, as all the variables were not normally distributed. P-values < 0.05 were considered statistically significant.

In study II the Spearman's correlation coefficient was computed to assess the relationship between the Emfit data and the EMG results. Based on the PLMI and PMI levels of 5 and 15 movements per hour, receiver operating characteristic (ROC) curves were derived and an area under the curve (AUC) was calculated. The Bland-Altman method of concordance was used to assess potential range-dependent agreement.

In study III to evaluate the performance of the various spiking detector versions ROC curves were determined, depicting the true positive rate as a function of the false positive rate. The ROC curve was determined by using a range of threshold values for each feature version.

In studies IV and V statistical analyses were performed with SPSS for Windows version 12.0® (SPSS Inc.).

In study IV multiple comparisons of independent variables were performed by the Kruskal Wallis test with the post hoc analysis by the Mann-Whitney U test. In all the post hoc analyses the Bonferroni correction factor was used. Spearman rank correlations were calculated to measure the strength of associations between the durations of the breathing episodes and the amount of $TcCO_2$ changes. Also the correlations between the changes of $TcCO_2$ and the durations of individual hypopneas and apneas inside hypopnea and apnea episodes were defined.

In study V the Friedman test was used in comparison between the dependent variables with the Wilcoxon signed-rank test in the post hoc analyses. In these post hoc analyses the Bonferroni correction factor was used to correct for multiple comparisons. The relationship between AHI and the percentage of thick, thin and plain periods was studied in two ways. First nonparametric correlation coefficients were computed. Secondly, a linear regression model between the ranks of both AHI and the thick period percentages was formed. To verify the model each patient's AHI based on the patient's thick period percentage was predicted.

4.2.6 Ethical aspects

Studies II, IV and V were approved by the Ethical Committee of the Pirkanmaa Hospital District. Study III was approved by the Medical Director of the Pirkanmaa Hospital District. All the subjects in these studies gave their written consent. Study I

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was performed retrospectively without the need of approval of the Ethical Committee.

5. Results

5.1 Performance of mattress movement sensors in detecting periodic limb movements (Studies I and II)

The demographic data and sleep parameters of the subjects of the studies I and II are presented in Table 4. The sleep parameters in study I are based the SCSB scorings and in study II polysomnographic results are presented.

Table 4. Demographic data and sleep parameters of the 47 (study I) and 27 (study II) subjects

	Study I			Study II	
	Mean	Range		Mean	Range
Age (years)	50.3	33 - 70		41.6	21 - 58
BMI (kg*m^{-2})	28.5	18.4 - 38.7		28.4	19.0 - 38.1
TIB (min)	401	228 - 513		475	286 - 598
P-1 (%)	3.5	0 - 31.8	TST (min)	408	228 - 531
OP-1 (%)	6.3	0 - 27.5	SEI (min)	85.7	66.3 - 98.6
OP-2 (%)	5.2	0 - 45.6	WASO (min)	56.1	6.5 - 163
OP-3 (%)	2.1	0 - 55.3	SL (min)	11.5	0.5 - 40
Spiking (%)	6.2	0 - 54.1	REMlat (min)	145.0	12.5 - 334.5
Wake (%)	7.3	0 - 32.4	S1 (%)	9.6	2.5 - 22.8
PM (%)	26.6	0 - 65.2	S2 (%)	64.1	45.5 - 79.0
			S3 (%)	8.4	0.0 - 24.4
			S4 (%)	1.2	0.0 - 10.3
			SREM (%)	16.8	4.4 - 26.2
			ARI (no/h)	16.2	4.2 - 38.7
			AHI (no/h)	12.0	0.1 - 59.0
			SaO$_2$min (%)	88	68 - 93
			ODI4 (no/h)	6.0	0.0 - 46.0

BMI, Body mass index; TIB, Time in bed; P-1 - OP-3, percentage of periodic breathing patterns referred to TIB; Spiking, percentage of spiking pattern referred to TIB; Wake, percentage of wake time referred to TIB; PM, percentage of periodic movement time referred to TIB; TST, Total sleep time; SEI, Sleep efficiency index; WASO, Wakefulness after sleep onset; SL, Sleep latency; REMlat, latency to the first REM-sleep period; S1 - SREM, percentage of the sleep stage referred to TST; ARI, arousal index; SaO2min, minimun oxygen saturation; ODI4, oxygen desaturation index.

5.1.1 Static-charge-sensitive bed

In study I two independent scorers divided 4.5 min length SCSB and anterior tibialis signal epochs into PLM positive and PLM negative categories. In addition to movement scoring also epochs with respiratory abnormalities and motor active wakefulness were determined. The number of positive anterior tibialis PLM and SCSB PMA epochs appearing as isolated, in conjunction with respiratory abnormalities or during motor active wakefulness combined are presented in Table 5.

Table 5. The number of all positive PLM and PMA epochs

	Total epochs	EMG		SCSB	
		PLM	Detected with SCSB	PMA	Detected with EMG
Scorer I	493	172	145	179	143
Scorer II	493	193	157	185	156

The sensitivity of the SCSB to detect epoch with PLM was 0.84 and 0.81 depending on the scorer. The specificities were 0.89 and 0.84, respectively. The results were better when simultaneous respiratory abnormalities or motor active wakefulness were excluded. All results are presented in Table 6.

The inter-rater reliability was 0.92 in scoring the EMG and 0.91 in scoring the SCSB. Forty per cent of the disagreements occurred in epochs with respiratory abnormalities and in epochs with motor active wakefulness. Eighteen per cent occurred in epochs containing only 4-5 movements as 4 movements was critical by definition.

Table 6. Sensitivities, specificities, positive and negative predictive values of the SCSB to detect epoch with PLM with and without simultaneous respiratory abnormalities (RA) or motor active wakefulness (MAW)

	Scorer I		Scorer II	
	All epochs	RA and MAW epochs excluded	All epochs	RA and MAW epochs excluded
Sensitivity	0.84	0.90	0.81	0.91
Specificity	0.89	0.91	0.84	0.94
Positive predictive value	0.80		0.84	
Negative predictive value	0.92		0.89	

The performance of the SCSB was also analysed in function of the frequency of PLM in each epoch. The highest sensitivity (0.98-0.99) was observed with 11-12 movements in the 4.5 minute epoch, which corresponds to the average movement interval of 24-27 seconds.

In study I also continuous recordings were analysed on the epoch-by-epoch as well as movement-by-movement basis. In the epoch-by-epoch analysis, the mean amount of periodic movement activity was 47.1% (range 21.9-80.0%) of time in bed. The inter-rater reliability analysing the EMG was 0.93 (range 0.86-1.00) and 0.91 (range 0.83-0.96) with the SCSB. The sensitivities to detect epoch with PLM were 0.93 / 0.94 and specificities 0.87 / 0.89. Thus the performance of the SCSB to detect periodic movements was better in continuous epochs than in random epochs (Table 7).

Table 7. The performance of the SCSB to detect periodic movements in epoch-by-epoch analysis in continuous recordings

	Scorer I		Scorer II	
	Mean	Range	Mean	Range
Sensitivity	0.93	0.85-1.00	0.94	0.84-1.00
Specificity	0.87	0.62-0.96	0.89	0.73-1.00
Positive predictive value	0.86	0.66-0.94	0.87	0.74-1.00
Negative predictive value	0.94	0.85-1.00	0.96	0.86-1.00

In the movement-by-movement comparison the mean sensitivities were 0.81 (range 0.63-0.93; scorer 1) and 0.80 (range 0.56-0.93; scorer 2). The mean PLM vs. PMA indexes were 65 vs. 63 or 55 vs. 53 movements per hour of time in bed,

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depending on the scorer (Figure 4). The SCSB did not fail to reveal periodic movement activity in any patient studied. As in the epoch based analysis, the presence of respiratory abnormalities and motor active wakefulness were the major source of discrepancy between EMG and SCSB. Since periodic movement activity in the upper limbs or trunk was not detected with our reference, bilateral anterior tibialis EMG, some 'false positive' epochs were observed with the SCSB, which reduced the calculated specificity.
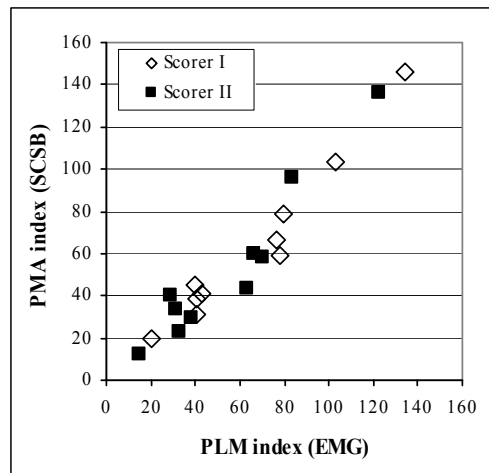

Figure 4. The relationship between periodic movement activity (PMA) and periodic limb movement (PLM) indexes in independent scorers.

5.1.2 Emfit

The study II included 27 subjects. Four subjects had completely normal findings. One subject had catathrenia. Fourteen subjects had sleep-disordered breathing. Three of them had abundant periodic limb movements. Four subjects had only periodic limb movements and four subjects were under CPAP treatment.

Number of single and periodic movements with indexes are presented in Table 8.

Table 8. Amounts of single and periodic movements and corresponding indexes

		EMG		Emfit	
		Mean	Range	Mean	Range
In sleep	Single (limb) movements (number)	222	38-742	193	32-666
	Periodic (limb) movements (number)	172	4-717	136	0-632
	Periodic movement index 1/h	27.6	0.5-108.8	23.0	0.0-101.1
With wake	Single (limb) movements (number)	290	55-801	233	56-714
	Periodic (limb) movements (number)	229	17-771	165	0-668
	Periodic movement index 1/h	31.4	2.6-99.8	23.8	0.0-111.4

The Spearman's correlation coefficient between the PMI of Emfit and the PLMI of EMG during sleep was 0.87 (Figure 5) and 0.88 with wake included.
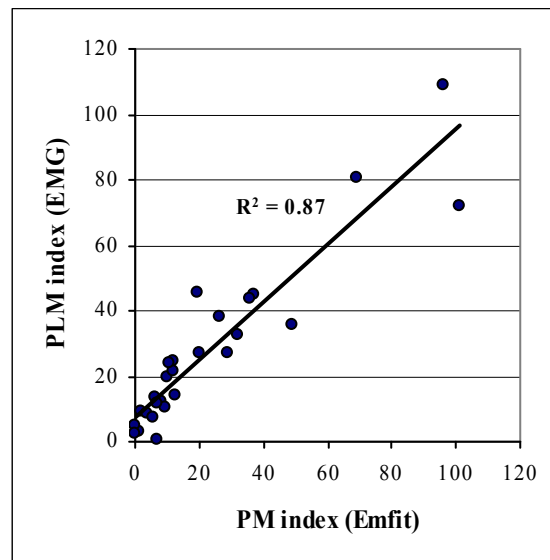

Figure 5. The relationship between the PLM index (EMG) and the PM index (Emfit) during sleep.

The sensitivity of the Emfit sensor to detect periodic limb movements in sleep was 0.91 at the cut-off level of 5 movements/h and 0.73 when the cut-off level was 15 movements/h. The specificities were 0.75 and 1.00, respectively. The overall performance of Emfit to detect periodic movements at the cut-off levels of 5 and 15 movements/h is presented in Table 9.

Table 9. The performance of Emfit to detect periodic movements at the cut-off levels of 5 and 15 movements/h

	In sleep		Wake included	
	5 mov/h	15 mov/h	5 mov/h	15 mov/h
Sensitivity	0.91	0.73	0.88	0.65
Specificity	0.75	1.00	0.67	1.00
Positive predictive value	0.95	1.00	0.95	1.00
Negative predictive value	0.60	0.75	0.40	0.63

The diagnostic capability of the Emfit sensor to detect periodic limb movements at the thresholds of a) 5 movements/h and b) 15 movements/h of the PLM index is presented with a ROC curve in the Figure 6. The area under curve (AUC) was 0.96 and 0.98, respectively.
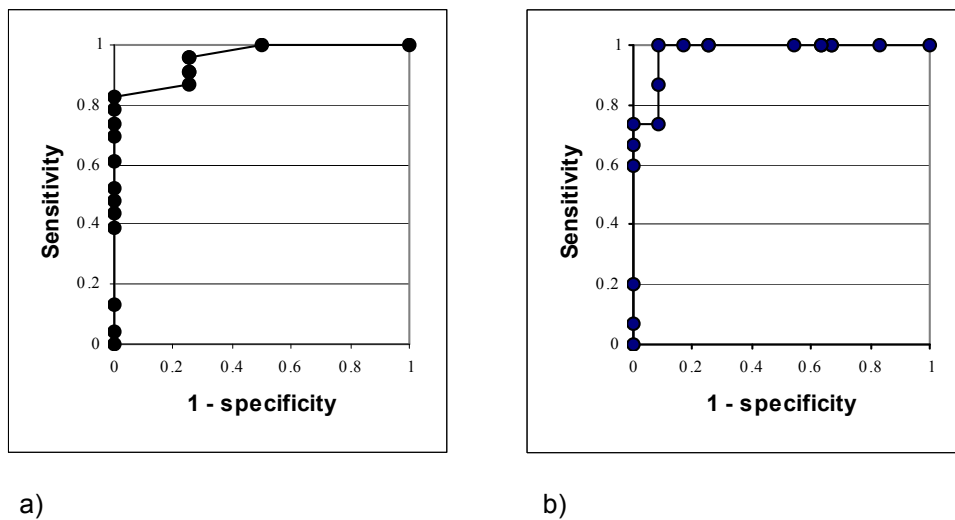

a)   b)

Figure 6. The ROC curves at the thresholds of a) 5 movements/h and b) 15 movements/h of PLM indexes based on EMG versus Emfit results.

According to the Bland-Altman method Emfit seems to underestimate motor events compared to combined EMG (mean PLMI / PMI difference of 4.5 movements/h, standard deviation of 9.8 movements/h (Figure 7).
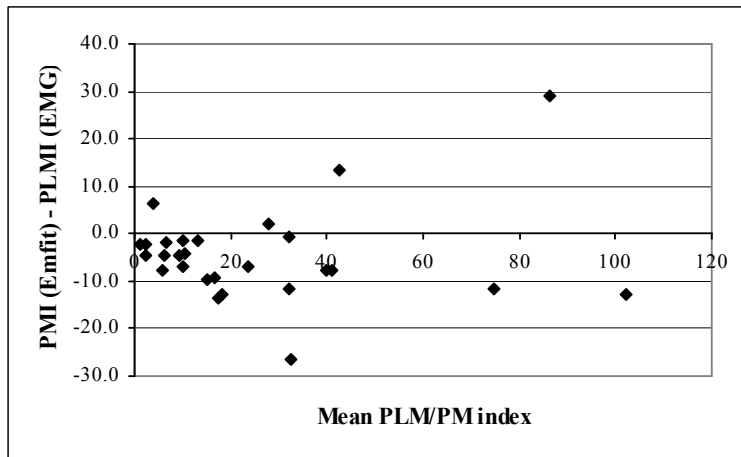
Figure 7. The distribution of the differences between PM index (Emfit) and PLM index (EMG) against the mean of PLM and PM indexes.

Based on the ICSD classification periodic movement indexes were divided into three severity categories. In 22 out of 27 cases the severity categories with Emfit and EMG were the same (Figure 8). In four cases Emfit suggested a milder and in one case a more severe disorder.
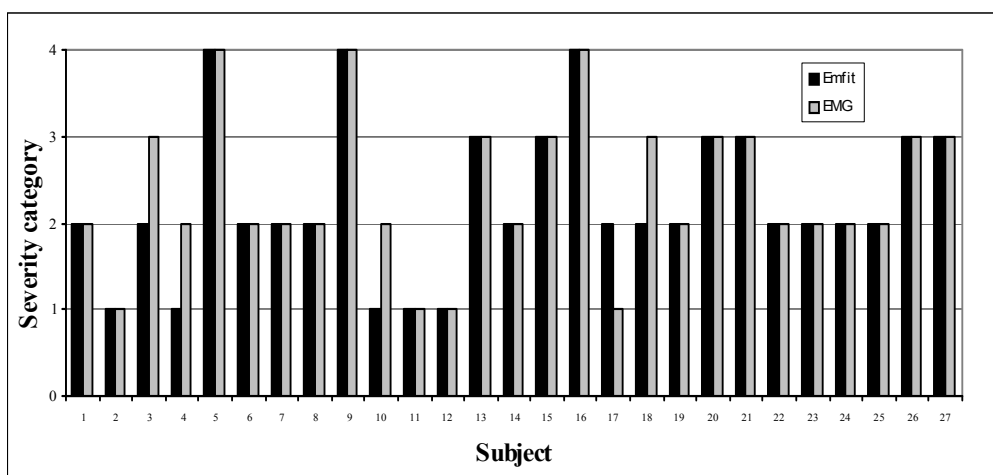

Figure 8. The severity categories of PM (Emfit) and PLM (EMG) indexes of the subjects. Categories: (1) PLM/PM index < 5; (2) 5 ≤ PLM/PM index < 25; (3) 25 ≤ PLM/PM index < 50; (4) PLM/PM index ≥ 50.

5.2 Automatic analysis of Emfit signal (Study III)

In the automatic detection of Emfit spiking events the features based on the frequency range 6–16 Hz provided overall better results than corresponding versions with 3–16 Hz. Method 10 provided the best performance of spiking detection. The ROC curve with true and false positive rates of 80% and 19% of the method 10 is presented in Figure 9.
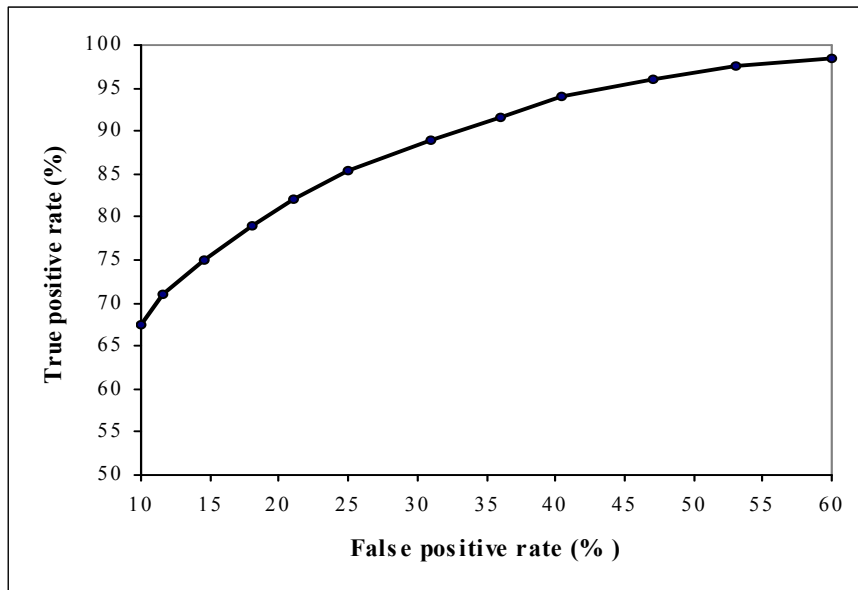

Figure 9. The comparison of spiking detection outcome of the method 10 presented with a ROC curve.

5.3 Sleep-disordered breathing (Studies IV and V)

The demographic data of the subjects and sleep parameters derived from the sleep recordings of studies IV and V are shown in Table 10.

Table 10. Demographic data and sleep parameters of the 27 (study IV) and 31 (study V) subjects

	Study IV		Study V	
	Median	Range	Median	Range
Age (years)	46	37 - 59	46	16 - 68
BMI (kg/m^2)	30.0	21.8 - 46.1	27.8	21.8 - 44.6
TIB (min)	488.0	364 - 602.5	488	328.5 - 602.5
TST (min)	379.0	233 - 529	396.5	222.5 - 529
SEI (min)	83.0	53 - 97	82.6	65.5 - 98
WASO (min)	72	13 - 199.5	59	6 - 154.5
SL (min)	10.5	2.5 - 40.5	9.8	0 - 53.5
REMlat (min)	118.0	79.5 - 441	124.8	1.5 - 514
S1 (%)	13.8	0.1 - 39.6	10.9	1.9 - 40.6
S2 (%)	62.4	48.5 - 86.8	67.6	49.5 - 85.6
S3 (%)	5.6	0 - 24.3	5.2	0 - 17.7
S4 (%)	0	0 - 2.8	0	0 - 21.3
SREM (%)	14.1	3.7 - 22.2	13.7	0.4 - 25.5
ARI (no/h)	32	11.8 - 68.7	44	11 - 108
AHI (no/h)	22	9 - 62	24.5	0 - 92
SaO$_2$min (%)	84	66 - 91	86	60 - 96
ODI4 (no/h)	15.0	1 - 58	7.5	0 - 81

BMI, Body mass index; TIB, Time in bed; TST, Total sleep time; SEI, Sleep efficiency index; WASO, Wakefulness after sleep onset; SL, Sleep latency; REMlat, latency to the first REM-sleep period; S1 - SREM, percentage of the sleep stage referred to TST; ARI, arousal index; SaO$_2$min, minimun oxygen saturation; ODI4, oxygen desaturation index.

5.3.1 Evaluating prolonged spiking

In study IV the total of 115 different breathing pattern episodes (periodic hypopnea, periodic apnea, prolonged spiking, normal breathing) from the accepted 19 recordings were adopted into the study. Forty-five episodes represented periodic hypopnea, 24 episodes periodic apnea, 33 episodes represented prolonged spiking and 13 episodes normal breathing. Apnea episodes were missing from six recordings, normal breathing episodes were missing from five recordings but hypopnea episodes were found in all recordings. As expected, no hypopnea or apnea was present during prolonged spiking episodes. The selected breathing pattern episodes occurred mainly in S2 but eight out of 25 prolonged spiking episodes appeared in S3 whereas no hypopnea or apnea episodes were present in S3.

The median cortical arousal frequency was lowest during the spiking and normal breathing pattern episodes differing significantly from arousal frequencies in hypopnea and apnea episodes. The results are presented in Figure 10.

Figure 10. Median cortical arousal frequency in various breathing pattern episodes. Statistical comparisons are presented above the horizontal lines (***, p<0.001), n.s., not significant).

In the detailed analysis of nasal pressure signal the percentage of normal, round flow shape was lowest in spiking episodes and highest in normal breathing episodes. The results of the analysis of the flow shapes in various breathing patterns are depicted in Figure 11.
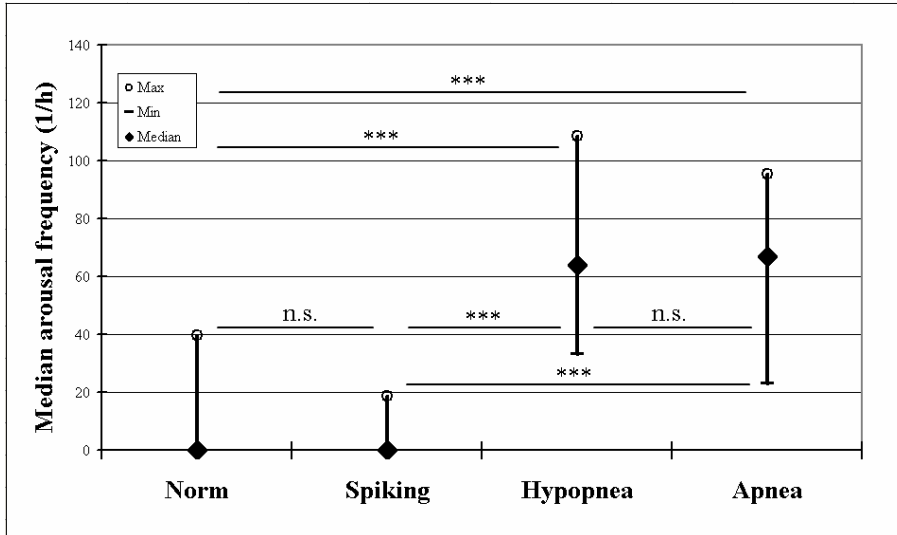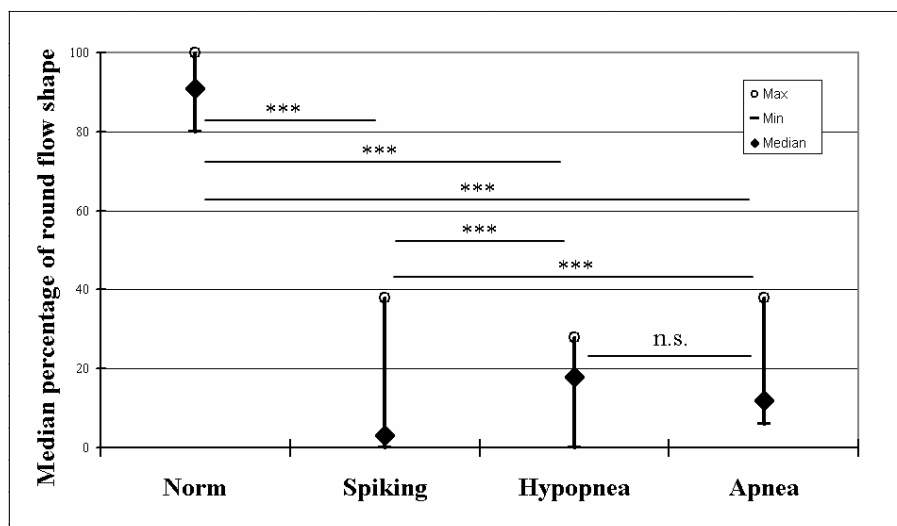
Figure 11. Median percentage of round flow shape in various breathing pattern episodes. Statistical comparisons are presented above the horizontal lines (***, p<0.001; n.s., not significant).

The median awake $TcCO_2$ of the subjects was 5.6 kPa (range 4.7-6.0). The median values of $TcCO_2$ changes per minute in the various breathing pattern episodes are presented in Figure 12. During normal breathing, 8 out of 13 episodes had a decreasing $TcCO_2$ trend, resulting in negative median $TcCO_2$ change per minute. The $TcCO_2$ increased during 28 out of 33 prolonged spiking episodes resulting in an overall median increase of 13 Pa/min. The median $TcCO_2$ change per minute for hypopnea episodes was slightly increasing, with 18 decreasing and 27 increasing values. For the apnea episodes the median $TcCO_2$ change was slightly increasing; 12 changes were decreasing, and another 12 were increasing.

The median $TcCO_2$ change in prolonged spiking episodes was statistically significantly higher than changes during other breathing episodes (Figure 12). The $TcCO_2$ changes between normal breathing, hypopnea episodes and apnea episodes did not show statistically significant differences.
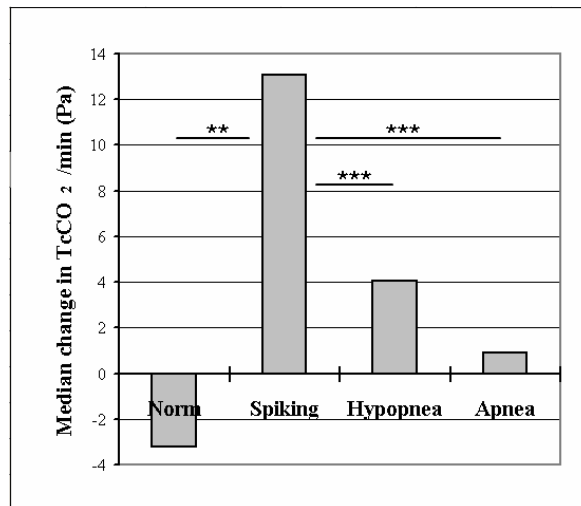
Figure 12.   The median changes in TcCO$_2$ in various breathing pattern episodes. Statistically significant comparisons are presented with asterisk   (**, $p<0.01$; ***, $p<0.001$).

The initial TcCO$_2$ -levels in various breathing episodes did not show statistically significant differences. No significant correlations between the durations of various breathing episodes and the changes per minute of TcCO$_2$ were observed indicating that the length of the breathing pattern episodes did not contribute to the level of the TcCO$_2$ change. Neither were there significant correlations between the durations of single hypopneas or apneas inside the breathing pattern episodes and the changes of TcCO$_2$ (all correlation coefficient values between -0.25 and 0.22, all p-values > 0.05).

SaO$_2$% tended to decline during normal breathing and in prolonged spiking episodes whereas in hypopnea and apnea episodes it rose slightly. The pulse rate showed a rising trend during prolonged spiking episodes, in hypopnea and apnea episodes the pulse rates slightly declined and practically no change occurred in normal breathing. The results and statistical differences are presented in Figure 13 a-b.

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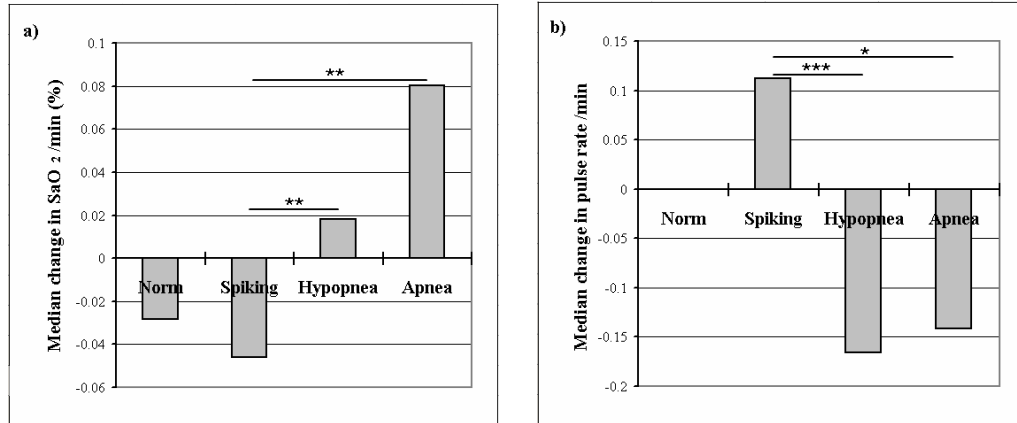
Figure 13.   The median values of a) SaO$_2$ and b) pulse rate changes per minute in various breathing pattern episodes. Statistically significant comparisons are presented with asterisk   (*, p<0.05; **, p<0.01; ***, p<0.001).

5.3.2 Evaluating tracheal sound

In study V a compressed tracheal sound curve was visually analysed. Twenty-nine subjects out of 31 possessed a pure thick sound pattern. Twenty-seven subjects had pure thin patterns. A pure plain pattern was found in 21 subjects. Examples of compressed tracheal sound patterns and simultaneous other signals are presented in Figures 14 and 15.
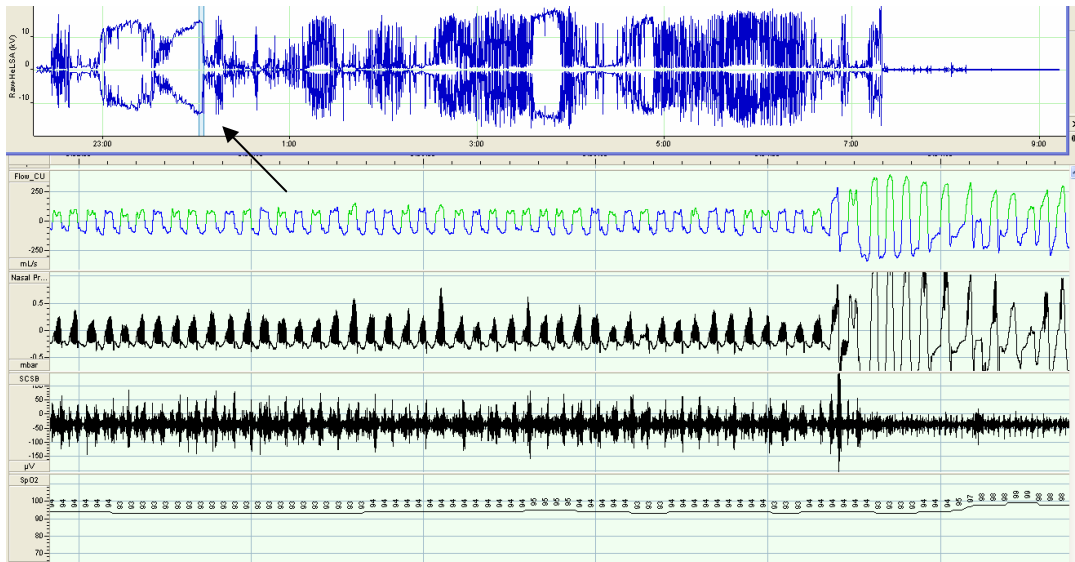
A.
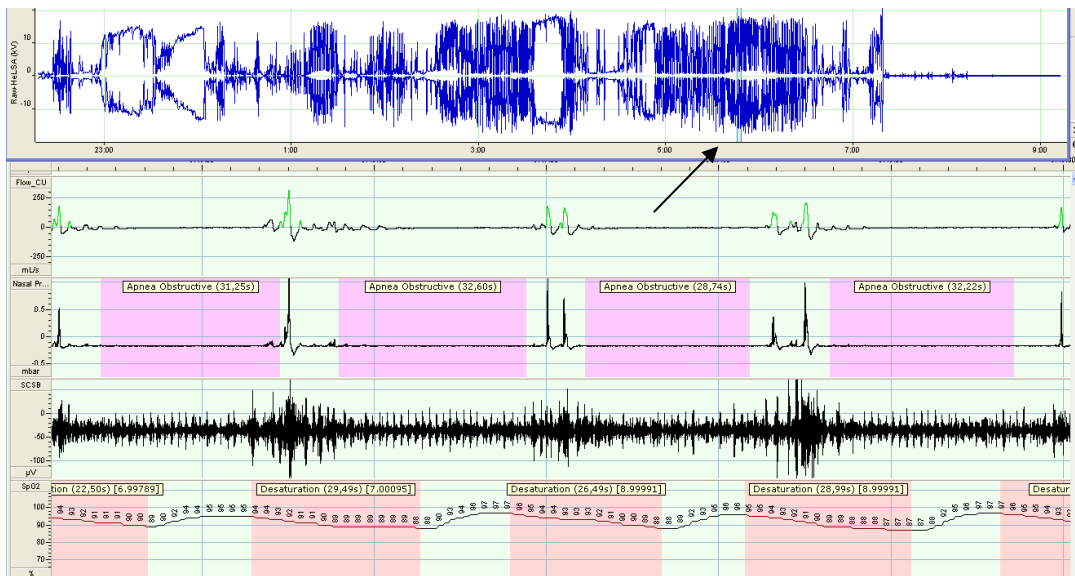
B.

Figure 14. The compressed tracheal sound signal (uppermost trace) of the whole night and a three minute episode of PSG with nasal flow, SCSB and oximetry (4 traces below) of a 45-year old male with AHI of 62/h. A) shows an example of the thin pattern. The 4 PSG traces are from the time point shown in the compressed tracheal sound curve with a vertical bar in the end of a thin period. Note the flow limitation findings in nasal flow and SCSB signals which end in arousals simultaneously with a drop in compressed tracheal sound curve. B) shows an example of the thick pattern with simultaneous obstructive apneas in PSG tracings.
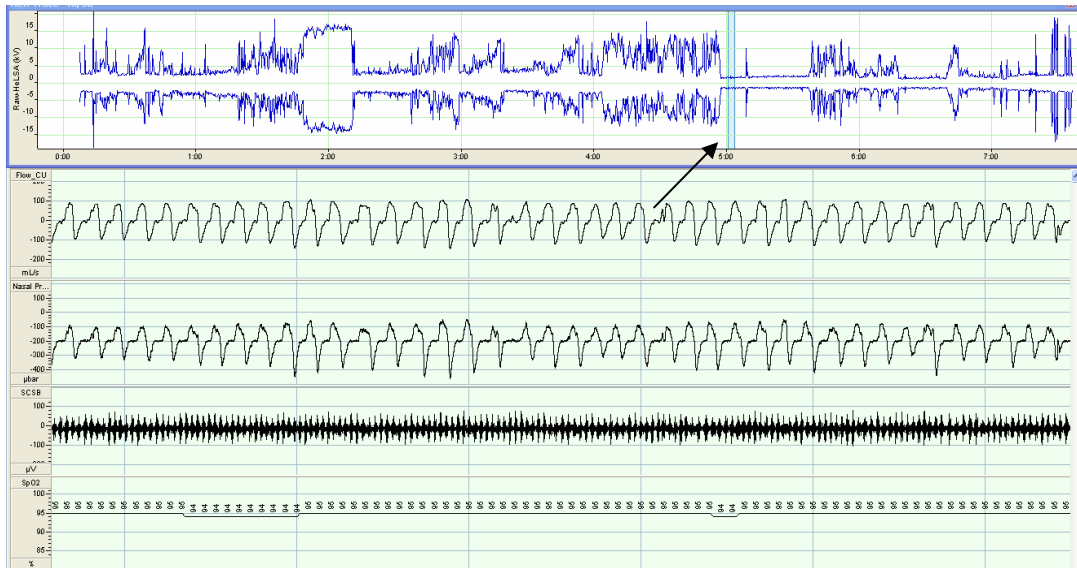
A.
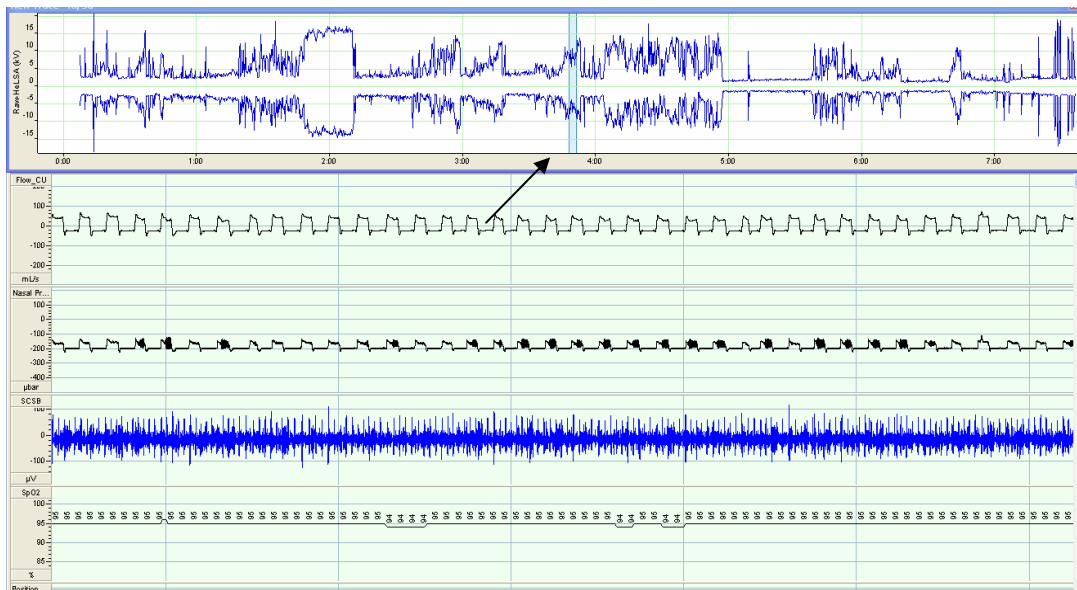
B.

Figure 15.   A) shows an example of the plain pattern with no obstruction in PSG traces in a 39-year old male with AHI of 10/h. B) shows thin pattern with flow limitation and partial mouth breathing in PSG tracings. The traces are as in the Figure 14.

From the nasal pressure channel AHI was calculated for each tracheal sound pattern. The median AHIs of various patterns in 10 minute periods are presented in Figure 16. AHI was highest in the pure thick periods differing statistically

significantly from both the pure thin AHI and the pure plain AHI but the difference between the pure thin and the pure plain AHI was not significant.
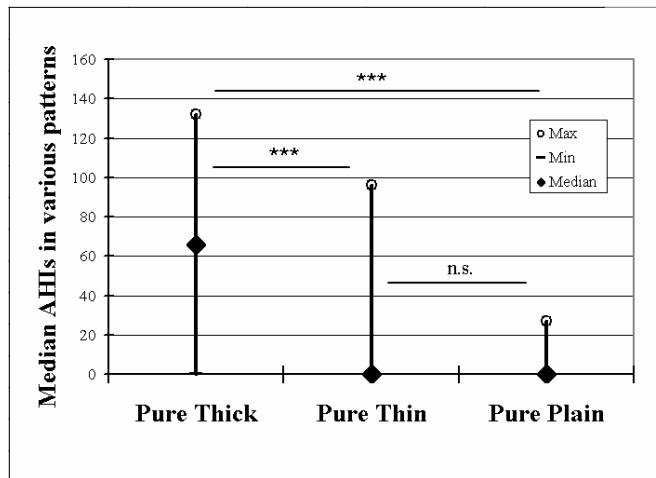
Figure 16. Median apnea-hypopnea indexes with statistics of various patterns in 10 minute periods. (***, $p < 0.001$; n.s., not significant).

Five subjects had AHI > 10/h during the pure thin period. These subjects had high AHI in the whole night analysis (34-59/h). The three subjects presenting AHI > 10/h during the pure plain periods had AHI in the whole night analysis between 12 and 24.

The results of the detailed nasal pressure channel flow shape analysis are presented in Figure 17. The proportion of round (normal, non-flattened) inspiratory flow shapes was highest during pure plain periods (median 42.5 %) and it differed significantly from the amount of round shapes during pure thick and thin periods. Although the amount of inspiratory round flow shapes seems to be lowest during pure thin periods (median 2.5 %), it did not differ statistically from the amount of round inspiratory shapes during the pure thick periods (median 19%).
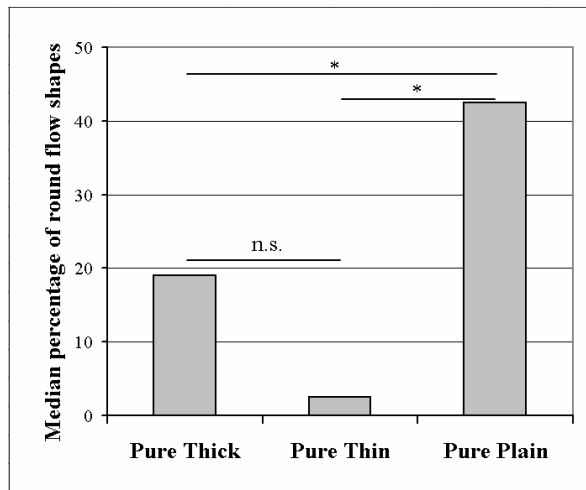
Figure 17. Median percentage of round flow shapes in various patterns. (*, p<0.05; n.s., not significant).

As thick periods seemed to comprise vast amount of apneas and hypopneas the correlation of nocturnal AHI and thick period time referred to TST was also calculated. Interestingly the nocturnal AHI was highly correlated with the thick period time (Figure 18). The Spearman's correlation coefficient (Spearman's ρ) appeared to be 0.91 (p < 0.001).
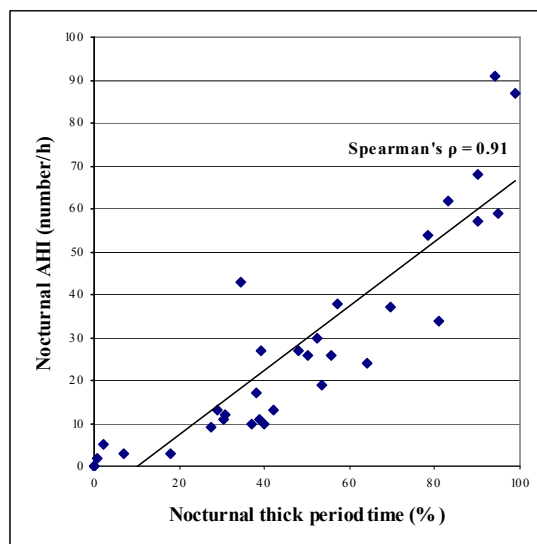
Figure 18. The relationship between the apnea-hypopnea measurements and the percentage of thick period time.

6. Discussion

6.1 Detection of periodic limb movements with the SCSB and Emfit sensors

Sleep disorders are very common and there is a constant need to develop simple but reliable devices (Flemons et al., 2003). The major advantage of the SCSB and Emfit sensors in sleep studies is that no electrodes have to be attached to patients. Studies I and II show that periodic limb movements can be reliably detected with the SCSB and Emfit sensors. Their sensitivity and specificity to detect PLMs in sleep was good and the PLM and PM indexes were quite comparable. In study II the specificity and the positive predictive value of the Emfit sensor in detecting periodic movements were 1.00 at the cut-off level of 15 movements/h. The PLM index of 15/h is considered to determine clinical significance (American Academy of Sleep Medicine, 2005) and so all the patients with indexes over 15/h were correctly classified. Additionally, the ROC curves with AUCs of 0.96 and 0.98 showed that the capability of the Emfit sensor in detecting periodic movements is very good. It was interesting to note that in study I the best performance of the SCSB to detect periodic movements was achieved in a setting where an epoch-based analysis was done from continuous recordings as in routine clinical work.

In standard movement measurements long EMG cables can be detached but movement sensors without any electrode connection to the patient assure that information about movement activity is always gained. The operational principles of the EMG and mattress movement sensors are diverse; EMG electrodes record muscle activity and movement sensors actual movements. This affects the results

when the methods are compared. It is evident that mere tonic muscle contractions without proper movements are not visible in the SCSB and Emfit. However, it is questionable if this tonic EMG activity or subtle movements are of clinical significance at all (Hening, 2004).

The SCSB and Emfit movement sensors display periodic movements irrespectively whether they occur in the lower limbs, upper limbs, shoulders, head or trunk. Their detection ability is high but the movements cannot be localized. The SCSB covers the whole area in bed but the Emfit is placed under the thoracic area of the sleeping subject. Despite of this size difference screening of periodic movements seems to be reliably possible also with the Emfit sensor. Presumably limb movements also slightly move the whole body causing movement detections in a rather small-sized but sensitive Emfit sensor.

6.2 Emfit in the evaluation of sleep-disordered breathing

Study III showed that the spiking phenomenon is clearly present also in the Emfit signal and an automatic analysis can successfully be applied in its detection. Ten different feature versions from two frequency bands for spiking detection were compared. All feature versions were independent on absolute signal amplitudes. This is important in the clinical work because of the large amplitude variation of the Emfit HFB signal among individual subjects.

All the studied feature versions were found to work well in the detection of the amplitude increase during the spiking phenomenon. A short-time FFT amplitude spectrum proved to be a useful tool in quantifying the amplitude of complexes with and without spiking. 0.5 s long overlapping windowing in FFT computation gave better results than a single 1 s window because of a better time-localisation of spiking complexes with 0.5 s segment. Overall the best results were obtained by method 10 using 0.5 s windowing. A possible reason for that might be that the mean taken first from the five 0.5 s sub-windows emphasizes the high-energy components

of the spiking complex. In the testing of different lengths of history window ranging from 10 to 80 s, they all seemed to work rather well. The median filtering in the history window provides a robust baseline during frequent spiking and also during normal Emfit HFB signal. The best results were obtained with the 60 s history window which is conveniently short as the characteristics of the signal can change during the night (Aittokallio et al., 2001a).

Study IV showed that a prolonged spiking phenomenon in the Emfit sensor during sleep in normocapnic patients seems to be related to a gradual elevation of $TcCO_2$ tension. The rises observed in normal breathing and in the episodes of apnea and hypopnea were minor or lacking. Also in SaO_2 and pulse rate measurements some significant differences between the various patterns were seen. SaO_2 values during prolonged spiking episodes showed a declining trend with a slight increase in the pulse rate while during hypopnea and apnea episodes the trend of SaO_2 was rising and pulse rates declined slightly. The non-round inspiratory flow shapes during sleep-disordered breathing confirmed that the investigated episodes were related to upper airway obstruction. The non-round inspiratory flow shapes were the most constant phenomenon during the prolonged spiking episodes whereas during the episodes of hypopnea or apnea also round shapes occurred repeatedly reflecting opening of the airway during arousals.

Sleep recordings routinely include only SaO_2 measurements to show the changes in respiratory gas balance during episodes of repetitive apnea and hypopnea. However, both the partial pressures of arterial oxygen and carbon dioxide are the basis of the metabolic control of breathing during sleep (Douglas, 2000). End-tidal carbon dioxide measurements are widely used in operating room, intensive care units and children's sleep studies. Still in adults the carbon dioxide measurements are infrequently used in sleep studies even if the rise in carbon dioxide pressure is the strongest stimulant of breathing during sleep. In adults with sleep-disordered breathing the transcutaneous method of carbon dioxide measurement might operate better compared to the end-tidal measurement of CO_2 because the $TcCO_2$ recording gives a continuous signal also during apnea. The measurement of $TcCO_2$ might therefore provide additional information about the characteristics of breathing disturbances during sleep. Changes in $TcCO_2$ occur quite slowly and the method is

most suitable in situations where alterations are not rapid. However, our clinical impression was that even minor arousals with changes in breathing caused a notable change in $TcCO_2$ level.

The findings in study IV suggest that prolonged spiking phenomenon measured by the Emfit sensor represents a variant of nocturnal breathing abnormality which differs from classical hypopnea and apnea periods. The Emfit might provide a valuable addition to sleep studies as a simple method that doesn't interfere with sleep. It can be used as a supplementary device because with a movement sensor a reliable signal is always gained contrary to nasal prongs, which can become detached. Also in the case of mouth breathing the Emfit provides a continuous signal. The Emfit can thereby reduce the risk of unsuccessful respiratory recordings.

Patients in study IV presented with various forms of breathing patterns with the spiking pattern as an inclusion criterion. Hence, sleep apnea patients with the most severe condition were not included in the study if they had only apneas and hypopneas. The prolonged spiking pattern can be the only respiratory abnormality during sleep in many symptomatic patients. It is particularly common in postmenopausal females (Polo-Kantola et al., 2003) who respond to nasal CPAP therapy with good adherence (Anttalainen et al., 2007). The spiking pattern is also more common in OSAS patients with narrow upper airway at tongue base or hyoid bone levels (Polo et al., 1991) or after shortening of the soft palate (Polo et al., 1989).

6.3   Compressed tracheal sound analysis in the evaluation of sleep-disordered breathing

Study V shows that tracheal sound analysis with a compressed signal might provide a reliable and efficient tool for screening sleep apnea. The tracheal sound analysis in sleep-disordered breathing has previously focused mainly only on visual or automatic analysis of respiratory events (Cummiskey et al., 1982, Van Brunt et al., 1997, Nakano et al., 2004). Likewise the severity and harmfulness of snoring has

mostly been associated simply with its intensity (Lugaresi et al., 1983, Itasaka et al., 1999, Wilson et al., 1999). As a new approach study V evaluated the characteristics of the compressed tracheal sound signal in which the whole night can be compacted into one page for visual examination of different respiratory patterns.

The HeLSA -tracheal sound analyzer, which has initially been developed for the diagnostics of lung diseases, was used in the study. The measurements were performed with the sampling rate of 11025 Hz. The pulmonological studies are quite short-term and the collected amount of data is still reasonable. Sleep recordings on the contrary last several hours and with the high sampling rate the amount of data is huge, which increases also the duration of the analysis. Thus a heavy data compression was used to reduce the amount of data substantially, to speed up the evaluation of the data and to improve the visualization of the signal.

Three distinct patterns of compressed respiratory sound signal could be observed: Thick, thin and plain. The patterns were easy to recognize and separate visually. During the periods of plain signal curves the minimum and maximum levels of the signals were quite constant and close to zero indicating that the sound levels recorded remained low. Also the proportion of round flow shape measured by nasal flow pressure was most substantial during the pure plain periods suggesting that this sound pattern represents mostly normal breathing.

When during the episodes of plain signal curve sound levels were low, the thick signal curve represented highly variable sound levels. If reviewed along with nasal pressure signal the thick sound pattern seemed to be composed by intermittent alternation of disturbed breathing during apneas or hypopneas with some normal breaths between them. The correlation studies between AHI and the total duration of different sound signal patterns during whole night indicated that the higher the AHI the higher is the total duration of the thick sound periods. Conversely, the higher the proportion of plain period time, the lower the AHI. The thin period time did not show significant correlation with AHI. The amount of 30% of the total nocturnal time of the thick signal could be used as a reliable limit of abnormal AHI of 5/h.

In the analysis of the representative 10-minute sequences the different compressed sound periods proved to differ in the amount of respiratory events (apneas and hypopneas). Apneas and hypopneas were most common during pure thick periods, whereas during pure thin and plain periods only few subjects had apneas and hypopneas. All the subjects presenting apneas or hypopneas during pure thin and plain periods had increased AHI in the whole night analysis.

The thin sound signal pattern differed substantially both visually and in characteristics from the thick and plain sound signals. In the thin pattern the compressed curves deviate clearly from the zero level reflecting episodes of continuous loud noise. During pure thin sound periods the proportion of flattened, non-round inspiration flow shape was highest and some subjects had apneas and hypopneas. Therefore the thin sound pattern seems to represent breathing with abundant flow limitation and sometimes apneas.

6.4   Evaluation of flow limitation

The reference standard for quantitative assessment of respiratory effort is the measurement of esophageal pressure with continuous overnight monitoring (1999). As this technique is not routinely used, indirect methods for estimating inspiratory effort non-invasively have been developed.

Nasal cannula can be used to assess breathing flow during sleep by recording pressure at the nostrils. The rationale of the method is that the airflow turbulences at the nostrils induce a pressure that is directly related to the magnitude of flow. The setting is extremely simple and has an excellent dynamic response. One limitation of this device is that the relationship between nasal prongs pressure and airflow is not linear (Montserrat et al., 1997) but can be linearised (Farre et al., 2001) to obtain an excellent surrogate signal of flow. Nasal cannula is also suitable for detecting flow limitation without hypopnea because the device is able to track the details of the inspiratory waveform contour.

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Guilleminault with co-workers have introduced the upper airway syndrome, which is characterized with flow limitation events occurring simultaneously with increased esophageal pressure and arousal (Guilleminault et al., 1993). Later the short-term (even merely one inspiration) non-round flow limitation pattern in nasal cannula has been associated to increased breathing effort and arousals (Johnson et al., 2005).

However, nasal cannula has some potential drawbacks in practice. First, the common experience is that it can detach during the night. Secondly, the device cannot detect mouth breathing obviously, with the result that the flow signal is completely lost when the patient breathes through the mouth, or partially lost when the patient inspires through the nose and expires through the mouth (Ballester et al., 1998). And it has also been suggested that nasal cannula with inadequate size could induce an increase in airway resistance (Lorino et al., 2000). However, in a study by Thurnheer an co-workers subjective nasal obstruction did not impair the accuracy of nasal pressure monitoring (Thurnheer et al., 2001) though in another study impaired nasal ventilation prevented adequate measurements of nasal pressure in 9% of the subjects (Series and Marc, 1999).

In studies IV and V inspiratory flow shapes were roughly divided only into two categories; round and non-round shapes. However, several flow shapes can be differentiated (Aittokallio et al., 1999) and pathophysiologic significances of different flow shapes are still unestablished (Saaresranta et al., 2003). In future it would be important to differentiate the obstructive flow patterns from the non-obstructive ones to facilitate the selection of correct treatment for the patients.

According to studies III and IV the mattress movement sensors can also be used for the evaluation of flow limitation. The mechanisms of the prolonged spiking phenomenon have been described and investigated in several studies. In SCSB prolonged spiking has been suggested to arise from increase in respiratory resistance during sleep (Alihanka, 1987, Polo et al., 1989) and breathing effort (Kirjavainen et al., 1996). Gradually growing spike amplitude correlates well with the increase in negative esophageal pressure (Kirjavainen, 1997). Study IV provides further evidence that the phenomenon of prolonged spiking recorded during sleep with

sensitive movement sensors such as Emfit arises from the increased drive of breathing caused by flow limitation and increasing transcutaneous CO_2. The Emfit sensor seems to provide a simple way to distinguish presumably clinically important progressive flow limitation with gradually increasing CO_2.

The compressed tracheal sound signal analysis visualized a distinct thin sound pattern which seems to represent breathing with abundant flow limitation. The sensitive but small microphone as a sound recording device may interfere with sleep less than the nasal pressure transducer, inductive plethysmography or the other attached transducers used for detecting breathing abnormalities. The fixation site of the microphone in jugulum is well sheltered from detachment. Thus tracheal sound analysis can provide an easy and reliable non-invasive way of studying characteristics of sleep-disordered breathing. The compressed recording is rapid to analyse visually but the method still needs further studies before it can be utilized in the investigations of sleep-disordered breathing.

6.5   Prolonged flow limitation

The prolonged flow limitation is found to be rather common (Hernandez et al., 2001). It has been evaluated in stepwise CPAP titration studies. The contour of inspiratory flow appears as the simplest variable that best correlates with lowest esophageal pressure during CPAP titration (Montserrat et al., 1995). Ayappa and co-workers found in their CPAP study that increasing obstruction caused by pressure decrease induced first prolonged flow limitation. Flow limitation was found to be even more sensitive indicator of obstruction than appearing of snoring (Ayappa et al., 1998).

However, the significance of the prolonged flow limitation is still unclear. Some results tend to show that correcting prolonged flow limitation in CPAP treatment is associated with a higher attentiveness and a higher efficiency in normalizing daytime vigilance than with conventionally titrated CPAP (elimination of only apnea/hypopnea and snoring) (Meurice et al., 1998). In a CPAP study it has also

been shown that the end-tidal carbon dioxide rises during induced prolonged flow limitation (Calero et al., 2006). Study IV demonstrated that prolonged spiking in the Emfit sensor occurring with simultaneous flow limitation, is associated with a spontaneous increment in transcutaneous carbon dioxide tension. Together these studies implicate that the prolonged flow limitation, even without apnea and hypopnea type periodic respiratory alternation, could indicate a breathing disturbance, which might have significant behavioural and metabolic effects.

6.6   Sleep recording systems

Polysomnography has been considered the gold standard for diagnosing obstructive sleep apnea (Kushida et al., 2005). The number of measured parameters in PSG usually varies from 16 to 32 leading to a large amount of cables, which are in danger to detach and demand continuous surveillance. Thus in-laboratory, technician attended PSG requires skilled performance of recording, scoring and professional interpretation. It is considered expensive and technically demanding, can be inconvenient for the patient, and it has been argued that many patients do not require such a comprehensive procedure to diagnose uncomplicated OSA. The term 'portable monitoring' is applied to technology that can be performed outside a sleep centre.

Portable monitoring is widely used as an alternative approach to PSG in countries other than the United States (Flemons et al., 2004). Recently a group from Switzerland concluded that unattended at least three channel cardio-respiratory recordings and PSG are complementary diagnostic tools, depending on the population and suspected diagnosis of the patients (Thurnheer et al., 2007). They stated that cardio-respiratory recordings are capable of obviating a great proportion of the more expensive and less widely available PSG.

Recent studies (e.g. Whitelaw et al., 2005, Mulgrew et al., 2007) have been able to demonstrate that compared with PSG, portable monitor could effectively diagnose patients with OSA and that the short-term outcome would not be adversely

affected. Based on these studies Centers for Medicare and Medicaid Services (CMS) has accepted the use of portable monitor devices if they utilize at least 3 channels to diagnose OSA in patients who will be treated with CPAP. It further stated that those patients have 12 weeks to demonstrate a positive response to CPAP. This is an important policy change for CMS to allow portable monitors to be used to diagnose OSA in the Medicare population (Collop, 2008).

6.6.1 Mattress movement sensors and compressed tracheal sound analysis in portable recordings

The facts that the SCSB and Emfit with no electrodes attached to a patient and a tracheal sound recording with a microphone interfering only minimally or not at all with sleep, emphasize their suitability for ambulatory sleep recordings. When studying patients with poor sleep quality and low arousal threshold, the advantage of non-invasive monitoring becomes particularly important.

The amount of body movements is a good estimate for the peacefulness or restlessness of the sleep (Gardner and Grossman, 1975, Kronholm et al., 1993). The number of movements per night in the SCSB studies in healthy volunteers has varied from 80-200 (Alihanka and Vaahtoranta, 1979) which is in line with studies combining direct observation, EMG and videomonitoring (Gardner and Grossman, 1975). It can be assumed that over 200 movements per night points to disordered sleep. However, in most sleep disorders the increases in movement activity are quite non-specific findings. The SCSB is also able to evaluate changes in the state of autonomic nervous system since almost all movements in bed are associated with heart rate acceleration and decelerations (Alihanka, 1982). Future studies focusing on movement frequency, duration and intensity spectra and associated changes in breathing pattern may clarify the underlying issues and help to discriminate more and less sleep disturbing movements.

The EEG is the standard method for classifying sleep as sleep stages and distinguishing sleep and wake is therefore not fully possible with movement sensors. Thus the movement counts measured with the SCSB and Emfit without a

full polysomnogram are combinations of periodic movements during sleep and wakefulness. However, including all the periodic movements in the analyses may even provide a better impression of the movement disorder during night (Michaud et al., 2002). Cortical arousals cannot either be observed with movement sensors but the SCSB and Emfit can provide complementary information through movement analysis since cortical arousals very often induce movements. Even simple duration measurement can offer additional information since a statistical correlation between the duration of PLM movements and EEG-arousals has been shown (Pollmächer and Schulz, 1993). The analyses differentiating movements with and without arousal have been considered important (Hornyak et al., 2006, Claman et al., 2006) but this matter is still controversial (Chervin, 2001, Al-Alawi et al., 2006).

Ambulatory recordings with SCSB and Emfit sensors are easy to perform. They can be conducted in ordinary wards and recordings are inexpensive compared to a full polysomnography. The SCSB is the same size as a bed, which causes some trouble with portability, but the Emfit is made of flexible material, is easy to carry and any size or shape of sensor can be used. Costs of the studies remain reasonable and recordings with the SCSB and Emfit can be repeated during several consecutive nights if necessary. The SCSB and Emfit can also be used for the evaluation of the efficacy of medical treatment in RLS and PLM disorder patients. Compared to actigraphy, mattress type movement sensors have the important advantage, i.e. sleep-related breathing disorders can be evaluated as well as periodic limb movements or increased movement activity. With the SCSB and Emfit normal breathing and breathing abnormalities presenting four types of periodic breathing or prolonged spiking episodes, can be distinguished. Thus movements coexisting with breathing disorders in the SCSB and Emfit PLM recordings do not pose a similar problem as with actigraphy.

The SCSB, Emfit and compressed tracheal sound analysis can all be parts of portable sleep recording systems or even operate as stand-alone devices. The SCSB and Emfit can be used to quantify periodic limb movements and all of them to evaluate sleep-disordered breathing. Additionally, they all have special features that help to detect and characterize short-term and especially prolonged flow limitation.

Future studies will investigate further the performance and impact of these devices in diagnostics of sleep disorders.

7. Conclusions

Studies I and II

The SCSB and Emfit sensors seem to be suitable for detecting periodic limb movements and estimating their frequency. Thus they might be suitable for screening purposes as stand-alone devices and, on the other hand, they can improve the reliability of polysomnographic recordings. Though the Emfit is placed under thoracic area of a subject, its performance is comparable with the SCSB. The mattress movement sensors have some advantages compared to anterior tibialis EMG and actigraphy. EMG electrodes might get detached but the SCSB and Emfit always provide information about periodic and general movement activity because no electrodes are connected to the subject. Breathing disorders appearing simultaneously with periodic movements disturb the interpretation of actigraphy but the SCSB and Emfit mattress movement sensors can characterize sleep-disordered breathing as well.

Study III

The spiking phenomenon previously observed with the SCSB is clearly present also in the Emfit signal and an automatic analysis can be applied in its detection successfully. The performance of the automatic method to detect the spiking phenomenon in the Emfit signal proved to be satisfactory. The algorithm monitors amplitude levels of spiking complexes and detects large relative increases. Clinically it is important that the detection is not dependent on absolute waveform amplitudes and therefore does not require any recording-specific tuning prior to application. The developed spiking detection system can be used as a part of automatic analysis of the Emfit signal.

Study IV

The prolonged spiking phenomenon recorded during sleep with the Emfit sensor is associated with flow limitation and increased drive of breathing leading to increasing transcutaneous CO_2. The prolonged spiking differs clearly from periodic breathing patterns and seems to represent a variant of breathing disturbance during sleep. The Emfit sensor seems to provide a simple way to distinguish presumably clinically important progressive flow limitation with gradually increasing carbon dioxide.

Study V

The compressed tracheal sound analysis provides a promising screening method for obstructive apneas and hypopneas. The compressed sound curve is highly visual and the proportions of various breathing patterns can be easily estimated. Further studies are still needed to validate the method and to enhance and speed up the analysis with automatic or semiautomatic methods. Besides the usability as a screening device of apneas and hypopneas, the compressed tracheal sound analysis can also distinguish a pattern, often with long duration, which seems to be related to flow limitation. The significance of prolonged flow limitation is insufficiently clarified and the compressed tracheal sound analysis might provide an efficient tool for studies in order to examine its possible behavioural and metabolic effects.

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Detection of periodic leg movements with a static-charge-sensitive bed

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SUMMARY We evaluated the performance of the static-charge-sensitive bed (SCSB), a non-invasive movement sensor, in detecting nocturnal periodic movement activity using simultaneous bilateral anterior tibialis electromyography (EMG_{at}) as a reference. Two different study setups were used, one with 500 random record extracts, another with 10 continuous recordings. The inter-rater reliability between two independent scorers was 0.92 in scoring EMG_{at} and 0.91 in scoring SCSB. In an epoch based analysis, depending on the study setup and scorer, the sensitivity of the SCSB to detect periodic leg movements was 0.81–0.94 whereas the specificity was 0.84–0.89. In a movement-by-movement analysis, despite incomplete concordance between the two methods the total number of movements per hour were comparable. Our findings support the use of the SCSB as a non-invasive alternative to anterior tibialis EMG recordings to reveal the presence of nocturnal periodic movement activity and estimate its frequency.

KEYWORDS anterior tibialis EMG, periodic leg movements, restless legs, sleep, static-charge-sensitive bed

INTRODUCTION

Periodic leg movements (PLM) are characterized by stereotyped, repetitive extensions of the big toe in combination with partial flexion of the ankle, knee and sometimes hip (Thorpy 1990). These movements occur during nocturnal wakefulness (Pollmächer and Schulz 1993) or sleep and are usually recorded with EMG electrodes attached on right and left anterior tibialis muscles. Similar repetitive movements may occur in the upper limbs but are rarely monitored. PLM is linked to restless legs syndrome and is commonly observed in association with various somatic disorders, including anaemia, uraemia, rheumatoid arthritis, chronic myelopathies, peripheral neuropathies, respiratory insufficiency, narcolepsy, snoring, or sleep apnoea (Montplaisir and Godbout 1989). PLM is common during pregnancy (Nikkola *et al.* 1996) or may be an isolated finding in asymptomatic subjects. The prevalence of PLM increases with age and reaches a frequency of 29% in the age group over 50 years (Bixler *et al.* 1982). PLM may interfere with sleep, induce arousals and go along with fatigue or

sleepiness. But they are also seen in normal subjects without changes to the overall structure of sleep. Recognition of PLM in association with a somatic disorder may therefore help to understand the overall clinical symptomatology. However, at present there are no simple and inexpensive methods available to establish the PLM diagnosis at ordinary wards. The static-charge-sensitive bed (SCSB; Alihanka *et al.* 1981) is a non-invasive movement sensor, which is well adapted for sleep studies, since no electrodes need to be attached to the subject. The SCSB has previously been used to study, e.g. sleep quality (Hyyppä and Kronholm 1987; Kronholm *et al.* 1987) and identify episodes of sleep apnoea (Polo *et al.* 1988) or partial upper airway obstruction (Polo 1992). We evaluated the performance of the SCSB in detecting nocturnal periodic movement activity using simultaneous bilateral anterior tibialis electromyography (EMG_{at}) as a reference.

SUBJECTS AND METHODS

Subjects

Fifty polysomnographic sleep recordings were carried out in 47 patients (35 men and 12 women) who had been referred to the sleep laboratory because of heavy snoring, suspected sleep

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Figure 1. A representative 4.5 min epoch with simultaneous periodic leg movements disclosed by the anterior tibialis EMG channels and periodic movement activity shown by the SCSB. BCG = ballistocardiogram, RESP = respiratory movement, R.EMG$_{at}$ = right anterior tibialis EMG, L.EMG$_{at}$ = left anterior tibialis EMG.

apnoea or nocturnal restlessness. Although the majority of the patients presented with PLM, recordings without any PLM episodes were also included. The mean age of the patient group was 50.3 years, and the age range was 33–70 years.

Sleep study

The sleep studies consisted of simultaneous all-night recordings of the following parameters: electroencephalogram (EEG: C$_4$A$_1$), electrooculogram, submental electromyogram (EMG), electrocardiogram (ECG), arterial oxyhemoglobin saturation (SaO$_2$), right and left anterior tibialis electromyogram (EMG$_{at}$), continuous video monitoring as well as respiratory movements and the heart pulse (ballistocardiogram) recorded with a static-charge-sensitive bed (SCSB). The signals were recorded on a Grass Model 79C pen recorder (Grass Instrument Co., Quincy, MA, USA) with a paper speed of 1.5 mm s^{-1}.

Movement detection

The EMG$_{at}$ was recorded with surface electrodes placed on the belly of the anterior tibialis muscle in both legs. To adjust the EMG signal amplitude within the dynamic range of the recorder, the patients were asked to dorsiflex each of their feet prior to the onset of the recordings. The composite movement signal recorded with the static-charge-sensitive bed (SCSB) was low and high pass filtered into two channels, one representing the respiratory movements, another displaying the heart pulse (ballistocardiogram). Body and limb movements were distinguished as high amplitude deflections displayed on either channel.

Data analysis

The frequency of PLM was determined from the EMG$_{at}$ recordings according to the criteria proposed in the International Classification of Sleep Disorders (Thorpy 1990). An epoch was defined as PLM-positive, if it contained a series of at least four consecutive EMG$_{at}$ events that fulfilled the criteria for duration (0.5–5 s) and the movement interval (5–90 s). On the SCSB recordings, episodes of movement activity were distinguished from the ballistocardiogram by higher signal amplitude or different wave form (Fig. 1). An epoch was defined as periodic movement activity (PMA) – positive, if there was a series of at least four consecutive movements in the epoch that fulfilled the above mentioned

criteria for duration and interval. The sensitivities, specificities and the positive and negative predictive values were computed using standard formulas (Polo *et al.* 1988).

Respiratory disturbances were analysed from the SCSB channels according to previously described scoring principles (Polo *et al.* 1988). If the epoch contained one or more periods of continuous motor activity exceeding 40 s, it was scored as motor active wakefulness (MAW).

In the first setup 500 samples, 10 from each recording, were randomly picked up for analyses. Each sample contained 4.5 min of recording which, according to the PLM definitions, would allow four repetitive movements to occur at three intervals of 90 s. The right and the left EMG$_{at}$ channels and the two SCSB channels (the ballistocardiogram and the respiratory movements) were separated from the rest of the recording, and cut apart from each other. The EMG$_{at}$ and SCSB extracts were randomly ordered and separately analysed by two scorers. Seven samples were eliminated from the analyses because of defects in the EMG channels.

In the second setup continuous all-night recordings from seven men and three women with periodic movement activity exceeding 15% of time in bed were analysed. Two independent scorers separately analysed the EMG$_{at}$ and the SCSB curves, which had been detached by a horizontal cut into two sets of paper recordings. The average length of the recordings was 424 min (range 253–523 min). The analysis epoch length was one page equal 200 s at the paper speed of 1.5 mm s^{-1}. In addition to epoch-based analyses, the periodic movements were also analysed on a movement-by-movement basis.

RESULTS

Setup 1

The number of positive PLM and PMA epochs appearing as isolated, in conjunction with respiratory abnormalities or during motor active wakefulness are presented in Table 1. Out of 172 (scorer 1)/193 (scorer 2) PLM epochs 145/157 were detected with the SCSB and out of 179/185 PMA epochs 143/156 were detected with the EMG$_{at}$.

The inter-rater reliability was 0.92 in scoring EMG$_{at}$ and 0.91 in scoring the SCSB. The two scorers agreed in 430 and 428 epochs and disagreed in altogether 98 epochs. Forty per cent of the disagreements occurred in epochs with respiratory abnormalities (Fig. 2) and 21% in epochs with motor active wakefulness (Fig. 3, upper panel). Eighteen per cent occurred

Table 1 The number of epochs with periodic leg movements (PLM) detected with the anterior tibialis EMG and periodic movement activity (PMA) recorded with the SCSB. ISOL = isolated movements, RA = respiratory abnormalities, MAW = motor active wakefulness, NA = not analysed (Setup 1)

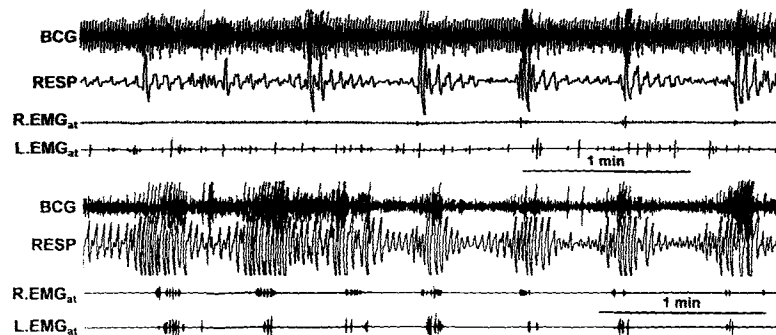| | | EMG | | | SCSB | | |
	Total epochs	*ISOL*	*PLM During RA*	*During MAW*	*ISOL*	*PMA During RA*	*MAW*
Scorer I	493	116	43	13	139	40	37
Scorer II	493	117	53	23	140	45	NA

Figure 2. Association of respiratory abnormalities was the most frequent reason for inconsistent scoring between the anterior tibialis EMG and the SCSB. A false positive epoch (upper panel) and a false negative epoch (lower panel). BCG = ballistocardiogram, RESP = respiratory movement, R.EMG$_{at}$ = right anterior tibialis EMG, L.EMG$_{at}$ = left anterior tibialis EMG.
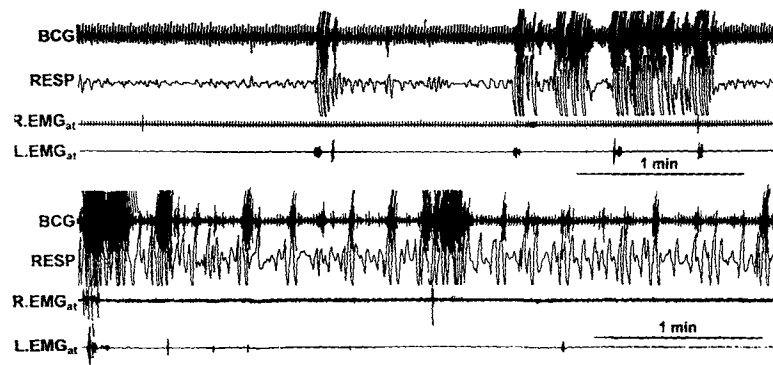
Figure 3. Upper panel: periodic leg movements during an epoch with SCSB-defined motor active wakefulness (MAW). Lower panel: an example of a false positive epoch: the SCSB discloses periodic movement activity in the absence of periodic leg movements. In the video playback the events were identified as stereotyped periodic movements of the shoulder. BCG = ballistocardiogram, RESP = respiratory movement, R.EMG$_{at}$ = right anterior tibialis EMG, L.EMG$_{at}$ = left anterior tibialis EMG.

in epochs containing only 4–5 movements as 4 movements was critical by definition.

The sensitivity of the SCSB to detect epoch with PLM was 0.84 (scorer 1)/0.81 (scorer 2). The specificity was 0.89/0.84, the positive predictive value 0.80/0.84 and the negative predictive value 0.92/0.89. In one epoch periodic movements in EMG$_{at}$ were not at all detected with the SCSB because the leg was not on the sensor. In three epochs the movements were slow and faint, and were noticeable only on the SCSB respiratory channel. In four epochs (1%) the scoring

disagreement was as a result of periodic movements appearing in the upper limb, detectable in our setting only by the SCSB (Fig. 3, lower panel).

The sensitivity and the specificity to detect isolated periodic movements without simultaneous respiratory abnormalities or motor active wakefulness were 0.90/0.91 and 0.91/0.94, respectively.

The performance of the SCSB was also analysed in function of the frequency of PLM in each epoch (Fig. 4). The highest sensitivity was observed with 11–12 movements in the 4.5 min

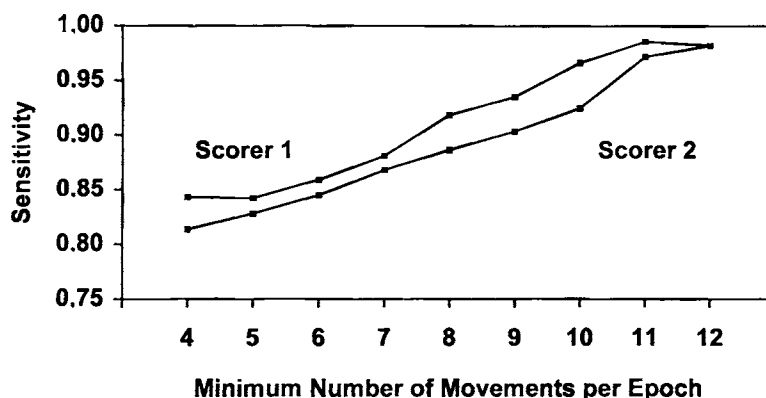**Figure 4.** The sensitivity of the SCSB to detect periodic leg movements in function of the frequency of movements in each epoch (Setup 1).

Table 2   The performance of the SCSB to detect periodic movements in continuous recordings (Setup 2)

| | Sensitivity | | Specificity | | Positive predictive value | | Negative predictive value | |
	Mean	*Range*	*Mean*	*Range*	*Mean*	*Range*	*Mean*	*Range*
Scorer I	0.93	0.85–1.00	0.87	0.62–0.96	0.86	0.66–0.94	0.94	0.85–1.00
Scorer II	0.94	0.84–1.00	0.89	0.73–1.00	0.87	0.74–1.00	0.96	0.86–1.00

epoch, which corresponds to the average movement interval of 24–27s.

Setup 2

The continuous recordings were analysed epoch-by-epoch as well as movement-by-movement. In the epoch-by-epoch analysis, the mean frequency of periodic movement activity was 47.1% (range 21.9–80.0%) of time in bed. The inter-rater reliability analysing the EMG_{at} was 0.93 (range 0.86–1.00) and 0.91 (range 0.83–0.96) with the SCSB. The performance of the SCSB to detect periodic movements was better in continuous epochs than in random epochs (Table 2).

In the movement-by-movement comparison the mean sensitivities were 0.81 (range 0.63–0.93; scorer 1) and 0.80 (range 0.56–0.93; scorer 2). The mean PLM vs. PMA indexes were 65 vs. 63 or 55 vs. 53 movements per hour of time in bed, depending on the scorer (Fig. 5).

The SCSB did not fail to reveal periodic movement activity in any patient studied. Again, the presence of respiratory abnormalities and motor active wakefulness were the major source of discrepancy between EMG_{at} and SCSB. Since periodic movement activity in the upper limbs or trunk was not detected with our reference, bilateral anterior tibialis EMG, some 'false positive' epochs were observed with the SCSB, which reduced the calculated specificity.

DISCUSSION

The present study showed that periodic movement activity in bed is reliably detected without electrodes, by using the SCSB.
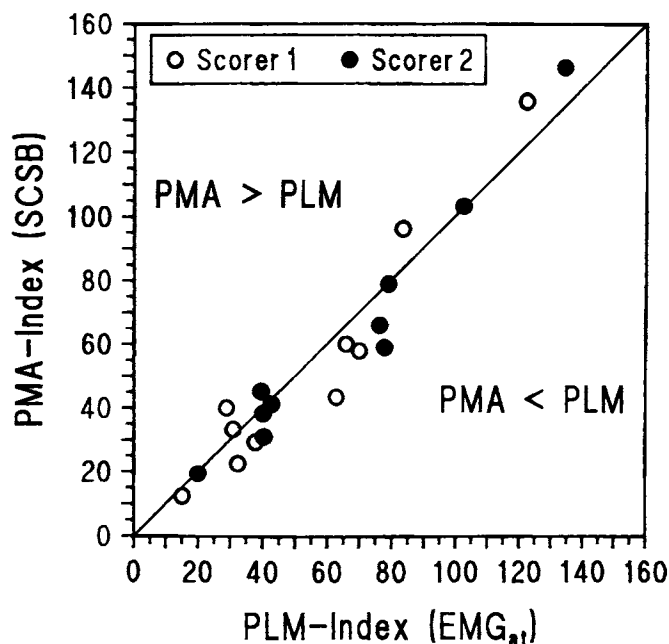
Figure 5. The PMA index provides a decent estimation of the PLM index. PLM = periodic leg movements, PMA = periodic movement activity (Setup 2).

When studying patients with poor sleep quality and low arousal threshold, the advantage of non-invasive monitoring becomes particularly important. Differentiation of isolated or respiratory events-related movement activity is possible since the SCSB also displays respiratory movements. Therefore, the

SCSB is an alternative to EMG_{at} to reveal the presence of periodic movement activity in bed and estimate its frequency. The recordings are easy to repeat at ordinary wards or in home environment at low running costs.

Periodic leg movements are not detected with the SCSB during gross body movements. Although movements that fulfilled the criteria for PLM were sometimes observed in the EMG_{at} during substantial amounts of gross body movement activity on the SCSB, it is possible that some of these movements would not be true PLM but repeated voluntary muscle contractions. Some authors report that wakefulness, NREM-sleep and REM-sleep can be differentiated with the SCSB in healthy subjects (Jansen and Shankar 1993), but it remains uncertain whether this also applies to patients with restless legs while falling asleep or having periodic movements during sleep. The clinical importance of PLM is appreciated according to the presence or absence of cortical arousals associated with PLM. It is clear that EEG-defined cortical arousals cannot be directly appreciated with the SCSB and for that purpose the EEG is needed. The movement properties may provide indirect evidence for associated cortical events. Pollmächer and Schulz (1993) have shown a statistical correlation between the duration of PLM movements and EEG-arousals. There is still insufficient proof for that only PLM with cortical arousal do interfere with sleep. Further studies into the movement frequency, duration or intensity spectra and associated changes in breathing pattern provided by automatical analyses might reveal additional variables to discriminate more and less 'sleep disturbing' periodic movements.

The SCSB displays periodic movements irrespectively whether they occur in the lower limbs, upper limbs, shoulders, head or trunk. Therefore, the method has a high sensitivity to detect any periodic movements in the body but is unable to localize them. Despite incomplete concordance between the two methods in the movement-by-movement analysis, the PLM and PMA-indexes (number of movements per hour) were comparable. This allows a direct comparison of results between studies using either of these methods. Combination of local EMGs and the SCSB is likely to provide more accurate determination of the total periodic movement activity than one method alone. This would also minimize the risk for losing information in case of detachment or poor contact of EMG electrodes.

According to our results, the best performance of the SCSB to detect periodic movements is achieved in a clinical setting where an epoch-based analysis is carried out from continuous recordings. In addition to detecting specific periodic movement activity, the SCSB provides more insight into the associated sleep disturbances as far as general motor activity and sleep-related breathing abnormalities are concerned.

ACKNOWLEDGEMENTS

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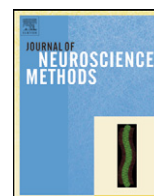
Periodic limb movement screening as an additional feature of Emfit sensor in sleep-disordered breathing studies

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ABSTRACT

Background: The standard method for recording periodic limb movements is anterior tibialis electromyography (EMG) but other methods are also used. A new movement sensor Emfit (ElectroMechanical Film) provides information about sleep-disordered breathing but also shows movements in bed. The aim of the study was to investigate the usability of a small Emfit sensor in revealing periodic movements.
Methods: Twenty seven consecutive patients were studied. Periodic movements in EMG and Emfit were scored blindly and periodic leg movement index (PLMI) for EMG and periodic movement index (PMI) for Emfit were counted. Spearman's correlation coefficient was used to assess the relationship between Emfit data and EMG results. Sensitivities and specificities were computed for PLMI and PMI levels of 5 and 15 movements/h. Additionally, receiver operating characteristic (ROC) curves were derived and the area under the curve (AUC) was calculated.
Results: The Spearman's correlation coefficient between the PMI of Emfit and the PLMI of EMG was 0.87. The sensitivity of the Emfit sensor to detect periodic limb movements was 0.91 at the level of 5 movements/h and 0.73 when the cut-off level was 15 movements/h. The specificities were 0.75 and 1.00, respectively. AUC in ROC analysis was 0.96 and 0.98 in the levels of 5 and 15 movements/h.
Conclusions: The results suggest that the Emfit sensor might be suitable for screening of periodic limb movements even if the sensor is placed under the thoracic area of the patient in sleep-disordered breathing studies.

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1. Introduction

Periodic limb movements (PLMs) are relatively common findings in polysomnograms. Around 70–90% of restless legs syndrome (RLS) patients have PLMs in their sleep recordings (Montplaisir et al., 1997). PLMs are quite common also among snorers and obstructive sleep apnea syndrome (OSAS) patients (Coleman et al., 1980; Chervin, 2001; Haba-Rubio et al., 2005) and during continuous positive airway pressure treatment (Fry et al., 1989). Besides, PLMs can be found in polysomnograms without other findings (Bixler et al., 1982; Polo-Kantola et al., 2001; Carrier et al., 2005).

PLMs are defined to last from 0.5 to 10 s and occur in a series of at least four movements at intervals of 5–90 s (Zucconi et al., 2006). The standard method for recording PLMs is the anterior

tibialis electromyography (EMG) (Zucconi et al., 2006), but many other methods, for instance activity monitors, piezoelectric devices, stretch sensors, video etc. are used (Hening, 2004). Actigraphs are used increasingly for PLM studies and recent standards for recording and scoring PLMs also include guidelines for using motion detector systems for activity monitoring of leg movements (Zucconi et al., 2006). In our previous study we found that the static-charge-sensitive bed (SCSB, Alihanka et al., 1981), which is widely used in Finland to monitor nocturnal breathing, is also suitable for detecting periodic movement activity (Rauhala et al., 1996). Lately in many sleep laboratories the SCSB has been replaced by a new movement sensor, Emfit (ElectroMechanical Film) because it is flexible and can be used with a regular mattress.

We use Emfit as part of polysomnography and our ambulatory cardio-respiratory polygraphy systems. Emfit is elastic, permanently charged ferro-electret plastic film that converts mechanical stress into proportionate electrical charge and conversely, it mechanically expands when voltages of opposite polarities are applied (Paajanen et al., 2000). When used to study sleep-disordered breathing the small Emfit sheet is placed under the

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thoracic area. Emfit has been shown to be useful in the analysis of breathing movements (Alametsä et al., 2003) and we have noticed that so called SCSB spiking phenomenon, which is related to partial airway obstruction during sleep, is also present with Emfit (Alametsa et al., 2006). We have also previously shown that a prolonged spiking phenomenon in the Emfit sensor is related to an increase in transcutaneous carbon dioxide and flow limitation pattern in a nasal pressure transducer (Rauhala et al., 2007).

Emfit is a highly sensitive movement sensor and also distant movements in limbs cause changes in the Emfit signal even if the sensor is placed under the thoracic area. We have observed in routine clinical work that in many patients periodic limb movements induce pronounced stereotyped deflections in the Emfit signal. Therefore the aim of the present study was to investigate the usability of a small Emfit sensor in revealing periodic movements.

2. Subjects and methods

2.1. Subjects

Twenty seven consecutive patients (18 men and 9 women) referred to the sleep laboratory of the Pirkanmaa Hospital District volunteered to participate in this study. All the subjects gave their informed consent. The study was approved by the Ethical Committee of the Pirkanmaa Hospital District. The reasons for referrals were suspicion of restless legs syndrome for five patients, control of continuous positive airway pressure treatment for four patients and possible sleep-disordered breathing for 18.

2.2. Methods

Embla N7000 and Somnologica software (Medcare, Iceland) were used as a recording system. The recording montage consisted of six EEG derivations (Fp1-A2, Fp2-A1, C3-A2, C4-A1, O1-A2, O2-A1), two EOG channels and submental electromyography, electrocardiogram, nasal pressure transducer, thermistor, thoracic and abdominal respiratory movements, body position, blood oxygen saturation and pulse by pulseoximetry, anterior tibialis muscle electromyography and the Emfit signals. The sampling rate of 1 Hz was used for pulseoximetry and 200 Hz for all other parameters.

In the present study an Emfit sensor with dimensions of $32\,cm \times 62\,cm$ was placed below a bed-sheet under the thoracic area of the sleeping patient (Fig. 1). Emfit was connected to a bipolar channel of the recorder. The signal was amplified and filtered into the high frequency (6–16 Hz) band analogously with the SCSB filtering (Polo, 1992; Kirjavainen et al., 1996).

Fig. 1. Emfit movement sensor placed on an ordinary mattress below a bed-sheet.

2.3. Visual analysis

The recordings were scored into sleep stages according to the Rechtschaffen and Kales (1968) criteria. Periodic movements in EMG and Emfit channels were scored blindly and independently by two neurophysiologists using the time window of 4 min/screen. Scorings of EMG recordings were performed using the World Association of Sleep Medicine (WASM) standards (Zucconi et al., 2006). The same principles were also used in the Emfit scorings following the leg activity monitoring standards in the WASM guidelines. PLMs were defined as repetitive muscle jerks lasting from 0.5 to 10 s, separated by an interval ranging from 5 to 90 s, with an EMG amplitude rise of 8 μV above resting level, organized in series of 4 or more consecutive leg movements. In the Emfit scoring the threshold of movement detection was set as double the basic Emfit signal amplitude, which seems to correspond well with the anterior tibialis EMG amplitude increase of $\geq 8\,\mu V$ (Fig. 2). After the independent movement scorings the two neurophysiologists formed together the EMG and Emfit consensus scorings which were used in further analyses.

To compare with Emfit the movements in the right and left EMG channels were combined and bilateral movements were counted as one. The movements occurring on the two sides were considered bilateral when they were overlapping or separated by less than 0.5 s (Zucconi et al., 2006). The total number of single and periodic movements were tabulated and the periodic limb movement index (PLMI) for EMG and the periodic movement index (PMI) for Emfit
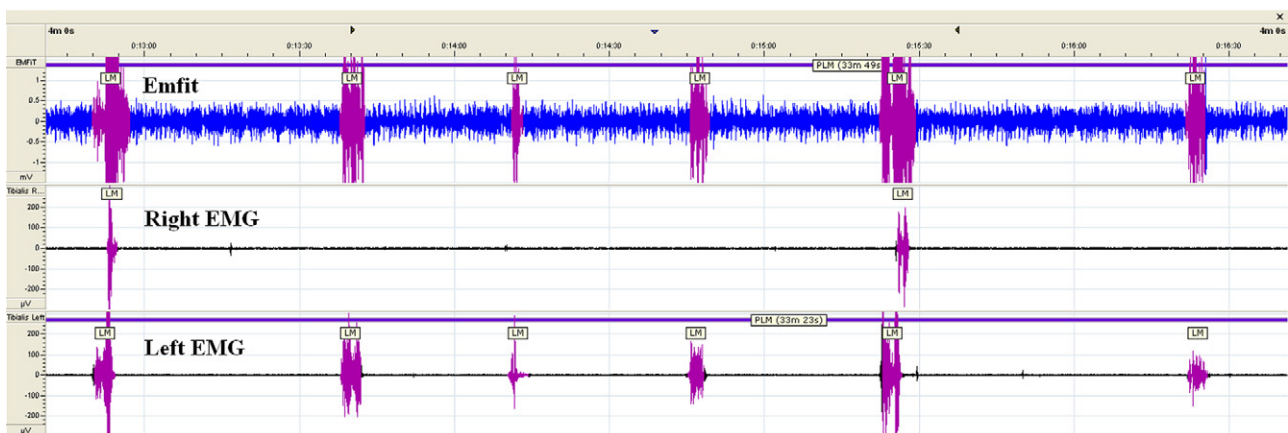

Fig. 2. An example of a 4 min polysomnography page presenting periodic movements in Emfit and left EMG channels. Two simultaneous single movements in right EMG channel. Durations of the movements in this sample were 1–7.5 s.

Table 1
Demographic data and sleep parameters of the 27 subjects.

	Mean	SD	Minimum	Maximum
Age (years)	41.6	11.2	21	58
BMI (kg m^{-2})	28.4	4.7	19.0	38.1
TIB (min)	475	76	286	598
TST (min)	408	81	228	531
SEI (%)	85.7	7.8	66.3	98.6
WASO (min)	56.1	32.4	6.5	163.0
SL (min)	11.5	9.6	0.5	40.0
REMlat (min)	145.0	77.6	12.5	334.5
S1 (%)	9.6	6.1	2.5	22.8
S2 (%)	64.1	8.2	45.5	79.0
S3 (%)	8.4	4.8	0.0	24.4
S4 (%)	1.2	2.8	0.0	10.3
SREM (%)	16.8	6.2	4.4	26.2
ARI (no/h)	16.2	10.3	4.2	38.7
AHI (no/h)	12.0	13.8	0.1	59.0
SaO$_2$min (%)	88.2	4.9	68	93
ODI4 (no/h)	6.0	10.3	0.0	46.0

BMI, Body mass index; TIB, Time in bed; TST, Total sleep time; SEI, Sleep efficiency index; WASO, Wakefulness after sleep onset; SL, Sleep latency; REMlat, latency to the first REM-sleep period; S1, SREM, percentage of the sleep stage referred to TST; ARI, arousal index; SaO$_2$min, minimun oxygen saturation; ODI4, oxygen desaturation $\geq 4\%$ index; SD, standard deviation.

were counted. Additionally, a software was developed for comparing the selections of single and periodic movements within Emfit and EMG channels.

2.4. Statistical analysis

The Spearman's correlation coefficient was used to assess the relationship between the Emfit data and the EMG results. The previous version of the International Classification of Sleep Disorders (ICSD) recommended a PLMI ≥ 5 to determine clinical significance (American Sleep Disorders Association, 1997) and the new ICSD recommended a cut-off of PLMI ≥ 15 (American Academy of Sleep Medicine, 2005). Therefore sensitivities and specificities were computed for PLMI and PMI levels of 5 and 15 movements/h. Based on these threshold definitions, receiver operating characteristic (ROC) curves were derived and an area under the curve (AUC) was calculated. The Bland–Altman method of concordance was used to assess potential range-dependent agreement.

3. Results

The sleep parameters derived from the polysomnograms of the 27 subjects are shown in Table 1.

Four subjects presented no significant findings in their polysomnograms. One subject had catathrenia. Fourteen subjects had sleep-disordered breathing and three of them had abundant periodic limb movements. Four subjects had only periodic limb movements and four studies were CPAP controls.

The number of selected single leg movements during sleep in combined EMG varied from 38 to 742 (mean 222) movements per night. Emfit presented 32–666 (mean 193) single movements per night. The corresponding numbers of periodic movements ranged from 4 to 717 (mean 172) and from 0 to 632 (mean 136), respectively. The corresponding mean periodic movement indexes were 27.6/h for EMG and 23.0/h for Emfit. With wake included the number of single leg movements in combined EMG ranged from 55 to 801 (mean 290) movements. Emfit detected 56–714 (mean 233) movements. The corresponding mean periodic movement indexes were 31.4/h for EMG and 23.8/h for Emfit.

The Spearman's correlation coefficient between the PMI of Emfit and the PLMI of EMG during sleep was 0.87 (Fig. 3) and 0.88 with wake included.
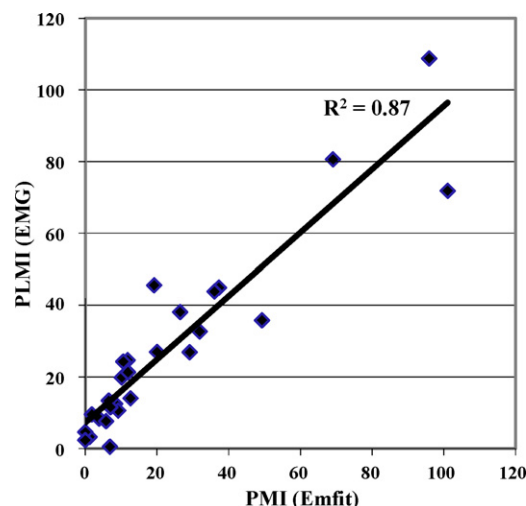

Fig. 3. Scatterplot showing the relationship between the PLM index (EMG) and the PM index (Emfit) during sleep.

The sensitivity of the Emfit sensor to detect criterial numbers of periodic limb movements in sleep was 0.91 at the cut-off level of 5 movements/h and 0.73 when the cut-off level was 15 movements/h. The specificities were 0.75 and 1.00, respectively. The positive predictive value for the cut-off level of 5 movements/h was 0.95 and for the level of 15 movements/h 1.00. The negative predictive values were 0.60 and 0.75, respectively. The results taking account also movements during wakefulness were about the same. At the cut-off levels of 5 and 15 movements the sensitivities were 0.88 and 0.65 and specificities 0.67 and 1.00, respectively.

Fig. 4 shows the ROC curve reflecting the diagnostic capability of Emfit sensor when the threshold of the PLM index was set at (a) 5 movements/h and (b) 15 movements/h for diagnosis of the PLM disorder, with an area under curve (AUC) of 0.96 and 0.98.

The Bland–Altman method showed that Emfit underestimates motor events compared to combined EMG (mean PLMI/PMI difference of 4.5 1/h, standard deviation of 9.8 1/h (Fig. 5).

4. Discussion

The Emfit movement sensor has proven to be valuable in clinical work in characterizing sleep-disordered breathing and therefore it is widely used in Finland. The results of the present study suggest that while polygraphy is set up to study sleep-disordered breathing, also periodic movements could be identified by the Emfit sensor. In other words, screening of periodic movements seems possible with Emfit even if the small-sized sensor is placed under the thoracic area of the patient. Presumably limb movements also slightly move the whole body causing movement detections in a rather small-sized but sensitive Emfit sensor. This option to discover periodic limb movements in recordings focusing on sleep-disordered breathing comprises a useful additional feature of Emfit.

The sensitivity of the Emfit sensor to detect periodic limb movements in sleep was good and the specificity and the positive predictive value at the cut-off level of 15 movements/h were 1.00. Hence all the patients with indexes over 15/h, which is considered to determine clinical significance (American Academy of Sleep Medicine, 2005), were classified correctly. The ROC curves with AUCs of 0.96 and 0.98 showed that the capability of detecting periodic movements of the Emfit sensor is very good.

Because of the characteristics of bed movement sensors, they detect all the movements in limbs, trunk or head, and this also interacts with results in the present study. The ability to record all the movements can be advantageous in a clinical setting, because in
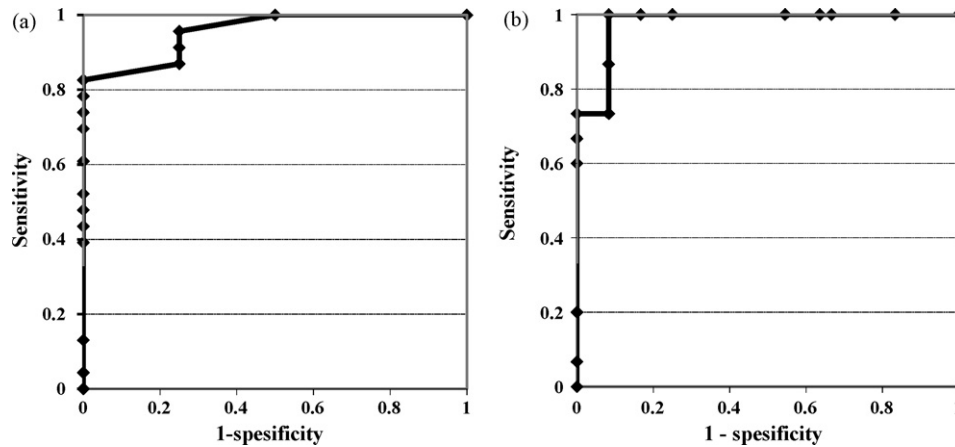
Fig. 4. Receiver operating characteristic (ROC) curve for identifying PLM indexes (a) 5 movements/h and (b) 15 movements/h based on EMG vs Emfit results.

addition to quantifying periodic movements, it gives an estimate of peacefulness or restlessness of sleep (Kronholm et al., 1987). The fact that EMG electrodes actually record muscle activity affects the results when compared with movement detecting sensors. It is evident that mere tonic muscle contractions without proper movements are not visible in Emfit. However, it is questionable if this EMG activity or subtle movements is of clinical significance as discussed in the review of Hening (2004).

Sleep disorders are very common and there is a constant need to develop simple but reliable ambulatory devices (Flemons et al., 2003). The advantage of the Emfit sensor in sleep studies is that no electrodes have to be attached to patients. Emfit can be used as a part of polysomnography and also as a stand-alone device. The matter that the device does not interfere with sleep emphasizes its suitability for ambulatory sleep recordings. Emfit is made of flexible material, any size or shape of sensor can be used, it is easy to carry and thus also easy to adapt to ambulatory recordings. Long EMG cables are always in danger to loose contact whereas Emfit without any electrode connection to the patient could provide more reliable and possibly wider insight into motor activity during night (Kronholm et al., 1993).

Ambulatory recordings with an Emfit sensor are easy to perform and recordings are inexpensive compared to full polysomnography. More patients can be studied with reasonable costs and recordings with an Emfit sensor can be repeated during several consecutive nights if necessary. PLMI and subjective symptoms in RLS patients have been shown to correlate quite well (Allen and Earley, 2001; Garcia-Borreguero et al., 2004) and thus PLMI could be used as a quantitative measurement for the severity of the disorder. Emfit could also be used for the evaluation of treatment efficacy in RLS and PLM disorder patients.

Distinguishing sleep and wake is not fully possible with a movement sensor (Morgenthaler et al., 2007). Therefore measuring movements with Emfit without a full polysomnogram is a combination of periodic movements during sleep (PLMS) and wakefulness (PLMW). However, it has been presented that including PLMW in the analyses may even provide a better impression of the movement disorder during night (Michaud et al., 2002). In this study Emfit performed well in detecting periodic movements during both sleep and wakefulness. Furthermore some investigators emphasize that counting both movements with and without arousal is very important (Claman et al., 2006; Hornyak et al., 2006) but this matter is still controversial (Chervin, 2001; Al-Alawi et al., 2006).

Emfit's performance in screening of PLM seems to be comparable with actigraphy (Sforza et al., 2005) which has lately been used increasingly for PLM studies. Actigraphy is also inexpensive and handy but it has its own limitations. Sforza and co-workers recommend that actigraphy should be used only in patients with clinical history suggesting the presence of an RLS syndrome and without sleep-disordered breathing. Sleep-disordered breathing and simultaneous periodic movements in sleep studies are indeed rather common findings (Haba-Rubio et al., 2005) and problems in their differentiation can be controlled better by Emfit because it is also used to detect nocturnal breathing disorders. Basically the Emfit movement sensor provides information about breathing movements and movements overall being very suitable for recordings where reducing the amount of electrodes is desirable. Hence the Emfit sensor can be used in cardio-respiratory polygraphy for characterizing sleep-disordered breathing and moreover for screening periodic movements.

Acknowledgements

Fig. 5. Bland–Altman plot showing the distribution of the differences between PM index (Emfit) and PLM index (EMG) against the mean of PLM and PM indexes. The dashed lines represent ±2 SD levels.

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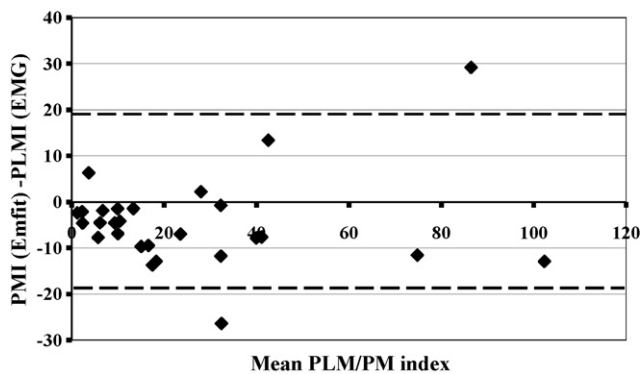
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Technical note

Automatic detection of spiking events in EMFi sheet during sleep

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Abstract

In this paper we present a new method for detection of spiking events caused by the increased respiratory resistance (IRR) from ballistocardiographic (BCG) data recorded with EMFi sheet. Spiking is a phenomenon where BCG wave complexes increase in amplitude during IRR. In this study data from six patients with a total of 1503 visually scored spiking events were studied. The algorithm monitors amplitude levels of BCG complexes and detects large relative increases. In this work 10 different variations of the algorithm were compared in order to find the best variation, which can cope with different recordings. The best variation of the algorithm was able to detect spiking events with 80% true positive and 19% false positive rates. The detection is not dependent on absolute waveform amplitudes and therefore does not require any recording-specific tuning prior to application. It is important to recognize spiking events in order to evaluate the severity of respiratory disturbance during sleep.
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Keywords: Spiking events; Detection; Sleep research; Apnea; Ballistocardiography

1. Introduction

Good vigilance is the fundamental basis for effective daily function. Poor sleep quality can cause a lot of trouble in daytime cognitive operations by undermining the quality of life. The impact of sleep disorders on general health is obvious [1]. Patients with respiratory disturbances during sleep have a higher risk of stroke, coronary artery disease, hypertension and myocardial infarction [2,3].

During sleep the muscles of the pharyngeal airway lose their tension causing narrowing of the airways and resistance of respiration increases even in non-snoring persons. When narrowing continues, snoring sounds caused by vibrating structures of pharyngeal airway appear. The upper airway dysfunction during sleep is a concept which ranges from partial upper airway collapse with increased upper airway resistance (IRR) associated usually with snoring to complete airway collapse with apnea episodes leading to increased inspiratory efforts [4].

Ballistocardiography (BCG) [5] means recording of the movements of the body caused by the pumping action of the heart. It represents movements that arise from heart contraction and displacement of blood in large vessels. Venous return into the right ventricle and right ventricle filling is increased during inspiration because of increased negative intrathoracic pressure leading to reinforcement of systolic recoil movement in the body [6]. As a result, the maximal amplitudes of the ballistocardiographic deflections are slightly increased during normal inspiration. The IRR or increased intrathoracic pressure variation reinforces the appearance of this respiratory variation [6].

Nowadays, the ballistocardiographic signal can be obtained with an EMFi [7–9] sheet and static charge sensitive bed (SCSB) mattress [10]. An example of the BCG waveform recorded with the EMFi sheet can be seen in Fig. 1. An EMFi sensor is basically a thin biaxially oriented plastic film coated with electrically conductive layers, which are permanently

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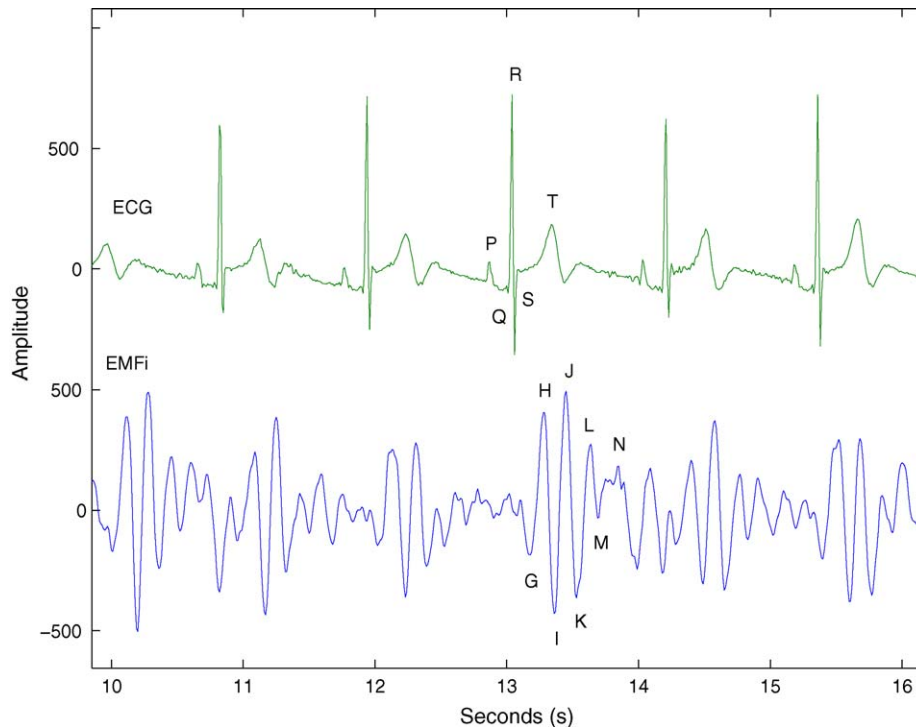*E-mail address:* jalam@cc.tut.fi (J. Alametsä).

Fig. 1. An example of ECG and corresponding (5–30 Hz filtered) EMFi signal segments of 6 s. Characteristic components of BCG can be seen in the EMFi signal so that the G, H, I, J and K denote the systolic deflections and L, M and N the diastolic deflections of the BCG complex [20]. The QRS complex of the ECG can be seen preceding the BCG complex during each heart cycle.

polarized. Changes in the pressure acting on the film generate a charge on its electrically conductive surfaces and this charge can be measured as a current or voltage signal [11]. As an unobtrusive method it can be used to acquire information from the heart activity and breathing patterns. SCSB mattress has been used especially in sleep studies for monitoring cardiorespiratory and motor activities where the need for long-term monitoring in order to detect periodic phenomena like apnea events is apparent [12]. The SCSB signal reveals besides apnea and hypopnea also episodes of IRR, which correspond to prolonged episodes of obstructive hypoventilation and flow limitations without arousals but with carbon dioxide retention [13] and increased intrathoracic pressure variation [14].

Spiking can be seen visually in EMFi sheet or SCSB mattress signal in the frequency range 6–16 Hz as sharp spiking episodes of main systolic complexes of BCG (duration about 0.4 s) with amplitudes much larger than the usual [6]. These spiking episodes repeat in synchrony with respiration being the result of increased inspiratory efforts as a response to partial upper airway obstruction [15]. The SCSB spiking has been called the IRR pattern [6]. Some spiking can be seen in normal recordings but moderate or heavy SCSB spiking has not been observed in normal healthy adult control subjects [6]. IRR is characterized by slowly increasing ballistocardiographic respiratory variation (BRV) accompanied with increased respiratory effort [6]. Spiky systolic waves are increased by the BRV synchronized with

the respiration [16]. Partial upper airway obstruction during sleep induces SCSB spiking in adults and in children [13]. Heavy snorers commonly present episodes of spiking in addition to episodes of apnea or hypopnea on the SCSB signal [15]. Kirjavainen et al. showed that the degree of respiratory effort correlated with the degree of spiking, which correlated with intrathoracic pressure variation during obstructive breathing in the SCSB signal. Also the amplitude of SCSB spiking complexes correlated highly with breathing frequency and variation on the oesophageal pressure, which is the most direct measure of respiratory effort [15].

There are very few previous studies on automatic detection of spiking events. The detection of high-frequency respiratory movement (spiking) patterns from SCSB signal which are related to increased respiratory efforts, was studied by four automatic methods in one work [17]. There were only two subjects in that study, with 1268 visually scored spiking events judged by an expert. The first and the second methods detected the local changes (outliers) in the signal amplitude, resulting in 74% and 89% sensitivity and 94% and 92% specificity. The third and the fourth methods detected overall changes in the spectral characteristics of the signal. The third method assumed that the signal model is known a priori, and it was based on autoregressive modelling whereas the fourth approach was non-parametric, resulting in 84% and 66% sensitivities and in 89% and 94% specificities. These methods required a separate training phase to determine appropriate

Table 1
Parameters of the patients

Subject	Patient 1	Patient 2	Patient 3	Patient 4	Patient 5	Patient 6
Episode duration[a]	629	1575	954	1268	632	935
Spiking events[b]	146	204	405	257	151	340
Gender[c]	Male	Female	Female	Male	Male	Male
Age[d]	44	58	42	47	46	43
BMI[e]	28.8	38.8	38.1	NA	25.4	23.0
TIB[f]	487.5	480.5	434.5	495.0	512.5	484.0
TST[g]	470.0	431.0	359.0	434.5	427.0	463.0
SEI[h]	96	90	83	88	83	96
S1[i]	2.2	8.2	5.2	6.1	13.8	6.3
S2[j]	66.9	72.7	60.7	70.7	68.0	74.2
S3[k]	7.0	0.1	10.3	6.8	5.6	3.8
S4[l]	0	0	2,8	0,2	0	0
REM[m]	23.8	18.9	20.8	16.1	12.5	15.8
SL[n]	3.5	9.5	40.5	7.5	5.5	4.0
REM Lat[o]	72.5	99	102	110	138	101
ARI[p]	23.6	27.0	9.2	26.7	17.7	24.0
AHI[q]	27	53	15	38	13	39
AI[r]	19	43	6	16	7	16
ODI4[s]	27	40	14	12	5	19
SaO2min[t]	81	79	77	83	86	85
SaO2mean[u]	93.2	93.3	94.6	95.6	95.6	94.8

[a] Duration of visually inspected episode (s).
[b] Number of visually scored spiking events.
[c] Gender of the subject.
[d] Age of the subject (years).
[e] Body mass index (kg/m^2).
[f] Time in bed (min).
[g] Total sleep time (min).
[h] Sleep efficiency index, TST/TIB (%).
[i] Proportion of sleep stage referred to TST (%).
[j] Proportion of sleep stage referred to TST (%).
[k] Proportion of sleep stage referred to TST (%).
[l] Proportion of sleep stage referred to TST (%).
[m] Proportion of sleep stage referred to TST (%).
[n] Sleep latency (min).
[o] REM latency (min).
[p] Arousal index (arousals/h).
[q] Apnea-hypopnea index (events/h).
[r] Apnea index (events/h).
[s] Oxygen desaturation index ($\geq$4% events/h).
[t] Minimum arterial oxygen saturation during the night (%).
[u] Mean arterial oxygen saturation during the night (%).

parameter values before they could be applied in signal analysis.

The purpose of our work was to develop an algorithm to detect the increase in the amplitude of BCG wave complexes during upper airway obstruction or during IRR in sleep.

2. Methods

2.1. Patient data

Six patients with nocturnal respiratory disturbances were studied (Table 1). All recordings were made in the Sleep Laboratory of the Department of Clinical Neurophysiology in Tampere University Hospital, Finland. The bedtime in the patients was from 23.00 to 07.00 h on average. A BCG signal was measured with an EMFi sheet below the mattress. The

data were converted into the European data format (EDF) [18]. The sampling rate was 200 Hz in two recordings and 100 Hz in four recordings, which were re-sampled to 200 Hz. The mean age of the patients was 46.7 years and the mean body mass index (BMI) was 30.8 kg/m^2.

From each recording one episode of frequent spiking was selected for the analysis. The total duration of these episodes was 5993 s. From these selections individual BCG spiking complexes were visually scored by an experienced clinical neurophysiologist. The total number of visually scored spiking events was 1503.

2.2. Detection algorithm

The amplitude of the systolic BCG complex increases when the spiking phenomenon occurs during IRR. This point
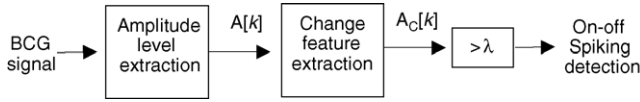
Fig. 2. Block diagram of the developed spiking detector. Letter k indicates time in seconds, $A[k]$ is the amplitude level feature and $A_c[k]$ is the change feature, formed on the basis of history of 60 s. The detector output is obtained with 1-s time resolution.

was used in the developed features which quantify the amplitude increases. The complexes of the cardiac cycle can be studied by using the frequency range of BCG (2–15 Hz frequency area), in which the main systolic components occur [12]. We used the frequency band of 6–16 Hz, which has been earlier found to be suitable in visual spiking analysis [16]. Additionally, also the 3–16 Hz band was used for comparison purposes. All features were calculated with 1 s time resolution. The overall scheme of the detection method is illustrated in Fig. 2.

2.3. Amplitude level extraction

In this work, two different fixed-length windowing approaches were applied. First, we simply used non-overlapping 1 s long Hanning windowed BCG signal segments. Each segment was zero-padded to length of 512 and the FFT was taken of the entire sequence. The frequency resolution of the spectrum was then $\Delta f = 0.39$ Hz. The resulting FFT estimate was properly scaled to get corresponding amplitude spectrum, denoted as $S_{one}[f]$. Then, mean of spectrum values $S_{one}[f_{band}]$ was extracted, where f_{band} denoted the frequency band included. For each 1 s segment, the resulting amplitude value is denoted as $A_{mean}[k]$, where k refers to time index. In the present analyses, frequency bands of 3–16 and 6–16 Hz were used.

Secondly, due to the fact that spiking events have typically duration of about 0.4 s, we used also shorter Hanning windows, in which five partly overlapping data segments (length of 0.5 s of each) centered at 0.1, 0.3, 0.5, 0.7 and 0.9 s were extracted from a 1.3 s signal segment. This was intended to give better time localization than simple 1 s segmentation described above making it possible to explore the properties of the signal components during different parts of the heart cycle. FFT with zero-padding to length of 512 taken from these five sub-windows were properly scaled to provide five corresponding amplitude spectra, denoted as $S_1[f], \ldots, S_5[f]$. Then, from each five spectra the maximum and mean values of $S_1[f_{band}], \ldots, S_5[f_{band}]$ were extracted, where f_{band} denoted the frequency band included (3–16 or 6–16 Hz). These values were denoted as $max_1, \ldots, max_5$ and $mean_1, \ldots, mean_5$, reflecting the BCG amplitude levels somewhat differently inside the five windows.

Finally, maximum and minimum of $max_1, \ldots, max_5$ and $mean_1, \ldots, mean_5$ were selected, denoted as $A_{max\_max}[k]$, $A_{max\_mean}[k]$, $A_{min\_max}[k]$ and $A_{min\_mean}[k]$. The maximum values ($A_{max\_max}[k]$ and $A_{max\_mean}[k]$) reflected the amplitudes during the peaking of the BCG complex, whereas the minimum values ($A_{min\_max}[k]$ and $A_{min\_mean}[k]$) reflected the valley between them.

2.4. Change feature extraction

By using the five versions of the extracted amplitudes above, different variations of change features could be obtained. Following notations are used to present the change feature calculus:

$A[k]_A = A_{mean}[k]$, in submethod A (1 s time resolution),
$A[k]_{B,C} = A_{max\_max}[k]$, in submethods B and C (0.5 s time resolution),
$A[k]_{D,E} = A_{max\_mean}[k]$, in submethods D and E (0.5 s time resolution).

For the calculation of change feature values, five versions of moving local base lines, denoted as $B[k]_{A,\ldots,E}$, were determined based on a 60 s history window:

$B[k]_A = \text{median}(A_{mean}[k-60:k-1]))$,
$B[k]_B = \text{median}(A_{min\_max}[k-60:k-1]))$,
$B[k]_C = \text{median}(A_{max\_max}[k-60:k-1]))$,
$B[k]_D = \text{median}(A_{min\_mean}[k-60:k-1]))$,
$B[k]_E = \text{median}(A_{max\_mean}[k-60:k-1]))$.

The different versions of amplitude $A[k]$ describe the peaking of BCG signal slightly differently. The different versions of $B[k]$ reflect the local base line of the amplitude in different phases of the BCG complexes inside the 60 s history window. The change features $B[k]_{C,E}$ reflect the amplitude level during the peaking of the BCG complex, whereas $B[k]_{B,D}$ reflect the valley of the BCG complex. By combining different versions of $B[k]$ and $A[k]$, five variations of change features were extracted as follows:

$$A_c[k]_{A,\ldots,E} = \frac{A[k]_{A,\ldots,E} - B[k]_{A,\ldots,E}}{B[k]_{A,\ldots,E}} \times 100\%$$

The change features were extracted using the two frequency bands of the spectra $S_1[f_{band}], \ldots, S_5[f_{band}]$, ($f_{band} = 3$–16 and 6–16 Hz) totalling 10 methods. Using the frequency range 3–16 Hz, the change feature $A_c[k]_A$ is denoted as method 1, $A_c[k]_B$ method 2, $A_c[k]_C$ method 3, $A_c[k]_D$ method 4 and $A_c[k]_E$ method 5. Using the frequency range 6–16 Hz, $A_c[k]_A$ corresponds to method 6, $A_c[k]_B$ method 7, $A_c[k]_C$ method 8, $A_c[k]_D$ method 9 and $A_c[k]_E$ method 10. A BCG spiking event was detected when $A_c[k]$ exceeded the threshold value λ.

2.5. Spiking detection performance evaluation

A true positive finding was calculated if the spiking event (change feature exceeding the threshold value λ) was detected simultaneously with a visually scored spiking event. If a spiking event was detected and there was no scoring of spiking, a false positive finding was counted. No false positive findings were calculated one second prior and after each scoring. This

was done because a scored spiking event near second border could be detected also in adjacent second. In this way in all data there were 2504 non-spiking seconds. Also, after one false finding there was 1 s latency time during which no additional false positive findings were counted. This way in all data there were 1452 non-spiking seconds. The true positive rate (sensitivity) was the number of detected spiking events divided by the total number of spiking events. The false positive rate was the number of false positive findings divided by the non-spiking seconds.

In order to evaluate the performance of the various spiking detector versions, receiver operating characteristics (ROC) curves were determined, depicting the true positive rate as a function of the false positive rate. The ROC curve was determined by using a range of threshold values for each feature version. The more to the top left corner a point on the ROC curve the better is the detection performance.

3. Results

Fig. 3 depicts an example of EMFi signal with scored spiking events and corresponding change features $A_c[k]$. The outcome of performance comparison of different feature
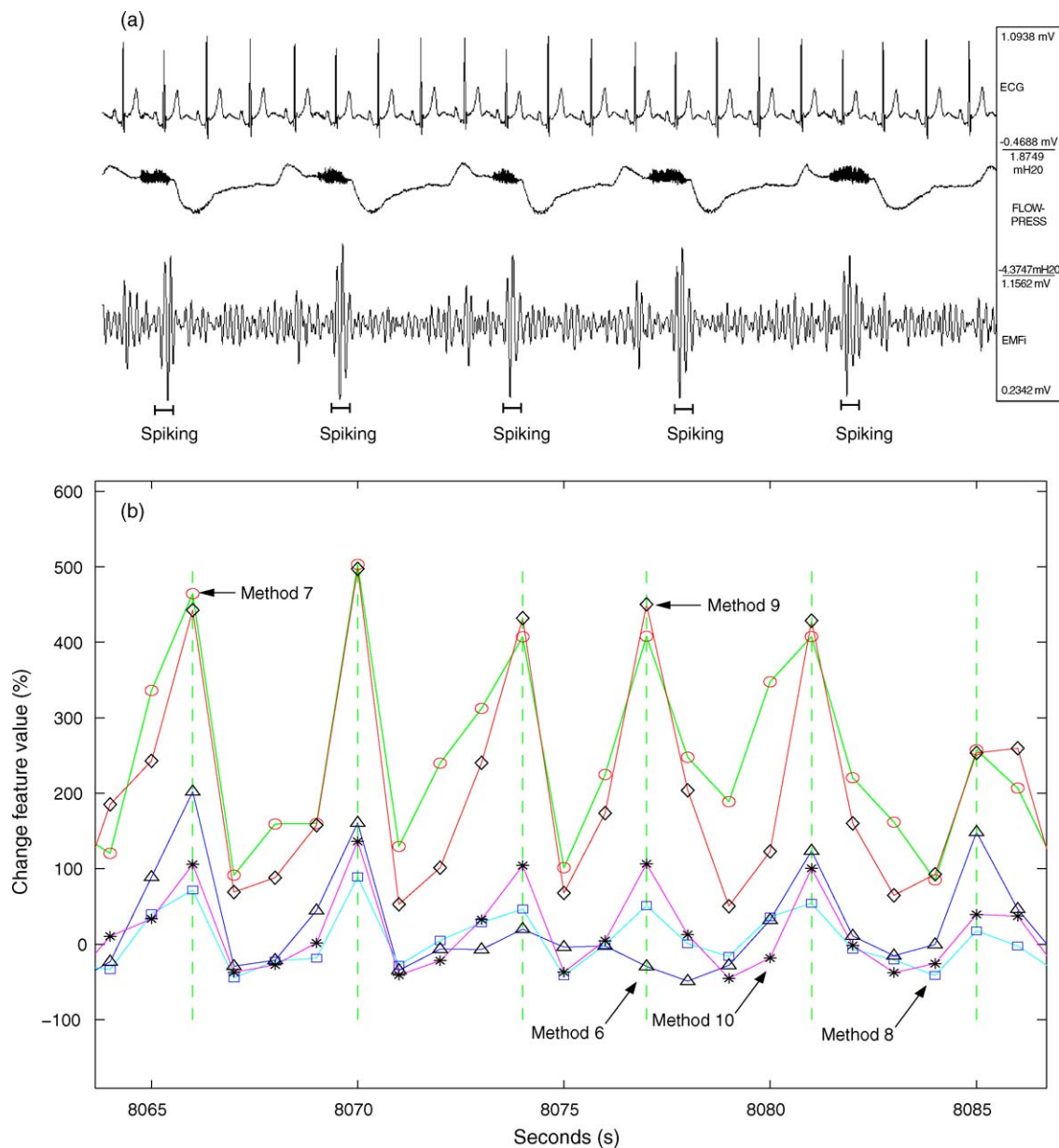
Fig. 3. An example of ECG, nasal flow pressure and (6–16 Hz filtered) EMFi signal segments of 20 s with scored spiking events (a). Occurrence of spiking seems to correlate with the flat flow pressure signal with high-frequency activity. Corresponding change feature $A_c[k]$ values of methods 6–10 are seen in (b). Dashed vertical lines indicate the time instants of spiking events. Method 6 is indicated with triangles, method 7 with circles, method 8 with squares, method 9 with diamonds, method 10 with stars. All these versions of change features seem to show large increases at the times of spiking events. Different versions are seen to act with different dynamics, as intended.
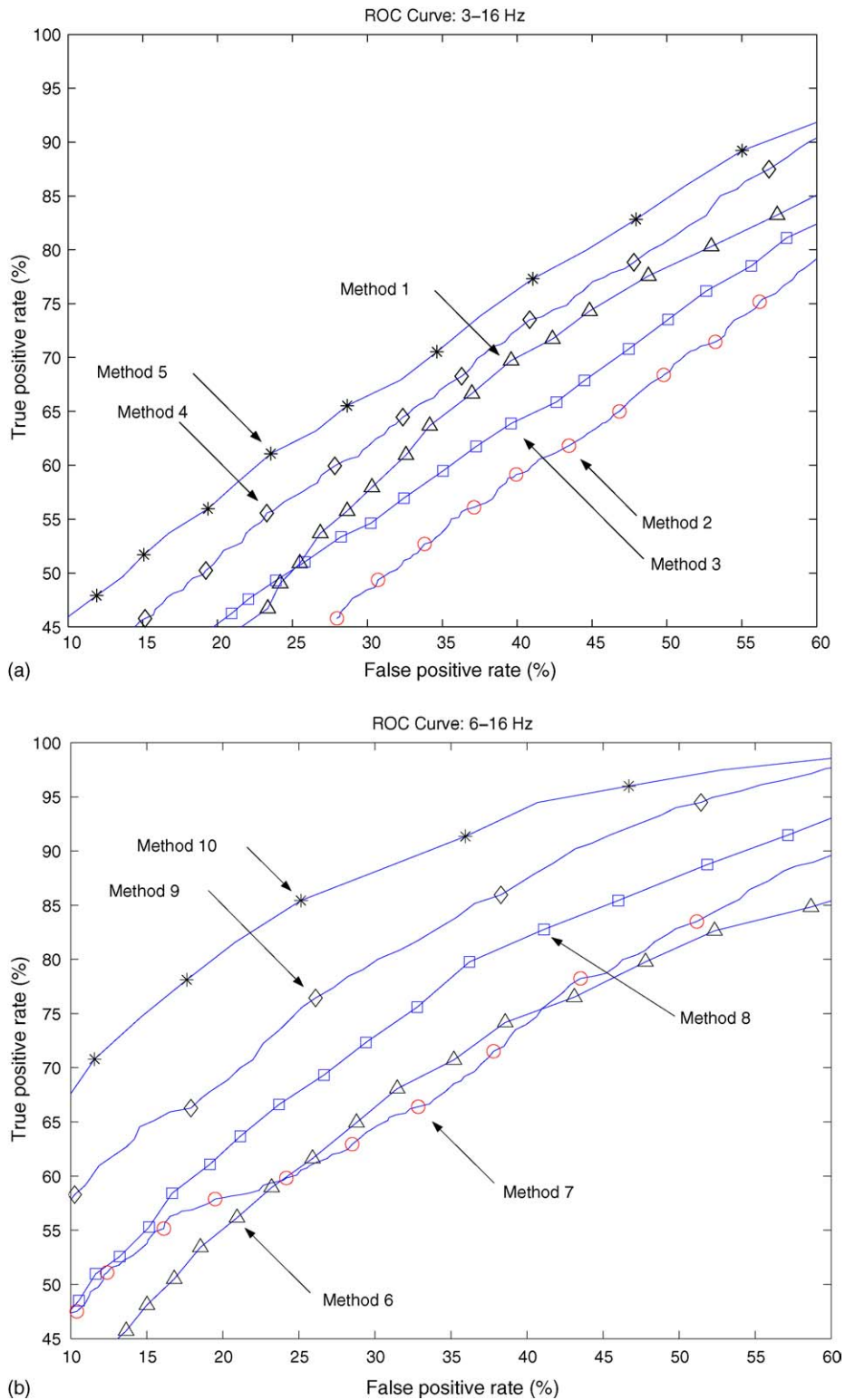
Fig. 4. Comparison of spiking detection outcome from all six night recordings obtained with different feature versions, with frequency range 3–16 Hz (a) and 6–16 Hz (b). ROC curves are labelled so that method 1 is indicated with triangles, method 2 with circles, method 3 with squares, method 4 with diamonds and method 5 with stars (a). Similarly, method 6 with triangles, method 7 with circles, method 8 with squares method 9 with diamonds, method 10 with stars (b).
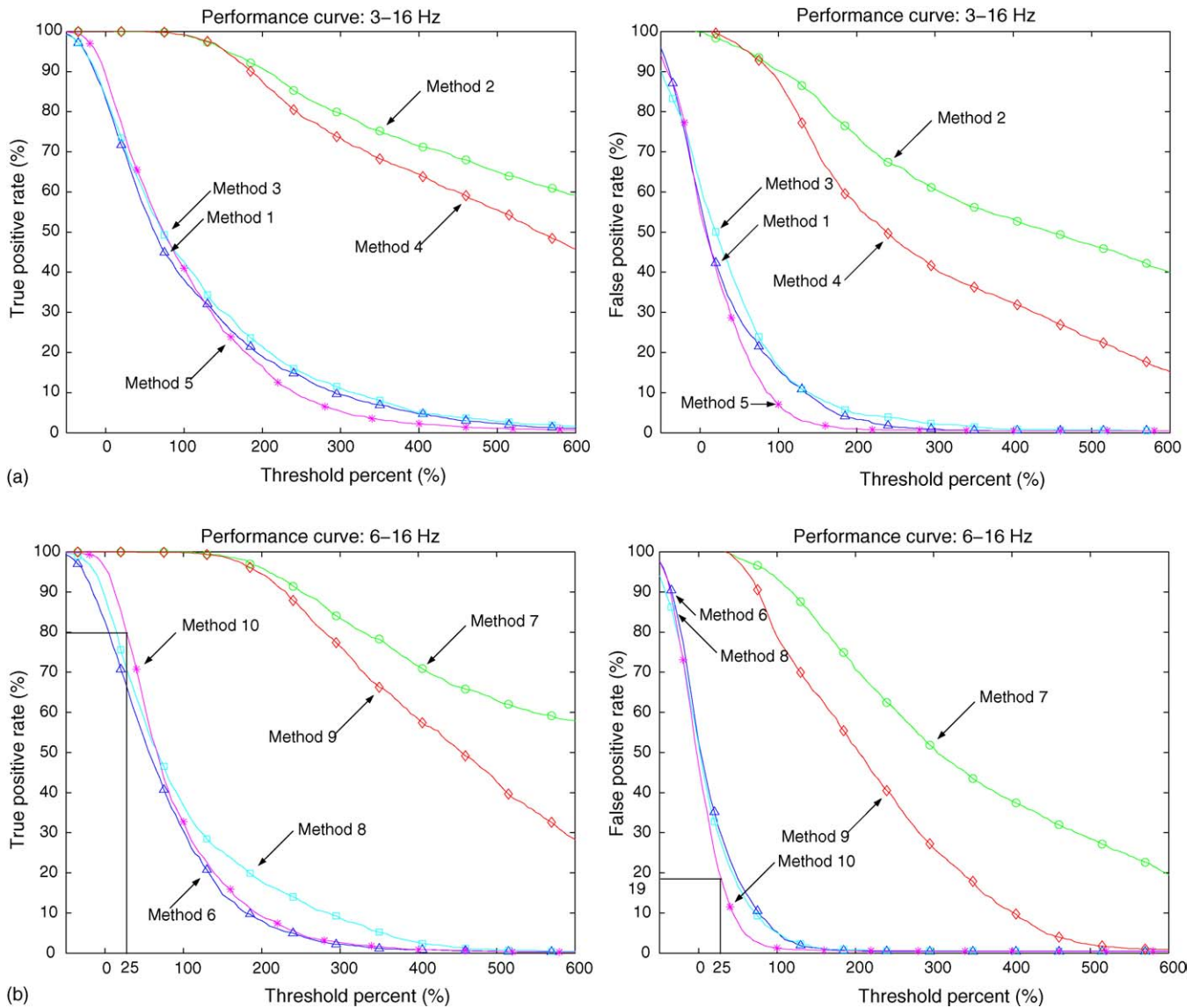
Fig. 5. Performance curves of different feature versions from all six night recordings from the frequency range 3–16 Hz (a) and 6–16 Hz (b); true positive rate on the left picture and false positive rate on the right picture. Method 1 is indicated with triangles, method 2 with circles, method 3 with squares, method 4 with diamonds and method 5 with stars. Similarly, method 6 is indicated with triangles, method 7 with circles, method 8 with squares, method 9 with diamonds, and method 10 with stars. The effect of threshold value λ into the operation of detection algorithm can be seen in these pictures. By example, analogy between pictures can be seen between performance curves 6–16 Hz and ROC Curve (Fig. 3; b). True positive rate 80% and false positive rate 19% were produced by the same threshold λ of 25%.

versions with ROC curves are depicted in Figs. 4 and 5. It can be seen in the ROC curves (Fig. 4) that features based on the frequency range 6–16 Hz (methods 6–10) provided overall better results than corresponding versions with 3–16 Hz. Method 10 provided the overall best spiking detection performance. Methods 8 and 9 provided better result than methods 1–5. Overall poorest results were seen in method 2.

As examples of spiking detection performance levels in frequency range 6–16 Hz, the (overall best) method 10 provided true and false positive rates of 80% and 19%, while

method 6 provided rates of 80% and 48%, respectively. In a similar way, in the frequency range 3–16 Hz, method 5 produced true and false positive rates of 80% and 45%, while (overall poorest) method 2 produced rates of 80% and 61%, respectively.

Likelihood functions of the best (method 10) and worst (method 2) feature versions during visually scored spiking and non-spiking seconds are shown in Fig. 6. In the best feature version the relative intersection of area under scored and un-scored likelihood functions was smaller (36.5%) than the worst feature version (42.6%).
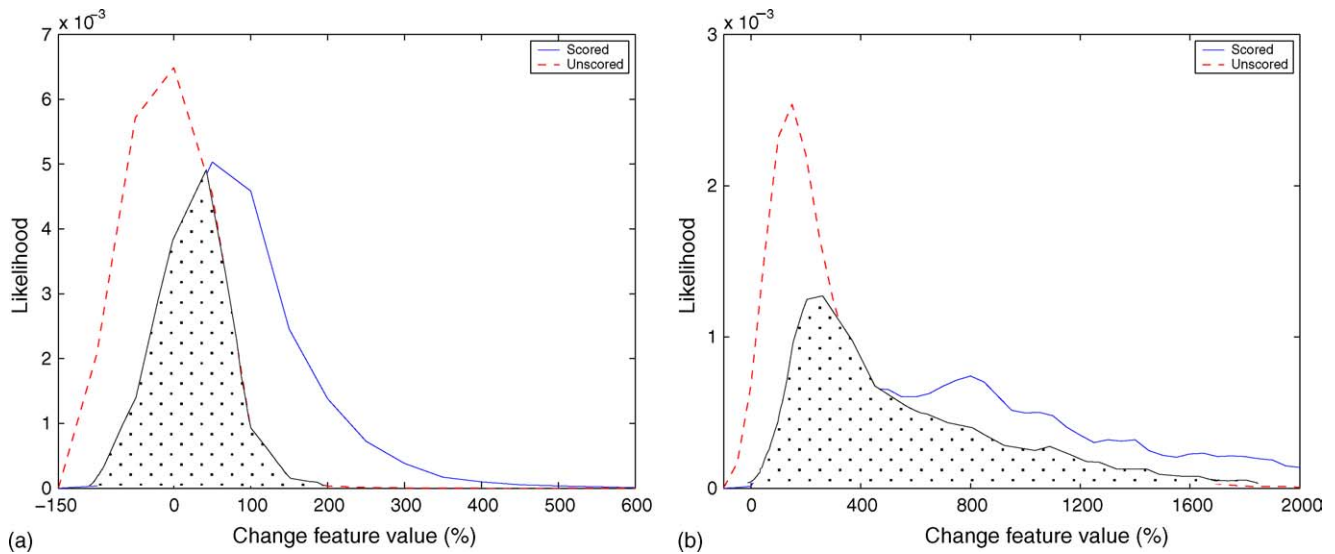
Fig. 6. Likelihood functions of two versions of change features from all six night recordings, method 10 (a) and method 2 (b). Method 10 provided the best results (Fig. 4) and method 2 the worst, which can be seen as smaller overlap (dotted area) between spiking and non-spiking seconds in method 10 than in method 2 (Fig. 3). In method 10 the relative area of dotted are was 36.5% and corresponding area in method 2 was 42.6%.

4. Discussion

In the present work, 10 different feature versions from two frequency bands for spiking detection were compared. All feature versions were independent on absolute signal amplitudes. This is a clear advantage because of the variation of the contractibility of the heart and anatomical differences of individual persons, the normal level of BCG amplitudes may vary. The individual differences might be even greater in patients with respiratory disturbances. All features proved to be rather useful in the detection of the amplitude increase of BCG complex caused by the IRR.

Spiking is a far more transient event (lasting approximately 0.4 s) than apnea event (mean duration of 40 s [19]). Normal systolic BCG complexes as well as spiking (increased amplitude) complexes consist of multiple wave elements going up and down. Therefore, a short-time FFT amplitude spectrum is a good tool in quantifying the amplitude of these complexes. In addition to this, the frequency range could be conveniently selected. In the present work, 0.5 s long overlapping windowing in FFT computation gave better results than a single 1 s window, because of a better time-localisation of BCG complexes with 0.5 s segment. Inside a 1 s segment an event of spiking BCG can locate also near the borders of the segment causing poorer detection of complexes with method 1 or 6. The best results were obtained by using 0.5 s windowing, also used in method 10. Possible reason for that might be that the mean taken first from the five 0.5 s sub-windows emphasizes the high-energy components of the BCG complex. Detecting the increase in the amplitude of the spiking events as compared to normal complexes is a trade-off between false and correct detections. The best detection performance was achieved when maximum was taken from the preserved amplitude spectrum band producing only loca-

tions, in which the greatest amplitude deflections occur in time domain.

In additional testing, different lengths of history window ranging from 10 to 80 s were used. All these lengths of history window seemed to work rather well. The median filtering in the history window provides a robust baseline during frequent spiking and also during normal BCG. Slightly the best results were obtained in this work with the reported 60 s history window. Such a relatively short history window is suitable because the characteristics of BCG signal can change during the night [17].

In this work we applied the well-known FFT approach, which uses sinusoidal waveforms, quite suitable for spiking analysis. There is naturally some uncertainty in the spectrum values extracted from the short signal segments. However, the uncertainty is known to be relatively small for (nearly) deterministic signals, like sinusoids, and therefore it can be assumed insignificant as compared to overall amplitudes of spectral components of BCG. It would naturally be possible to explore possibilities of techniques making use of dictionaries of other waveforms. The matching pursuit approach is the most versatile one, covering sinusoidal and wavelet waveforms [21].

In healthy subjects moderate or heavy spiking rarely occurs during the night. In apnea patients or heavy snorers like in this study, spiking may happen in variable amounts. Partial airway obstruction producing the increased respiratory resistance pattern has shown to be, for instance, in postmenopausal women the predominant breathing abnormality, manifesting about ten times as often as the apnea episodes [4]. The present method can be used to quantify the spiking events in EMFi sheet during the night and thus help to evaluate the impact of partial airway obstruction on sleep.

Acknowledgements

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Prolonged spiking in the Emfit sensor in patients with sleep-disordered breathing is characterized by increase in transcutaneous carbon dioxide

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Abstract

A phenomenon of prolonged spiking in movement sensors, such as static-charge-sensitive bed or Emfit (electromechanical film) sensors, has been connected to an increase in carbon dioxide tension in wakefulness. Spiking is also a common finding in sleep studies. This made us hypothesize that carbon dioxide changes might also happen in sleep during prolonged spiking episodes in Emfit sheet. We examined four different kinds of breathing pattern episodes: normal breathing, episodes of repetitive apnea, episodes of repetitive hypopnea and episodes with prolonged spiking lasting at least 3 min. One hundred and fifteen episodes from 19 polysomnograms were finally admitted to the study according to the protocol. The changes in the transcutaneous carbon dioxide tension ($TcCO_2$) were defined for different breathing patterns. During prolonged spiking episodes the $TcCO_2$ increased significantly and differed statistically from the $TcCO_2$ changes of normal breathing and periodic breathing patterns (episodes of apnea and hypopnea). The rise in $TcCO_2$ during prolonged spiking episodes might suggest that prolonged spiking is representing another type of breathing disturbance during sleep differing from periodic breathing patterns. The Emfit sensor as a small, flexible and non-invasive sensor might provide useful additional information about breathing during sleep.

Keywords: Emfit, movement sensor, sleep apnea, sleep-disordered breathing, spiking, transcutaneous carbon dioxide

1. Introduction

The standard method to determine sleep-disordered breathing is to count apneas and hypopneas per hour of sleep measured by a nasal pressure transducer or thermistor (Silber *et al* 2007). Obstruction in upper airways during sleep can also be studied with special techniques, such as movement sensors SCSB (static-charge-sensitive bed) or Emfit (electromechanical film) (figures 1(a)–(c)). In high-frequency SCSB and Emfit channels rapid spikes with waxing and waning amplitudes correlate well with obstructive events (Polo *et al* 1988).

Besides obstructive events, prolonged flow limitation in nasal cannula preceding an arousal is considered as a significant respiratory event (respiratory effort-related arousal, RERA (Silber *et al* 2007). Prolonged flow limitation periods up to several minutes are a common finding in sleep studies (Hernandez *et al* 2001). Bao and Guilleminault (2004) present in their article a figure with prolonged flow limitation and simultaneous sustained increase in esophageal pressure. Also in movement sensors a prolonged respiratory pattern can be recognized (Alametsa *et al* 2006, Polo *et al* 1989). It appears as prolonged high-frequency spiking with high-amplitude spikes sticking out from the background activity lasting even tens of minutes (Alihanka 1987). Scoring rules for waxing and waning respiratory events and for prolonged spiking phenomena are specified in a scoring manual (Alihanka 1987). During the last 20 years the movement sensor based scoring systems have been widely used in clinical practice in Finland.

The experience with SCSB and Emfit has shown that prolonged spiking is common among patients referred to a sleep laboratory (Polo *et al* 1988). Spiking can be seen in patients with sleep apnea but can also appear alone. Prolonged spiking is a frequent finding in clinical sleep studies among postmenopausal women (Polo-Kantola *et al* 2003). Experimentally, in awake subjects, prolonged spiking is found to be related to the elevation of intrathoracic pressure (Kirjavainen *et al* 1996) and to the increase in end-tidal carbon dioxide tension (Kirjavainen 1997).

Transcutaneus carbon dioxide tension ($TcCO_2$) measurements can also be used to evaluate arterial partial pressure of carbon dioxide. Although a $TcCO_2$ measurement is widely used in operating room and intensive care units its position in sleep studies is still unestablished (Chhajed *et al* 2004). Already in 1984 Midgren with co-workers found that in some apnea patients $TcCO_2$ increased with sleep (Midgren *et al* 1984). Somewhat later Gislason with co-workers noticed a slight increase in $TcCO_2$ during obstructive sleep apneic events and a cumulative increase in $TcCO_2$ with repetetive long apneas (Gislason *et al* 1989). As prolonged spiking has been linked to the increase in the carbon dioxide tension in wakefulness (Kirjavainen *et al* 1996), we hypothesize that carbon dioxide tension changes might also happen in sleep during prolonged spiking episodes.

The aim of the present study was to study the transition trend of $TcCO_2$ in prolonged spiking episodes, using episodes of repetitive hypopnea, apnea or normal breathing as a reference. To deepen the insight into the breathing disturbances, also the changes of arterial oxyhaemoglobin saturation (SaO_2) and pulse rate were calculated. As the control of breathing is different in rapid eye movement (REM) sleep and in non-REM (NREM) sleep (Krieger 2000), the present study was focused solely on NREM sleep.

2. Methods

2.1. Subjects

115 consecutive polysomnograms (from different patients) equipped with $TcCO_2$ measurement were analysed retrospectively. The study was approved by the Ethical Committee of the Pirkanmaa Hospital District.

(a)

(b)
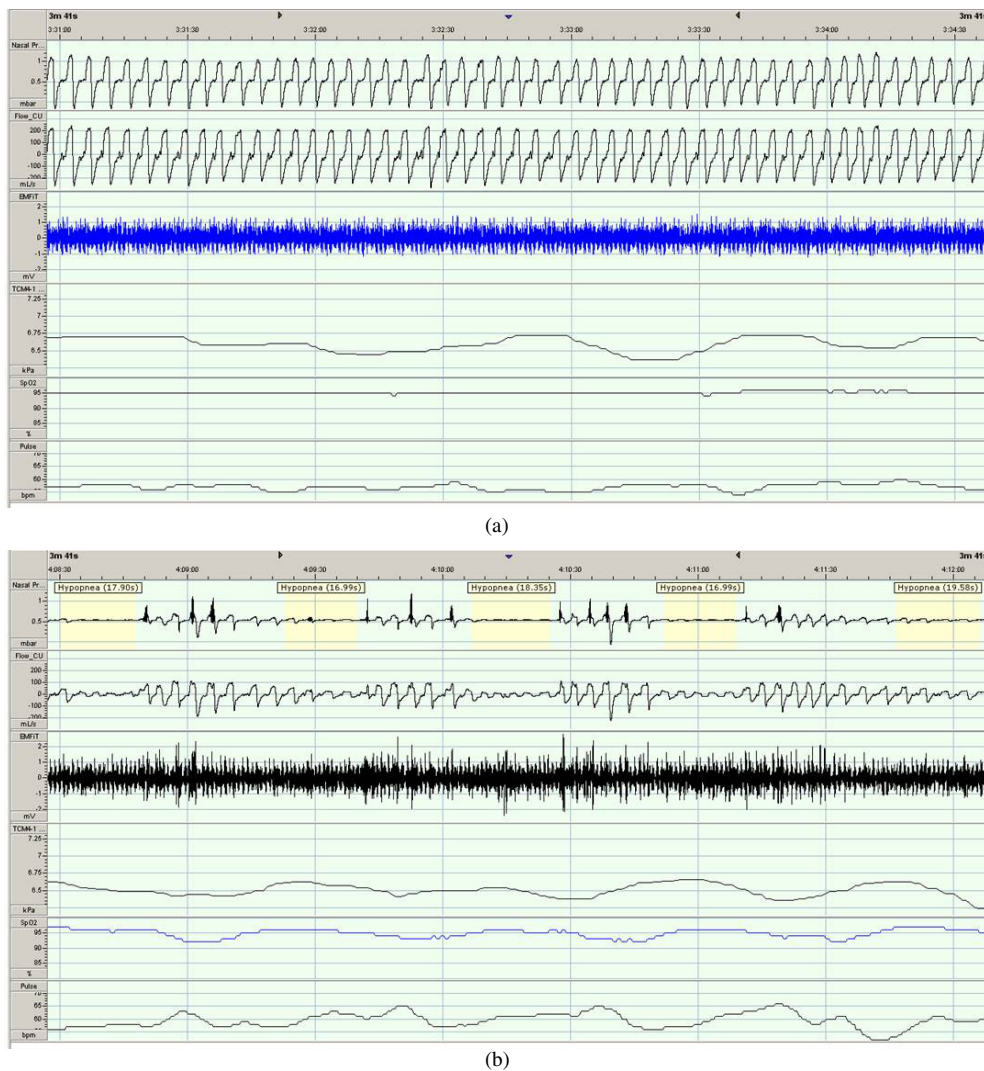
Figure 1. Signals from nasal prongs, Emfit sensor, TcCO$_2$, SaO$_2$ and pulse rate in various breathing patterns. Three minute and 41 s long tracings are pictured. (a) Normal, steady breathing with round flow shape in nasal prongs and without spikes in Emfit signal. (b) Periodic breathing with hypopneas in nasal prongs and varying amplitude of spikes in Emfit. (c) Gradually growing and sustained series of spikes (prolonged spiking) in Emfit signal.

(This figure is in colour only in the electronic version)

2.2. Recording equipment

Embla N7000 and Somnologica software (Medcare, Iceland) were used as the recording system. The recording montage consisted of six EEG derivations, two EOG channels and submental EMG, electrocardiogram, nasal pressure transducer, thoracic and abdominal respiratory movements, body position, anterior tibialis muscle EMG, arterial oxyhaemoglobin saturation and pulse rate by a finger pulse oximeter (Nonin XPOD®, Nonin Medical Inc.,

(c)
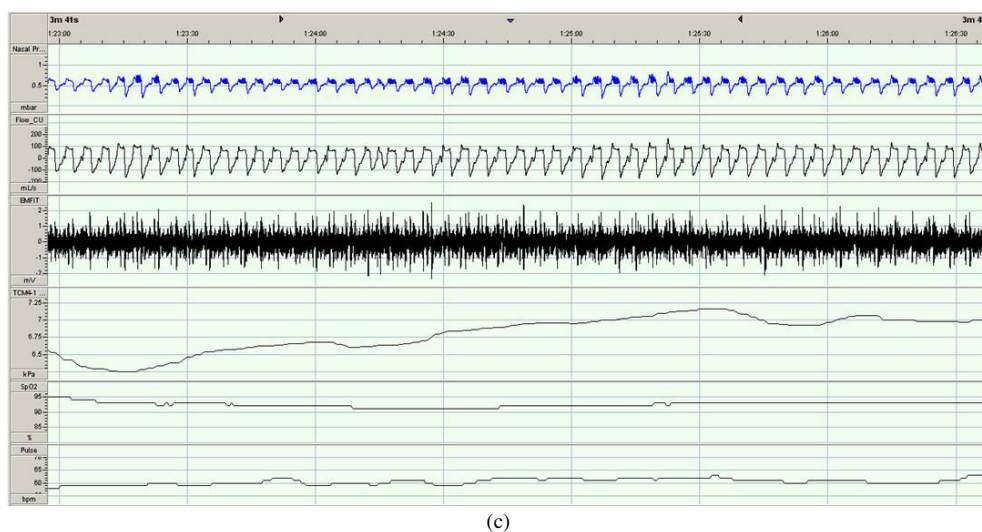
Figure 1. (Continued.)

USA), transcutaneous measurement of carbon dioxide (TcCO$_2$, Tina TCM4, Radiometer, Denmark) and the Emfit sensor (Emfit Ltd, Finland). The Emfit sensor is a small mattress, which consists of a thin, flexible, lightweight, biaxially oriented plastic film coated with electrically conductive, permanently polarized layers. Changes in the pressure acting on the film generate a charge on its electrically conductive surfaces and this charge can be measured as a current or voltage signal. When used as a sensor, it converts mechanical energy into an electrical signal (Paajanen *et al* 2000). The Emfit sensor with dimensions of 32×62 cm was placed under the thoracic area of the sleeping patient. The Emfit signal was amplified and filtered into the 6–16 Hz band (Alametsa *et al* 2006, Kirjavainen *et al* 1996). The sampling rate of 1 Hz was used for oximetry (SaO$_2$ and pulse) and TcCO$_2$ measurements, and 200 Hz for all other parameters. TcCO$_2$ recordings were performed for 8 h with the probe temperature of 43 °C (Janssens *et al* 2001). The TcCO$_2$ electrode was placed on the sternum three fingers breadth below the jugulum.

2.3. Analysis of signal

Recordings were scored into sleep stages according to the old standard criteria (Rechtschaffen and Kales 1968). Microarousals were scored according to the criteria of the American Sleep Disorders Association (1992). The apnea–hypopnea index (AHI) was calculated as the number of cessations or diminutions $\geqslant 50\%$ of airflow lasting over 10 s per hour of sleep. NREM sleep periods of the recordings were selected for further analysis with the help of the hypnogram. In the Emfit analysis all the severity categories of spiking (light, moderate and heavy, see Kirjavainen *et al* (1996)) were combined and included in the study. Awake TcCO$_2$ values just before falling asleep were collected.

2.4. Study design

In the first phase the Emfit signal of the recordings was visually evaluated and all prolonged spiking episodes lasting at least 3 min were selected. The scoring was performed by two independent scorers and the agreement of the overlapping prolonged spiking episodes was

72% (50–83%). All the events, which were overlapping, were accepted. For non-overlapping episodes, consensus selection was performed by the two independent scorers together. No other traces were visible during this procedure. The recordings containing at least one accepted prolonged spiking episode passed to the second phase. In the second phase all at least 3 min episodes of repetitive hypopnea and repetitive obstructive apnea (maximum four breaths were accepted between the respiratory events) were selected from the nasal pressure and the thoracic and abdominal movement channels. The same consensus scoring procedure was used in this phase. If at least one 3 min hypopnea episode or apnea episode was found, the recording was eventually included in the study. 19 patients fulfilled the criteria. There were several patients with shorter than 3 min episodes of spiking, hypopnea and apnea, but we chose to accept only the episodes $\geqslant 3$ min because changes in $TcCO_2$ are found to be rather slow. In addition to hypopnea, apnea and prolonged spiking episodes, also $\geqslant 3$ min episodes of normal breathing were collected.

The inspiratory flow shapes were visually classified on a breath-by-breath basis according to principles earlier described (Aittokallio *et al* 1999). For the purpose of this study the rounded inspiratory flow shapes (without flow limitation) were distinguished from the non-rounded (or flow-limited) ones.

Breathing was defined as normal if more than 80% of inspiratory flow shapes were round. The share of round flow shapes was also calculated for prolonged spiking, hypopnea and apnea episodes. The agreement for flow shape scoring was 96% (89–98%).

To study the long-term transition of $TcCO_2$, SaO_2 and pulse rate signals in collected breathing pattern sections, the episodes were divided into 50 parts. Median values of every part were used to define a trend line and to calculate the change per minute values for $TcCO_2$, SaO_2 and pulse rate. In addition, the initial values for $TcCO_2$, SaO_2 and pulse rate signals in the beginning of the breathing episodes were collected.

2.5. Statistical analysis

Statistical analyses were performed with SPSS for Windows version 12.0® (SPSS Inc.). Non-parametric tests were used, as all the variables were not normally distributed. Multiple comparisons of independent variables were performed by the Kruskal Wallis test with the post hoc analysis by the Mann–Whitney U test. In all the post hoc analyses the Bonferroni correction factor was used. P-values < 0.05 were considered statistically significant. Spearman rank correlations were calculated to measure the strength of associations between the durations of the breathing episodes and the amount of $TcCO_2$ changes. Also the correlations between the changes of $TcCO_2$ and the durations of individual hypopneas and apneas within hypopnea and apnea episodes were defined.

3. Results

3.1. Polysomnographic and respiratory results

A total of 115 polysomnograms were evaluated. Ten recordings had technical shortcomings and were therefore discarded. Twenty-two recordings were pediatric and were not included in the study. Nineteen subjects (14 men and 5 women) passed all the study phases and were accepted into the further evaluation. The indications of the remaining 64 recordings were variable; 32 patients had sleep-disordered breathing not fulfilling the inclusion criteria, 7 patients presented with periodic leg movements, 9 patients had insomnia, 2 patients had narcolepsy, 1 patient had nocturnal groaning, 4 recordings were performed with

Table 1. The demographic data and sleep parameters of the subjects included in the study and 32 subjects with sleep-disordered breathing not fulfilling the inclusion criteria.

	Included patients			Not included patients			
	Median	Min	Max	Median	Min	Max	*p*-value
Age (years)	46	37	59	50	25	65	ns[n]
BMI[a] (kg m^{-2})	30	21.8	46.1	28	14.5	44	ns
TIB[b] (min)	488	364	602.5	487.5	374.5	577	ns
TST[c] (min)	379	233	529	429.5	234	538	ns
SEI[d] (%)	83	53	97	86	54	97	ns
SL[e] (min)	10.5	2.5	40.5	9.5	1.5	156.5	ns
REMlat[f] (min)	118	79.5	441	84	12.5	387	<0.01
%S1[g]	13.8	0.1	39.6	8	2.2	31.3	<0.05
%S2[g]	62.4	48.5	86.8	65.1	45.5	84.9	ns
%S3[g]	5.6	0	24.3	6.4	0	10.6	ns
%S4[g]	0	0	2.8	0	0	3.9	ns
%SREM[g]	14.1	3.7	22.2	20	4.5	36.5	<0.01
ARI[h]	32	11.8	68.7	15.7	4.2	38.1	<0.001
AHI[i]	22	9	62	17	0	59	ns
AI[j]	6	1	57	6	0	51	ns
ODI4[k]	15	1	58	5	0	57	ns
SaO$_2$ min[l] (%)	84	66	91	88	68	93	<0.05
SaO$_2$ mean[m] (%)	94.5	90.5	96.3	94.5	91.8	97.9	ns

[a] Body mass index.
[b] Time in bed.
[c] Total sleep time.
[d] Sleep efficiency index.
[e] Sleep latency.
[f] Latency to REM (rapid eye movement) sleep.
[g] Percentage of sleep stage (S1–SREM) referred to TST.
[h] Arousal index.
[i] Apnea–hypopnea index, apneas and hypopneas per hour of TST.
[j] Apnea index, apneas per hour of TST.
[k] Oxygen desaturation index, number of desaturations $\geqslant 4\%$ per hour of TST.
[l] Minimum of oxygen saturation.
[m] Mean of oxygen saturation.
[n] Not significant.

nasal continuous positive airway pressure therapy and 8 subjects were healthy volunteers participating in another study.

The demographic data and sleep parameters of the 19 subjects included in the study and the 32 subjects with sleep-disordered breathing not included were compared. The study group had less REM sleep, more stage 1 sleep and had longer REM latency. They had more arousals and their minimum oxygen saturation was lower. The demographic data and sleep parameters from the recordings are shown in table 1. In the study group seven patients had a low apnea–hypopnea frequency ($5 < AHI < 15$), five had an intermediate ($15 \leqslant AHI < 30$) and seven patients had a high apnea–hypopnea frequency ($AHI \geqslant 30$).

Altogether 115 different breathing pattern episodes from the accepted 19 recordings were adopted into the study. Forty-five episodes represented periodic hypopnea, 24 episodes periodic apnea, 33 episodes represented prolonged spiking and 13 episodes normal breathing. Apnea episodes were missing from six recordings, normal breathing episodes were missing from five recordings but hypopnea episodes were found in all recordings. As expected, no

Table 2. The median changes per minute, 25th and 75th percentiles, range and the change direction of TcCO$_2$, SaO$_2$ and pulse rate in different breathing episodes.

		Median change per min	25th percentile	75th percentile	Range	Change positive
Normal breathing episode	TcCO$_2$[a] (Pa)	−3.218	−9.820	15.817	−39.729–25.788	5/13
	SaO$_2$[b] (%)	−0.028	−0.100	0.041	−0.467–0.362	5/13
	Pulse (min^{-1})	−0.0005	−0.004	0.001	−0.033–0.007	5/13
Prolonged spiking episode	TcCO$_2$ (Pa)	13.074	6.224	37.909	−12.321–156.758	28/33
	SaO$_2$ (%)	−0.046	−0.348	0.001	−0.780–0.073	9/33
	Pulse (min^{-1})	0.113	−0.004	0.440	−0.519–1.455	24/33
Hypopnea episode	TcCO$_2$ (Pa)	4.060	−9.618	16.872	−98.820–78.652	27/45
	SaO$_2$ (%)	0.018	−0.068	0.127	−0.457–1.574	27/45
	Pulse (min^{-1})	−0.166	−0.387	0.079	−1.105–1.037	15/45
Apnea episode	TcCO$_2$ (Pa)	0.898	−9.609	10.640	−68.026–41.291	12/24
	SaO$_2$ (%)	0.080	−0.098	0.218	−0.426–0.685	16/24
	Pulse (min^{-1})	−0.142	−0.350	0.153	−0.990–0.895	10/24

[a] Transcutaneous carbon dioxide tension (unit: pascal).
[b] Oxygen saturation.

hypopnea or apnea was present during prolonged spiking episodes. The selected breathing pattern episodes occurred mainly in sleep stage 2 (S2) but 8 out of 25 prolonged spiking episodes appeared in sleep stage 3 (S3) whereas no hypopnea or apnea episodes were present in S3.

The median arousal frequency was similar in prolonged spiking episodes (0 per min) and in normal breathing episodes (0 per min, ranges 0–0.3 and 0–0.7 per min, respectively). The frequencies of arousals in hypopnea episodes and apnea episodes did not differ (1.1 per min versus 1.1 per min, ranges 0.6–1.8 and 0.4–1.6 per min), but were higher than in prolonged spiking episodes ($p < 0.001$) or during normal breathing ($p < 0.001$).

In the analysis of the flow shape the median percentage of round flow shape during normal breathing was 91% (range 80–100), during prolonged spiking 3% (0–38), during hypopnea episodes 18% (0–28) and during apnea episodes 12% (6–38). Statistically the share of round flow shapes was significantly higher during normal breathing than in the other breathing episodes (all p-values < 0.001). The frequency of round flow shape was lowest during prolonged spiking episodes (all p-values < 0.001) but the hypopnea episodes did not differ from the apnea episodes in this respect.

3.2. TcCO$_2$ results

The median awake TcCO$_2$ was 5.6 kPa (range 4.7–6.0). The median values of TcCO$_2$ changes per minute in the various breathing pattern episodes are presented in table 2. During normal breathing, 8 out of 13 episodes had a decreasing TcCO$_2$ trend, resulting in an overall negative median TcCO$_2$ change per minute. The TcCO$_2$ increased during 28 out of 33 prolonged spiking episodes, with an overall median increase of 13 Pa min^{-1}. The median TcCO$_2$ change per minute for hypopnea episodes was slightly increasing, with 18 decreasing and 27 increasing values. For the apnea episodes the median TcCO$_2$ change was also slightly increasing; 12 changes were decreasing and another 12 were increasing. Examples of respiratory signals and TcCO$_2$ curves during normal breathing, hypopnea and prolonged spiking episodes are seen in figures 1(a)–(c).

Table 3. Statistical comparison of the changes per minute in different breathing patterns. Bonferroni corrected *p*-values are presented.

	Normal versus spiking	Normal versus hypopnea	Normal versus apnea	Spiking versus hypopnea	Spiking versus apnea	Hypopnea versus apnea
$TcCO_2$[a]	0.0093	ns	ns	0.0005	0.0006	ns[c]
SaO_2[b]	ns	ns	ns	0.0024	0.0054	ns
Pulse rate	ns	ns	ns	0.0004	0.0167	ns

[a] Transcutaneous carbon dioxide tension.
[b] Oxygen saturation.
[c] Not significant.

Statistically the $TcCO_2$ change in prolonged spiking episodes differed significantly from the change in other breathing episodes (table 3). The $TcCO_2$ changes between normal breathing, hypopnea episodes and apnea episodes did not show statistically significant differences.

The initial levels of $TcCO_2$ in various breathing episodes did not show statistically significant differences. The length of the breathing pattern episodes did not contribute to the level of the $TcCO_2$ change as no significant correlations between the durations of various breathing episodes and the changes per minute of $TcCO_2$ were observed. Neither were there significant correlations between the durations of individual hypopneas or apneas inside the breathing pattern episodes and the changes of $TcCO_2$ (all correlation coefficient values between -0.25 and 0.22, all *p*-values > 0.05).

3.3. SaO₂ and pulse rate results

In normal breathing and during prolonged spiking episodes, SaO_2 tended to decline and in hypopnea and apnea episodes, it slightly rose. Pulse rate during prolonged spiking episodes showed a rising trend, in hypopnea and apnea episodes the pulse rates slightly declined and practically no change occurred in normal breathing (table 2).

Statistically, the changes in SaO_2 and pulse rate during the prolonged spiking episodes differed significantly from the changes of the hypopnea and apnea episodes but the other comparisons showed no significant differences (table 3).

4. Discussion

In this study we found that a prolonged spiking phenomenon in the Emfit sensor during sleep in normocapnic patients seems to be related to a gradual elevation of $TcCO_2$ tension. This trend differed clearly from the changes observed in normal breathing and periodic breathing patterns (episodes of apnea and hypopnea). Additionally, there were divergent changes in SaO_2 and pulse rate. During prolonged spiking episodes SaO_2 values showed a declining trend with a slight increase in the pulse rate while during hypopnea and apnea episodes the trend of SaO_2 was rising and pulse rates declined slightly. However, the changes in SaO_2 and pulse rate between normal breathing and prolonged spiking episodes were not statistically significant.

A number of earlier studies have described and investigated the mechanisms of the spiking phenomenon. In SCSB studies, prolonged spiking has been suggested to arise from the increase in respiratory resistance during sleep (Alihanka 1987, Polo *et al* 1989) and breathing effort (Kirjavainen *et al* 1996). Gradually growing spike amplitude correlates well with the

increase in negative esophageal pressure (Kirjavainen *et al* 1996). In addition, increased breathing effort with an increase in the negative esophageal pressure has been associated with flow limitation pattern in nasal cannula (Johnson *et al* 2005). We and others have previously described a rise in end-tidal carbon dioxide pressure during prolonged periods of flow limitation (Calero *et al* 2006, Kirjavainen 1997). The non-round inspiratory flow shapes during sleep-disordered breathing in the present study confirmed that the investigated episodes were related to upper airway obstruction. The non-round inspiratory flow shapes were the most constant phenomenon during the prolonged spiking episodes whereas during the episodes of hypopnea or apnea also round shapes occurred repeatedly, reflecting arousals (Aittokallio *et al* 2001).

The partial pressures or arterial oxygen and carbon dioxide are the basis of the metabolic control of breathing during sleep (Douglas 2000). Sleep recordings routinely include SaO$_2$ measurements to show the desaturations during episodes of repetitive apnea and hypopnea. End-tidal carbon dioxide measurements are widely used in operating rooms, intensive care units and children's sleep studies. However, in adults the carbon dioxide measurements are still infrequently used in sleep studies even if the rise in carbon dioxide pressure is the strongest stimulant of breathing during sleep. Thus TcCO$_2$ measurements might provide important information about characteristics of breathing disturbances during sleep. Changes in TcCO$_2$ occur quite slowly but in the present study it caused no problems because our focus was on long-lasting alterations. The transcutaneus method of carbon dioxide measurement was also used because the TcCO$_2$ recording gives a continuous signal also during apnea unlike the end-tidal measurement of CO$_2$.

Two early descriptive studies suggested that in obstructive sleep apnea syndrome (OSAS) patients, the changes in TcCO$_2$ values were influenced by the sleep stage, the severity of the disease and the patient's body mass (Gislason *et al* 1989, Midgren *et al* 1984). In our study the AHI and BMI (body mass index) of the subjects varied substantially. Since the number of patients remained low we decided not to divide subjects into subgroups, e.g. by severity of the disease or BMI. Only 19 recordings could be included in the study because severe apnea patients did not show long enough prolonged spiking episodes and, on the other hand, patients with prolonged spiking episodes had often only minor amounts of other forms of sleep-disordered breathing.

Chronic hypercapnia (while awake) is frequently found in severe OSAS patients. Ventilation during sleep between obstructive events is tightly linked to CO$_2$ kinetics during periodic breathing, appearing to be an important mechanism that defends against the development of acute hypercapnia in very obese patients with severe OSAS (Berger *et al* 2000, 2002). In the present study, the change in TcCO$_2$ was studied in representative episodes, not in the entire time of sleep. Our patients were eucapnic, only moderately obese and their OSAS was also moderate. Thus our patient group differed essentially from Berger's. Presumably for these reasons we did not discover a significant rise in TcCO$_2$ in hypopnea and apnea episodes. We did not either see correlations between the durations of individual hypopneas or apneas inside the breathing pattern episodes and the changes of TcCO$_2$. Mechanisms for CO$_2$ accumulation in very severe OSAS and in sleep-disordered breathing with hypopneas and long-standing flow limitation as well seem to be different.

Altogether these findings suggest that the prolonged spiking phenomenon measured by the Emfit sensor might represent another type of breathing disturbance during sleep, which differs from classical hypopnea and apnea periods. Thus Emfit would provide a valuable addition to sleep studies as a simple method that does not interfere with sleep, because no electrodes are attached to the patient. Emfit is made of flexible material, any size or shape of sensor can be used, it is easy to carry and therefore it is also suitable for ambulatory

sleep recordings. In Finland, Emfit is often used as a supplementary device because with a movement sensor a reliable signal is always gained, contrary to nasal prongs, which can become detached. Nasal prongs can also markedly increase the nasal airflow resistance in subjects presenting with nare narrowness and/or deviated nasal septum (Lorino *et al* 2000). However, in a study by Thurnheer *et al* (2001) subjective nasal obstruction did not impair the accuracy of nasal pressure monitoring though in another study impaired nasal ventilation prevented adequate measurements of nasal pressure in 9% of the subjects (Series and Marc 1999). As an additional device Emfit can reduce the risk of unreliable respiratory recordings.

For the purpose of the present study, patients presenting with various forms of breathing patterns were selected. Therefore, clinically they do not represent sleep apnea patients with the most severe condition. In clinical patient populations there are many symptomatic patients in whom the prolonged spiking is the only respiratory abnormality during sleep. This is an abnormality that is particularly common in postmenopausal females (Polo-Kantola *et al* 2003) and which responds to nasal CPAP therapy with good adherence (Anttalainen *et al* 2007). The spiking is more common in OSAS patients with narrow upper airway at tongue base or hyoid bone levels (Polo *et al* 1991) or after shortening of the soft palate (Polo *et al* 1989). Our clinical experience is that patients with spiking have a higher tendency for hypercapnic respiratory failure. Therefore, it seems that the spiking episodes have both anatomic and functional determinants and may not just represent a stage of developing obstructive sleep apnea.

Our results provide further evidence that the phenomenon of prolonged spiking recorded during sleep with sensitive movement sensors such as Emfit arises from the increased drive of breathing caused by flow limitation and increasing transcutaneous (or end-tidal) CO_2. The Emfit sensor provides a convenient method for distinguishing simple flow limitation from the clinically more relevant form of progressive flow limitation with gradually increasing CO_2 and gradually increasing respiratory drive, which in a subset of patients with sleep-disordered breathing may explain the symptoms of sleepiness or morning headaches.

Acknowledgments

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Compressed tracheal sound analysis in screening of sleep-disordered breathing

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Abstract

Objective: To evaluate the suitability of compressed tracheal sound signal for screening sleep-disordered breathing.
Methods: Thirty-three consecutive patients underwent a polysomnography with a tracheal sound analysis. Nineteen patients were healthy except for the sleep complaint, 9 were hypertonic and 3 were hypertonic and had elevated cholesterol. Minimum and maximum values of each consecutive, non-overlapping segment of 15 s of original sound data were extracted. All these compressed tracheal sound traces were divided into plain, thin and thick signal periods. Also pure, 10-min episodes of plain, thin and thick tracheal sound periods were selected and the nasal pressure flow shapes during these pure sound periods were examined.
Results: There was a significant positive correlation between the total nocturnal amount of thick periods and AHI. Apneas and hypopneas were most common during the 10-min episodes of thick sound periods. The proportion of round (normal, non-flattened) inspiratory flow shape was highest during the pure plain periods.
Conclusions: Breathing consisting of apneas and hypopneas can quite reliably be visualised with compressed tracheal sound analysis. The other interesting outcome of the study is that even prolonged flow limitation might be revealed with the method.
Significance: Compressed tracheal sound analysis might provide a promising screening method for obstructive apneas and hypopneas.
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Keywords: Breathing sound; Sleep; Sleep apnea; OSAS; Flow limitation

1. Introduction

Sleep-disordered breathing is common, and its prevalence is estimated to be 2–4% (Young et al., 1993). Hence the need for sleep studies is vast. Sleep-disordered breathing is usually diagnosed by calculating the number of hypopneas and apneas per hour of sleep (apnea–hypopnea index, AHI) from the flow-channel. Apneas can be detected by the thermistor, but a nasal pressure transducer is recommended for the detection of hypopneas (Iber et al., 2007). A nasal pressure transducer signal (cannula signal) also provides information of snoring and of the shape of the inspiratory flow curve. Flattened, non-round inspiration is present during hypopneas but it may also appear in prolonged periods up to several minutes. This prolonged flow limitation is found to be rather common (Hernandez et al., 2001).

Apneas and hypopneas are most often selected or confirmed visually. Various methods with semi-automated analyses have been developed to speed up this procedure and to make it more objective, independent of the analyst. One of the new methods to study respiratory events during sleep is the analysis of tracheal sound. Respiratory events

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can be selected from a tracheal sound channel both visually and automatically by means of intensity or power changes of the sound signal (Van Brunt et al., 1997; Cummiskey et al., 1982; Nakano et al., 2004).

When the amount of collected data is large, compressing the raw data is often used to enhance the review procedure. This is used, e.g. in a long-term EEG-monitoring where a compressed EEG-signal is used to facilitate visual detection of waveform alternations (Agarwal et al., 1998; Hellstrom-Westas et al., 2006). Even clinical and treatment decisions are made based on these compressed signal changes. This principle was utilized when we first decided to reduce tracheal sound data by compressing the sound signal.

In our pilot study, we measured breathing sounds from the neck by a microphone using 11,025 Hz sampling rate. To reduce the sound signal data only the maximum and minimum sound signal values of each consecutive 20 s epochs were taken. The resultant two compressed sound signal traces that together form a compressed sound signal curve (Fig. 1) were analysed visually. In the compressed tracheal sound curve three different patterns were dominating: periods with a plain signal curve close to zero, periods with a thin signal curve deviating clearly from zero and periods with a thick signal curve (Himanen et al., 2007). With a small group of apnea patients ($n = 10$) we found a good correlation between the amount of thick-pattern and the conventional AHI of the polysomnography during sleep. In the other pilot study, we examined the compressed sound signal curve of five 7-year-old children (Saarenpää-Heikkilä et al., 2007). Their compressed sound curves were mostly plain. Thick signal episodes were sparse and all AHIs were below 2/h. Two of the children had one longer flow limitation period each in the nasal cannula tracing. During these periods of flow limitation they both had a

representative thin sound signal curve in the tracheal sound channel. No other thin signal curves were present.

Based on the presented pilot studies we hypothesize that the use of compressed sound analysis may aid the identification/visualization of various sleep related respiratory episodes. To test the hypothesis we compared the different compressed sound trace patterns with the parameters derived from nasal pressure tracing.

2. Methods

Thirty-three consecutive patients (27 male, 6 female) referred to the Sleep Unit of Pirkanmaa Hospital District volunteered to participate in this study. All the subjects gave their written consent. One female and one male were referred due to the suspicion of narcolepsy, one male subject had restless legs, and the rest were referred due to the suspicion of sleep-disordered breathing. Nineteen patients were otherwise healthy, 9 were hypertonic and 3 were hypertonic and had elevated cholesterol. The study was approved by the Ethical Committee of the Pirkanmaa Hospital District.

Embla N7000 and Somnologica software (Medcare, Iceland) were used as a recording system. The recording montage consisted of six EEG derivations (Fp1–A2, Fp2–A1, C3–A2, C4–A1, O1–A2, O2–A1), two EOG channels and submental EMG, electrocardiogram, nasal pressure transducer, thermistor, thoracic and abdominal respiratory movements, body position, anterior tibialis muscle EMG, blood oxygen saturation and pulse by pulseoximetry and tracheal sound recording.

Night recordings were scored into sleep stages according to the old standard criteria (Rechtschaffen and Kales, 1968). The apnea–hypopnea index (AHI) was calculated
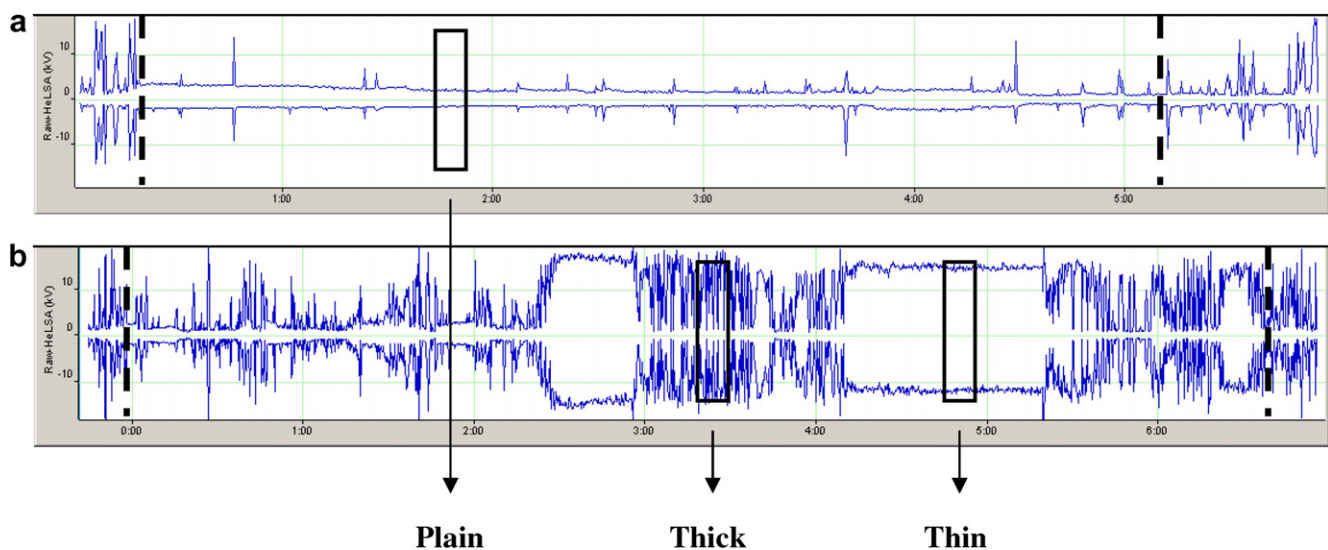

Fig. 1. (a) Compressed tracheal sound signal curve of a subject with AHI of 2/h and the proportion of thick periods of 1%. The prevailing pattern was plain with some big deflections which were caused by movements. (b) Example of recording with plenty of thick and thin periods. AHI was 26/h and the proportion of thick periods was 56%. Representative periods of pure thick, thin and plain patterns are marked on the recordings. The lights-off–lights-on time is between the dashed lines. Note the artifacts during wakefulness.

from the nasal pressure channel as the number of cessations or diminutions $\geqslant 50\%$ of airflow lasting over 10 s per hour of sleep. For hypopneas new scoring rules were used (Iber et al., 2007). Microarousals were scored according to the criteria of the American Sleep Disorders Association (1992).

The tracheal sound recordings were performed with a small electret microphone, Panasonic WM-60A (Matsushita Electric Industrial Co., Ltd, Kadoma Osaka, Japan). The microphone has conical air cavity of 25 mm in diameter and 3 mm deep. The sensitivity of the microphone is 10 mV/Pa and the frequency range in the free field is 20 Hz–20 kHz, ± 2 dB (Sovijarvi et al., 1998). The microphone is attached to the suprasternal fossa with an adhesive tape ring and an additional taping on the top (Fig. 2). The measured breathing sound signal is amplified and high-pass filtered with a preamplifier unit. After the preamplifier unit, the signal is fed into an external sound card USB Sound Blaster Audigy 2 NX (Creative Labs, Singapore) for A/D conversion followed by USI-01 USB isolator (MESO, Mittweida, Germany) providing galvanic isolation between the measuring device HeLSA and the PC. The sampling frequency for the sound signal is 11,025 Hz. SuperHeLSA software (Pulmer, Helsinki, Finland) provides the raw data from the sounds recorded over the trachea, which is converted into Embla data format.

Based on the visual and mathematical evaluation of the compressed signal curves we decided to increase the sampling rate from three to four samples per minute. Internal dynamics of each consecutive, non-overlapping segment of 15 s of original sound data was estimated by extracting the minimum and maximum value inside the segment. This compressed sound information (four samples per minute) was used for the actual analysis. As in the pilot studies, simultaneous visualization of the two compressed sound traces through time revealed three dominating patterns: a plain signal curve close to zero, periods with a thin signal curve deviating clearly from zero and periods with a thick signal curve (see Fig. 1). Additionally, the fourth pattern, where the line was at zero, was present in two recordings. Further examination of these recordings revealed that the microphones had become detached during the night. Therefore, we decided to exclude these recordings from the data set.

All the compressed tracheal sound traces during the sleep period were manually analysed by two different scorers and divided into plain, thin and thick periods. Altogether 780 different pattern periods were scored in the recordings. The agreement between the classifiers was 86.3%. The two individual scorers classified together consensus scorings based on the individual scorings. The consensus scorings were used in the analyses.

After that one representative episode of plain, thin and thick tracheal sound periods (later pure plain, pure thin and pure thick) lasting 10 min were selected from each subject, whenever present. AHIs during these pure periods were calculated.

Breathing and tracheal sound was explored in more detail by comparing the flow shapes of the nasal pressure signal in different pure sound periods. Every inspiration during pure plain, pure thin and pure thick patterns was divided into a round or non-round flow shape category (Calero et al., 2006; Rauhala et al., 2007) by two different classifiers. The scoring agreement between the classifiers was 95% (range 87–98%). The proportion of round (normal, non-flattened) breathing was calculated for each pure sound period.

To examine the hypothesized connection between AHI and the total nocturnal amount of different sound patterns, we computed percentages of plain, thin and thick periods during the night for each subject.

2.1. Statistics

Statistical analyses were performed with SPSS for Windows version 12.0® (SPSS Inc.). Non-parametric tests were used, as all the variables were not normally distributed.

The Friedman test was used in comparison between the dependent variables with the Wilcoxon signed-rank test in post hoc analyses. In these post hoc analyses the Bonferroni correction factor was used to correct for multiple comparisons. p-Values < 0.05 were considered statistically significant.

The relationship between AHI and the percentage of thick, thin and plain periods was studied in two ways. First, non-parametric correlation coefficients were computed. Second, a linear regression model between the ranks of both AHI and the thick period percentages was formed. To verify the model, each patient's AHI based on the patient's thick period percentage was predicted.

Fig. 2. The recording site of the microphone in the suprasternal fossa. The fixation taping is not in place to show the exact site.

3. Results

The sleep parameters derived from the sleep recordings of the 31 valid subjects are shown in Table 1. Three subjects had no significant findings in their polysomnograms. One subject had periodic leg movements and one subject had narcolepsy. The rest of the patients had sleep-disordered breathing.

Twenty-nine subjects out of 31 possessed a pure thick sound pattern. Twenty-seven subjects had pure thin patterns. A pure plain pattern was found in 21 subjects. The median AHI in pure thick sound 10-min periods was 66/h (range 0–132). In pure thin periods and pure plain periods the median AHIs were 0 (ranges 0–96/h and 0–27/h, respectively). The pure thick AHI differed statistically significantly from both the pure thin AHI and the pure plain AHI (p-values 0.0002 and 0.0012, respectively). The difference between the pure thin and the pure plain AHI was not significant. Five subjects had AHI > 10/h during the pure thin period. These subjects had high AHI in the whole night analysis (34–59/h). The three subjects presenting AHI > 10/h during the pure plain periods had AHI in the whole night analysis between 12 and 24.

The proportion of round (normal, non-flattened) inspiratory flow shapes was highest during pure plain periods (median 42.5%) differing significantly from the amount of round shapes during pure thick and thin periods (p-values 0.024 and 0.019, respectively, Fig. 3). Although the amount of inspiratory round flow shapes was lowest during pure thin periods (median 2.5%), it did not differ statistically from the amount of round inspiratory shapes during the pure thick periods (median 19%).
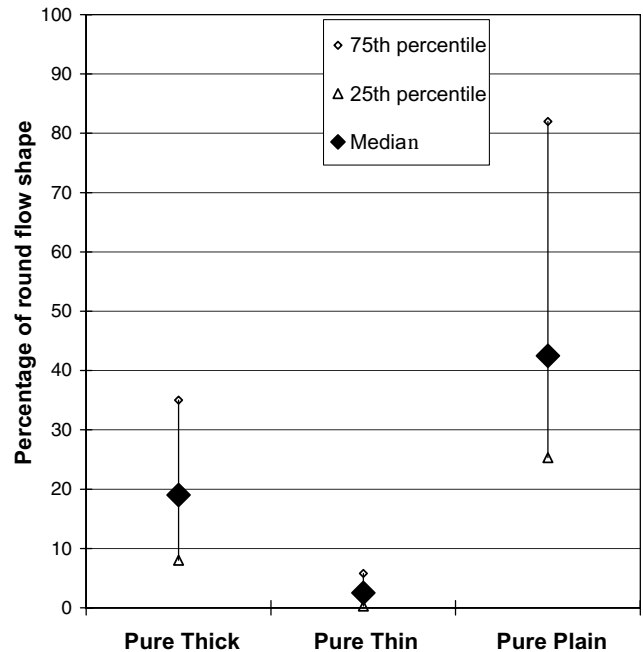

Fig. 3. Percentages of round flow shape during various representative pure breathing pattern periods. Median values and 75th and 25th percentiles are presented.

The study group consisted of consecutive patients. This led to a group with wide age range and a variety of causes of referral. Thus we formed a subgroup of patients aged from 30 to 50 years who all were referred due to a suspicion of sleep-disordered breathing. Seventeen patients were included in this group. Two subjects had no significant findings in their polysomnogram. Fifteen patients had sleep-disordered breathing.

The results were essentially the same as with the 31 subjects. All the 17 subjects possessed a pure thick sound pattern, 16 had pure thin patterns and a pure plain pattern was found in 15 subjects. The median AHI in pure thick sound 10 min periods was 48/h (range 0–132). In pure thin periods and pure plain periods the median AHIs were 0 (ranges 0–6/h and 0–10/h, respectively). The pure thick AHI differed statistically significantly from both the pure thin AHI and the pure plain AHI (p-values 0.0044 and 0.0056, respectively). The difference between the pure thin and the pure plain AHI was not significant. As with 31 subjects the amount of round inspiratory flow shapes was highest during pure plain periods (median 42.2%) differing significantly from the amount of round shapes during pure thick and thin periods (p-values 0.019 and 0.027, respectively).

Fig. 4 contains a plot of the AHI measurements and the percentage of thick periods for all 31 patients. The latter variable shall be referred as thick%. To study the relationship of the (AHI, thick%)-pair Spearman's correlation coefficient (Spearman's ρ) was calculated. We computed Spearman's ρ for two sets of measurements: (1) the entire set of 31 patients and (2) the subset of 17 patients between

Table 1
Demographic data and sleep parameters of the 31 subjects

	Median	Minimum	Maximum
Age (years)	46	16	68
BMI (kg∗m^{-2})	27.8	21.8	44.6
TIB (min)	488	328.5	602.5
TST (min)	396.5	222.5	529
SEI (min)	82.6	65.5	98
WASO (min)	59	6	154.5
SL (min)	9.75	0	53.5
REMlat (min)	124.75	1.5	514
S1 (%)	10.9	1.9	40.6
S2 (%)	67.6	49.5	85.6
S3 (%)	5.2	0	17.7
S4 (%)	0	0	21.3
SREM (%)	13.7	0.4	25.5
ARI (no/h)	44	11	108
AHI (no/h)	24.5	0	92
SaO$_2$ min (%)	86	60	96
ODI4 (no/h)	7.5	0	81

TIB, Time in bed; TST, Total sleep time; SEI, Sleep efficiency index; WASO, Wakefulness after sleep onset; SL, Sleep latency; REMlat, latency to the first REM-sleep period; S1-SREM, percentage of the sleep stage referred to TST; ARI, arousal index; SaO$_2$min, minimum oxygen saturation; ODI4, oxygen desaturation index.
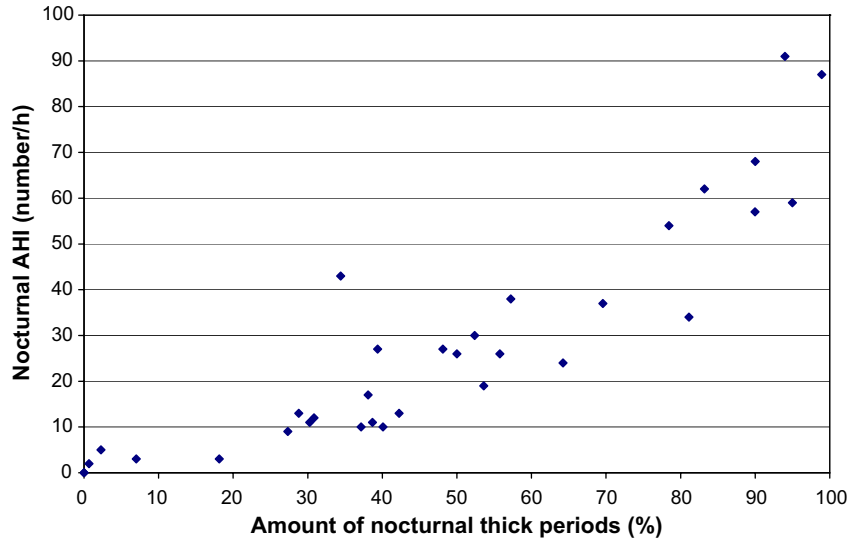
Fig. 4. The relationship between nocturnal AHI and the total nocturnal amount of thick periods. Spearman's correlation coefficient 0.91.

Table 2
Spearman's ρ correlation coefficients and p-values for two data sets

Data set	(AHI, thick%)-pair		(AHI, thin%)-pair		(AHI, plain%)-pair	
	ρ	p-value	ρ	p-value	ρ	p-value
All	0.91	$1.2e^{-12}$	−0.87	$1.2e^{-10}$	−0.06	0.73
30–50 years	0.85	$1.9e^{-5}$	−0.40	0.11	−0.03	0.91

AHI, apnea–hypopnea index; thick%, percentage of thick periods; thin%, percentage of thin periods; plain%, percentage of plain periods; ρ, Spearman's ρ correlation coefficient.

30 and 50 years. Spearman's ρ was computed also for the (AHI, thin%)-pair and the (AHI, plain%)-pair. Table 2 contains the correlation coefficients and the corresponding p-values. The p-values are small for the (AHI, thick%)-pair and the (AHI, thin%)-pair indicating strong positive and negative correlations, respectively.

In addition to these correlation coefficients we also formed a rank-based regression model as follows. The AHI rank, rank(AHI) and the thick period percentage rank, rank(thick%) were computed for each patient. A linear regression model between rank(thick%) and rank(AHI) was formed. Using this rank-based linear model we then predicted the patient's AHI value from his/her thick period percentage. Connover (1999, Section 5.6) provides more details on this model. Fig. 5 contains the results of this model. The proximity of measured and predicted points indicates that this model could be used for prediction purposes.
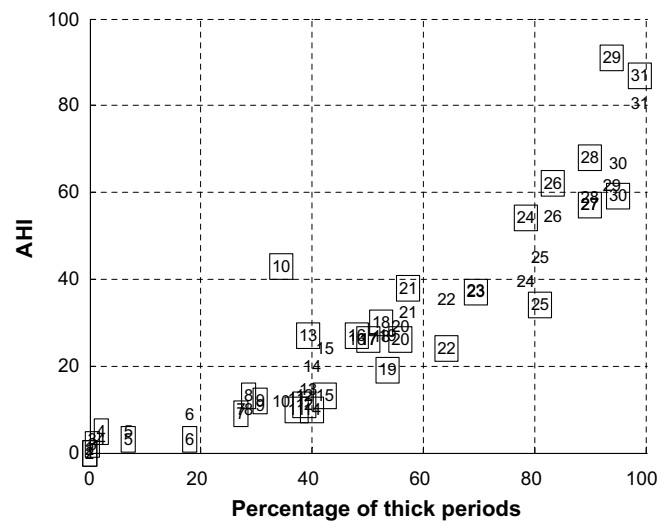

Fig. 5. Comparison of measurements and predicted values from regression model. Measurement points are boxed numbers, predicted points are unboxed numbers.

4. Discussion

The results of this study suggest that tracheal sound analysis with a compressed signal might provide a reliable and efficient tool for screening sleep apnea. Previous studies of tracheal sound analysis have mainly focused on visual or automatical analysis of respiratory events (Van Brunt et al., 1997; Cummiskey et al., 1982; Nakano et al., 2004). This study concentrates on the compressed tracheal sound characteristics in which the whole night can be squeezed into one page for visual examination of different respiratory patterns.

Three distinct patterns of compressed respiratory sound signal could be observed: thick, plain and thin. The patterns were easily recognized. During periods of plain signal curves the minimum and maximum levels of the signals were rather constant and close to zero indicating that the sound levels recorded remained low. The proportion of round flow shape measured by nasal flow pressure was most substantial during the pure plain periods suggesting that this sound pattern represents mostly normal breathing. However, few subjects had apneas and hypopneas during pure plain periods. Therefore, the assumption that a plain signal curve reflects normal breathing needs further examination.

The thick signal curve represents highly variable sound levels. Visually, it seemed to be composed by intermittent alternation of disturbed breathing during apneas/hypopneas with some normal breaths between apneas/hypopneas. The correlation studies between AHI and the amounts of different sound signal patterns during whole night indicated that the higher the AHI the higher is the amount of the thick sound signal. Conversely, the higher the proportion of plain period time, the lower the AHI. The thin period time did not show significant correlation with AHI. Based on Figs. 4 and 5, it seems that when the nocturnal total time of the thick signal is over 30% the limit of abnormal AHI (>5/h) is clearly outreached.

In addition, the representative 10-min sequences of compressed sound periods proved to differ in the amount of respiratory events (apneas and hypopneas). Apneas and hypopneas were most common during pure thick periods, whereas during pure thin and plain periods only few subjects had apneas and hypopneas. All the subjects presenting apneas or hypopneas during pure thin and plain periods had increased AHI in the whole night analysis.

The thin sound signal pattern was substantially distinct from the thick and plain sound signals. In this pattern, both curves deviate clearly from the zero level reflecting episodes of loud continuous noise. During pure thin sound periods the proportion of flattened, non-round inspiration flow shape was highest and some subjects had apneas and hypopneas. Therefore, the thin sound pattern seems to represent breathing with abundant flow limitation and sometimes apneas.

The importance of prolonged flow limitation is still unclear. End-tidal carbon dioxide has been shown to rise during induced flow limitation (Calero et al., 2006). Our previous study demonstrated a rise in transcutaneous carbon dioxide tension during prolonged spiking in Emfit sensor, which occurred with simultaneous flow limitation in the nasal pressure signal (Rauhala et al., 2007). Together these studies implicate that a prolonged flow limitation, even without apnea and hypopnea type periodic respiratory alternation, could indicate a breathing disturbance, which might have significant metabolic effects.

In this study, we used HeLSA-tracheal sound analyzer, which has initially been developed for the diagnostics of lung diseases. To maximize the accuracy of the measure-

ments a 11,025 Hz sampling rate is used. The pulmological studies are quite short and the collected amount of data is still reasonable. However, sleep recordings last several hours and with the high sampling rate the amount of data is huge which increases also the duration of analysis. In this study heavy data compression was used to reduce the amount of data substantially, to speed up the evaluation of the data and to improve the visualization of the signal.

Besides tracheal sound analysis, a wide range of devices have been used for portable monitoring of sleep apnea. An extensive review of the subject has been produced by the American Academy of Sleep Medicine, the American College of Chest Physicians and the American Thoracic Society (Flemons et al., 2003). The review stated that the gold standard polysomnography requires technical expertise, and is labor-intensive and time consuming. Thus, there is a growing interest in the alternative approaches to diagnosis. In the review, portable monitors were categorized into four types. The simplest type 4 monitors (continuous single or dual bioparameters) measured one, two or three variables and reported on these either individually or in combination. The single variables measured included oximetry, heart rate, airflow, respiratory inductance plethysmography, esophageal pressure and snoring. In the review, it was summarized that several portable monitors show a promise for excluding disease, other ones for confirming disease, and some for doing both. All the different monitors have their advantages and disadvantages.

The nasal cannula is today the method of choice in detecting airflow limitation and upper airways obstruction. Thermistor is no longer recommended alone. Common experience is, however, that the nasal pressure transducer can detach during the night. Nasal cannulas can markedly increase the nasal airflow resistance in subjects presented with nare narrowness and/or deviated nasal septum (Lorino et al., 2000). Although subjective nasal obstruction does not necessarily impair the accuracy of nasal pressure monitoring (Thurnheer et al., 2001), it is possible that impaired nasal ventilation prevents adequate measurements of nasal pressure in some subjects (Series and Marc, 1999). Also, the oximeter, often causes discomfort and can be detached. All the patients are not able to put the transducer on by themselves. It is neither considered reliable to be used alone. Heart rate measurement is indirect and cannot be used for patients with impaired autonomic nervous system function. Respiratory inductance plethysmography is reliable when used appropriately. However, it needs special equipment and a set of transducers of different sizes for different subjects. The initiation of the recording also takes time because the system needs to be adjusted.

The sound recording device, a sensitive but small microphone, may interfere with sleep less than the nasal pressure transducer or the other transducers used. Among two of the first patients the microphone became detached, but after the fixation was standardized the detachments ceased. Thus tracheal sound analysis can provide a more reliable

non-invasive way of showing prolonged flow limitation during sleep in some subjects. The compressed recording is also much more rapid to analyse visually than the methods presented above. It is so far not recommended to use automatic analysis of sleep recordings alone.

Compressed tracheal sound analysis might provide a promising screening method for obstructive apneas and hypopneas. The compressed sound curve is highly visual and the proportions of various breathing patterns can be easily estimated. Naturally, the method has to be replicated and validated. Although it has been previously shown that the tracheal sound patterns are rather constant across nights in patients, the method should be tested by repeated recordings over multiple nights (Sanchez and Vizcaya, 2003). In the future also automatic or semiautomatic analysis could be developed to speed up the analysis. Many other screening devices can be used to distinguish between apneas and normal breathing, but with the presented compressed analysis also a third pattern with flow limitation can be observed. This additional feature warrants future studies. Recent work has shown that the flow limitation might be more important than has previously been understood. Compressed tracheal sound analysis seems to provide an efficient tool for large patient studies in order to examine the clinical significance of continuous flow limitation.

Acknowledgements

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