Evaluation of a non-invasive respiratory monitoring system for sleeping subjects

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Abstract. In a previous paper we introduced a non-invasive respiratory monitoring system (NIRMS) for monitoring respiratory movements during sleep. Unlike standard sleep laboratory methods, the NIRMS can be used in frail older adults to describe breathing patterns during sleep that will mark individuals with declining neurological function. The present study evaluates the use of the NIRMS as a respiratory monitor and identifies variables that can reliably detect changes in breathing patterns and the presence of other body movements.

Data were obtained from eleven healthy adults (six women, five men) whose body mass indices ranged from 20 to 47 kg m⁻², and whose baseline respiratory rates ranged from 4 to 19 breaths per minute. We evaluated three variables derived from frequency and amplitude measurements of the NIRMS: (1) the interval between breathing cycles (the interbreath interval or IBI); (2) the period between breathing cycles (the interbreath frequency or IBF); and (3) the amplitude of breath cycles (AMP).

The frequency of NIRMS waveform deflections correlated highly with the frequency of visually observed chest movements (r = 0.99). Compared with a subject's baseline, a standard deviation of IBI > 3.0 s consistently identified time segments with three or more apnoeic events. The IBF and AMP differentiated respiratory from other body movements. An IBF > 20 cycles/s or an AMP > 0.4 V identified experimentally introduced body movements much more accurately than wrist accelerometry (NIRMS detected 99% of events, kappa = 0.90; WA detected 50%, kappa = 0.70). These findings support the use of the NIRMS in monitoring changes in breathing patterns during sleep, especially in frail and cognitively impaired subjects.

Keywords: breathing, movements, non-invasive monitoring device, sleep, older adults

1. Introduction

Frail and cognitively impaired older persons often cannot tolerate the intrusiveness of the laboratory environment and the apparatus normally used to monitor respiratory activity during sleep. Less intrusive methods using piezoelectric sensors have been used successfully to monitor sleep, movement and respiratory activity in infants (Holditch-Davis 1990) and healthy older adults (Acebo *et al* 1991). In a previous paper (Hernandez *et al* 1995), we described a pressure-sensitive sensor (the non-invasive respiratory monitoring system or NIRMS), designed to monitor respiratory movements of frail and cognitively impaired subjects during sleep. Unlike standard sleep laboratory methods (which requires the use of face masks, eosophageal balloons and thermocouples), the NIRMS can be used for long-term monitoring in the home and thus introduces the possibility of describing breathing patterns during sleep that may mark individuals with declining neurological function (Carlson 1997).

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NIRMS development and operation is described fully in our earlier paper, but we review here briefly the design of the system. The NIRMS uses a pressure transducer to measure changes in air pressure inside an inflatable mattress (figure 1). A crucial part of the design is a mechanical filter that protects the transducer from large pressure changes caused by gross body movements. This allows detection of subtle movements of the subject's chest. The pressure sensor does not require skin contacts or wires that increase the subject's risk for shock or mechanical entanglement, a concern in those who may have incontinence during sleep or need to get out of bed during the night (Gardner and Hujcs 1993, Bliwise et al 1990).



Figure 1. Schematic illustration of the NIRMS. (From Hernandez et al 1995, reprinted with permission.)

An example of the waveform derived from the NIRMS pressure transducer is shown in figure 2. The horizontal axis indicates the time, in seconds; the vertical axis plots the amplitude of the waveform, expressed in volts. Downward deflections of the NIRMS waveform correspond to the rise of the chest during inspiration (I), and upward deflections correspond to the return of the chest to a resting state during expiration (E). We can use these deflections to identify breath cycles. And since the NIRMS senses full body pressure, the amplitude of the NIRMS waveform can identify body movements during sleep and yield a rough estimate of the continuity of sleep.

In this study we wanted to see whether we could reliably identify breathing cycles, apnoeic events and other body movements using the NIRMS. A secondary aim was to identify derived variables that would allow on-line, computer-generated scoring of breathing patterns and detection of non-respiratory body movements.

2. Methods

2.1. Subjects

Data were collected on eleven healthy subjects (five men and six women), aged 18 to 41 years and having a wide range of body mass index (20 to 49 kg m^{-2}). The subjects, wearing loosefitting clothes, were placed on a hydraulic bed, with the head of the bed flat. The trials were conducted in the morning, between 8 and 10 a.m. Room temperature was maintained between



Figure 2. Example of an NIRMS waveform. Time scale = 47 s (1.33 div^{-1}) . Each division along the vertical axis equals 0.008 V. A breathing cycle is defined by a set of negative and positive deflections. I = end inspiration, E = end expiration. The higher-frequency oscillations are cardiac in origin (Carlson 1997).

22 and 24 °C, and the relative humidity between 19 and 20%. The study was approved by the University of North Carolina School of Nursing Institutional Review Committee for the Protection of Human Subjects.

2.2. Subject procedure

2.2.1. Measurement of breathing cycles. While breathing at their own inherent rates, the subjects were monitored for 5 min in the supine position and for another 5 min in left side-lying position (baseline breathing). Between the supine and side-lying recordings, three apnoea simulations were carried out during a 5 min period in the supine position. During the apnoea simulations, subjects took three deep breaths, stopped breathing for 10 s after expiration, and finished with three more deep breaths.

Two methods, visual observation and capnography, were used to identify inspiration and expiration independent of the NIRMS waveform. Visual observation required that an observer press an event marker, which recorded a 1 V signal, indicating a visible rise of the chest or abdomen. A capnograph (Oxicap 4700, Ohmeda, Boulder, CO), which records the carbon dioxide concentration of air at the nose, was used to identify the expiratory phase of each breathing cycle. The visually observed and capnograph signals were each collected at a rate of 15 samples/s and stored as separate channels on the same computer file as the NIRMS waveform.

2.2.2. *Measurement of other body movements.* A 21 min recording was made to evaluate the sensitivity of the NIRMS for detecting non-respiratory movement. Experimentally introduced

movements, designed to simulate those typically observed in sleeping subjects, consisted of (1) turning to the right or left and then supine, (2) crossing and uncrossing the legs, (3) bending and straightening the legs, (4) placing one or both hands on the chest, to the side of the body, and under the pillow, and (5) bending and straightening the arms.

In addition to the NIRMS, body movements were monitored with an accelerometer (HMS-5000, Vitalog Monitoring, Redwood City, CA) worn on the subject's left wrist. Signals from the accelerometer were stored directly in a second computer. The accelerometer and NIRMS waveform files were synchronized by recording a shared event marker at the beginning and end of each trial. Finally, an observer marked the time of each body movement on a protocol checklist, using a stopwatch to determine the time elapsed from the start of each recording. This observer was blinded to both the NIRMS and wrist accelerometer waveform data, so the times indicated on the protocol checklist were used as a standard for comparing the sensitivity of the NIRMS data against wrist accelerometry.

2.2.3. NIRMS waveform analysis. To evaluate the sensitivity of the NIRMS in detecting and counting breathing cycles, we generated separate NIRMS waveform and visual observation recordings of the 5 min periods of regular (supine and left side-lying baseline) and irregular (simulated apnoeic) breathing. The 66 waveforms (22 supine baseline, 22 side-lying baseline and 22 simulate apnoea records) and simultaneous visual observation recordings were given to a rater (BWC) who was instructed to count the number of downward and upward deflections on the NIRMS waveform and the number of 1 V signals on the event marker waveform. The visual observation and NIRMS waveforms were independently assigned a random alphanumeric code so that the rater could not match NIRMS waveforms with their corresponding visual observation waveforms. The reliability of counting breathing cycles over a 5 min period was determined by subtracting the visual observation count from the corresponding NIRMS and visual observation measures, was used to evaluate reliability.

We used a computer-assisted procedure to calculate variables to be used in detecting apnoeic events and other body movements. A commercially available peak-valley capture algorithm (Windaq, Dataq Instruments, Columbus, OH), which allows the user to specify a threshold, identified points of maximum (expiration) and minimum (inspiration) voltage on the NIRMS waveform. The computer-identified events were manually compared with the subject event and capnograph waveforms, and the threshold lowered until all maximum and minimum points were recognized. Then each pair of verified, consecutive, maximum and minimum values was analysed with a statistical package (SAS, SAS Institute, Cary, NC).

We used the points of minimum amplitude (nadir) to mark the occurrence of breathing cycles (figure 3). We then measured the interbreath interval (IBI) and the amplitude of each breathing cycle (AMP). Measured in volts, the amplitude of each breathing cycle is a measure of chest expansion during inspiration. IBI measures the duration, in seconds, of a respiratory cycle. From the IBI, we derived a third variable called the interbreath frequency (IBF). Expressed in cycles/s, the IBF is obtained by multiplying the inverse of the IBI by 60 (60/IBI) and is used to provide an instantaneous measure of the frequency of each breathing cycle.

Using the UNIVARIATE procedure in SAS, we determined the mean and standard deviation of IBI, IBF and AMP values for each of the 11 5-min segments of baseline and 11 5-min segments of simulated apnoea. Differences between the mean and standard deviation of each variable on baseline and on simulated apnoea recordings were tested using paired *t*-tests (df = 9); a probability of less than 0.05 was used to identify statistically significant differences between baseline and simulated apnoea values.



Figure 3. Breathing cycle measurements. Bold vertical arrows = breathing cycles, thin vertical arrow = AMP, thin horizontal arrow = IBI.

We established thresholds for identifying non-respiratory movements on the NIRMS waveform as follows: differences in the frequency of IBI and IBF deflections were established by pooling data across each type of record and determining the percentage of IBF values that were >20 cycles/s. At rest, the IBI ranges between 3 and 5 s (Milic-Emili 1982). The IBI of an apnoeic event lasts at least 10 s (Acebo *et al* 1991), and in sleep can be up to 120 s (Krieger 1994). The IBF is the product of the reciprocal of IBI times a constant (60). This means that the IBF should be between 12 and 20 cycles/s at rest and 6 cycles/s or less for apnoeic events. Since there are no established norms for classifying AMP values, we chose a threshold based the maximum and minimum AMP values by comparing the subjects' simulated apnoea recordings against their non-respiratory movement recordings.

The 21 min recordings with experimentally introduced body movements were used to examine the accuracy of these threshold values. We divided the 11 NIRMS records into 21 1-min epochs. We assigned a score of '1' to epochs that demonstrated body movement and a score of '0' to epochs that did not. In a similar fashion, we divided the 11 accelerometer records and assigned a score of '1' to 1 min epochs that showed a rise above baseline. Prior to scoring, identifying information (subject identification code, dates and times) was removed from each record and the records were randomly assigned an alphanumeric code. This procedure prevented us from matching the subject's NIRMS record with their accelerometer record.

The accuracy of each method was evaluated by examining the level of agreement of each method against a protocol checklist, which documented movements by direct visual observation. Agreement amongst the three methods was determined by calculating the kappa coefficient, an index of agreement beyond that expected by chance (Cohen 1960). A kappa coefficient of 1 represents perfect agreement; values of greater than 0.70 are considered to be significantly greater than chance and were used to establish accuracy.

3. Findings

3.1. Frequency of breathing cycles (baseline and simulated apnoea)

NIRMS and visual observation counts during each 5 min period for each subject are shown in table 1. The frequency of breathing ranged from 4 to 19 breaths per minute. Of the 33 paired

Table 1. Frequency of breathing cycles of each 5 min segment: NIRMS = non-invasive respiratory monitoring system, VO = visual observation.

Subject	Supine		Side-lying		Simulated apnoea	
	NIRMS	VO	NIRMS	VO	NIRMS	VO
1	51	50	62	62	36	35
2	21	21	36	36	26	27
3	84	84	66	66	48	48
4	72	73	61	61	48	47
5	73	72	46	47	52	53
6	41	41	43	40	40	39
7	74	73	65	67	51	52
8	97	97	92	92	44	44
9	79	78	78	78	43	40
10	27	27	39	40	38	39
11	96	95	95	96	65	61

(visual and NIRMS) counts of breathing cycles, 29 agreed perfectly or differed by no more than one count. A high correlation between the frequency of breathing measured by the NIRMS and by visual observation was found for both baseline recordings ($r_{supine} = 0.99$, $r_{side} = 0.99$) and for the simulated apnoea recordings ($r_{periodic} = 0.99$). There was disagreement between the NIRMS and visual observation in four paired observations (highlighted pairs). Unlike the supine baseline recordings, the side-lying and simulated apnoea recordings contained a higher number of spontaneous body movements.

3.2. Comparison of the NIRMS waveform recordings

Figures 4 and 5 show representative baseline and simulated apnoea recordings. Each contains separate channels displaying the NIRMS waveform (top), capnograph (middle) and the visual event marker (bottom) signals. In both recordings, the downward deflection on the NIRMS waveform occurred at approximately the same time as the 1.0 V signals on the event marker waveform, indicating inspiration (I). Conversely, each upward deflection on the NIRMS waveform corresponded with the rise and plateau on the capnograph waveform, indicating expiration (E).

The amplitude of the baseline waveform (figure 4) ranges from 0.04 to 0.06 V and the interval between points of minimum amplitude ranges from 3.8 to 10.5 s. The inspiratory phase of each cycle tends to be twice the duration of the expiratory phase, which is characteristic of a respiratory signal (Milic-Emili 1982). At the point indicated by the arrow, the interval between cycles abruptly shortens after the fifth breathing cycle. The increased frequency of breathing that follows the arrow can be seen on all three waveform channels and corresponds to the onset of snoring. Although we asked the subjects to remain awake during the trials, this subject reported that he had not slept well the night before and had fallen asleep during this 5 min segment.

Compared with baseline, the simulated apnoea records had greater changes in both the amplitude and frequency of breathing (figure 5). The amplitude of the NIRMS waveform increased from approximately 0.06 at baseline to over 0.1 during deep breathing. The three apnoeic events are denoted by the three lines above the NIRMS waveform. Although the three deep breaths preceding the apnoea events caused the waveform to exceed baseline expiratory level, the waveform returned to baseline during each apnoeic event.



Figure 4. Representative supine baseline record. Time scale = 77 s (2.67 s div⁻¹). Each division along the vertical axis equals 0.053 V. I = inspiration, E = expiration. Note that all three waveforms show the spontaneous increase in the frequency of breathing cycles.



Figure 5. Representative simulated apnoea waveform record. Time scale = $280 \text{ s} (9.33 \text{ s} \text{ div}^{-1})$. Each division along the vertical axis equals 0.053 V. The three vertical lines on the NIRMS waveform indicate the occurrence of an apnoeic event. Maximum amplitude is 0.2 V.

Non-respiratory body movements produced greater changes in amplitude of deflection on the NIRMS waveforms than did the 'deep breaths' of the simulated apnoea records. Figure 6 shows a representative record with a small (marked 1) and a large body movement (marked 2). The full lines on the NIRMS channel indicate the start and end of each other body movement. Movement 1 occurred when the subject raised his right arm. This movement produced a large upward deflection on the NIRMS signal (amplitude = 1.781 V). Movement 2 occurred when the subject turned from the supine to the left side-lying position. This movement caused both a downward and an upward deflection on the NIRMS waveform (>5.632 and 1.12 V, respectively).



Figure 6. Example of two body movements on the NIRMS waveform. Time scale = 133.3 s (5.33 s div⁻¹). Each division along the vertical axis equals 0.333 V. Movement 1 = raising the right arm. Movement 2 = turning from supine to left side-lying position.

The above analysis suggests that variables derived from measurements of the interval between negative deflections on the NIRMS could be used to characterize patterns reflecting changes in the frequency of breathing. In addition, body movements produced large increases in the amplitude and frequency of the NIRMS signal that together could differentiate respiratory from other body movements. We then applied a peak valley capture algorithm to mark the waveform and derive variables that would identify within-subject changes in breathing pattern (difference between baseline and simulated apnoea) as well as identify other body movements on the NIRMS waveform.

3.3. NIRMS waveform analysis

3.3.1. Simulated apnoea recordings. The interval between breaths (IBI) and two calculated values (the mean and standard deviation of IBI) reliably identified changes in a subject's breathing patterns. IBIs ranged from 1.8 to 20.7 s, with larger values reflecting a lower

frequency of breathing. Pooling across the 22 5 min segments, the mean IBI of segments with simulated apnoeic events ($M_{apnoea} = 11.2 \text{ s}$) was significantly greater than the baseline ($M_{baseline} = 5.59 \text{ s}$), paired t (df = 9) = -2.38, p < 0.044. However, the mean IBI is a less sensitive indicator of apnoea in those with baseline respiratory rates of less than 8 breaths/min. Two of our subjects had a baseline respiratory rates of less than 8 breaths/min. In these cases, it was difficult to differentiate slow regular breathing patterns from breathing patterns which contain apnoeic events.

Regardless of the subject's baseline respiratory rate, the standard deviation of IBI clearly identified segments containing simulated apnoeas. The standard deviation of IBI increased from 1.34 s during baseline to 5.84 s during simulated apnoea (paired t = -3.64, p < 0.003, df = 9). All segments with simulated apnoeic events had a standard deviation of IBI greater than 3.0 s.

3.3.2. Non-respiratory movements. Both the amplitude of breathing cycles (AMP) and the interbreath frequency (IBF) were sensitive measures for identifying non-respiratory movements. Ninety per cent of non-respiratory movements had AMP values of greater than 0.40 V and 80% had IBI values of less than 3 s. The corresponding IBF of these non-respiratory movements was 20 cycles per second or greater. By using a combined threshold (either an IBF of greater than 20 cycles per second or an AMP of >0.40 V or both), we were able to detect a greater number of non-respiratory movements.

The protocol checklist and wrist accelerometry data were used to test the sensitivity of the NIRMS thresholds for non-respiratory movement. Of the 231 1-min movement epochs, 139 were identified on the protocol checklist as containing non-respiratory body movements. Ninety-nine per cent of these movement epochs were identified using the NIRMS thresholds (kappa = 0.97). The NIRMS thresholds identified all those detected by wrist accelerometer, but only 50% of the movements noted on the protocol checklist were accompanied by detectable deflections of the wrist accelerometer waveform (kappa = 0.56). The movements detected by the wrist accelerometer detected changes in body position and left arm movements, but failed to detect isolated leg and right arm movements.

4. Discussion

A number of non-invasive methods can be used to monitor breathing during sleep. Most systems use microprocessors and microcomputers, and a combination of monitoring techniques such as inductance plethysmography, nasal/oral thermocouples, or pulse oximetry to monitor respiratory activity (Krieger 1994). Although portable, all of these methods run cables from the subject to a bedside recording unit. The cables not only limit the subject's mobility, but important data can be lost if they become detached over the course of the night (Bliwise *et al* 1990).

Our experience with the NIRMS demonstrates the strengths and limitations of using a fullbody, pressure-sensitive sensor to monitor breathing. Our analysis of this sample of subjects with a wide range of body weights showed that the NIRMS was as accurate as visual observation in counting breathing cycles during quiet breathing and during simulated apnoea. By estimating the standard deviation of interbreath intervals, we were able to discriminate between subjects' resting and simulated apnoea waveform records. Given the inherent difficulty of using visual observation to quantify the regularity of breathing (Krieger 1994), the IBI and its derived variable, IBF, offer the opportunity to quantify the magnitude of change in the breathing patterns.

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Secondly, our findings show that the NIRMS is an excellent movement sensor. In general, the magnitude of the deflections during body movements are significantly greater than during respiration. After a body movement, it takes 5–10 s to return to the typical respiratory movement voltage range of 0.01–0.2 V. Since the signals generated by non-respiratory body movements differ drastically from those during breathing cycles, we were able to develop a protocol, based on AMP and IBF, to discriminate respiratory from other body movements on the NIRMS waveforms. We found that non-respiratory body movement signals had amplitudes in excess of 0.40 V, a value much greater than that seen in either quiet breathing or simulated apnoea recordings. In addition, body movements were characterized by IBF values of greater than 20 cycles/s. Our body movement detection procedure, based on IBF and amplitude thresholds, was more sensitive than even a generous rule for marking movements on the wrist accelerometer waveform.

Not shown were three left side-lying recordings that showed a shift in the polarity of the NIRMS waveform. This shift in polarity did not adversely affect the ability to count and measure the frequency or amplitude of breathing cycles, because the shift was consistent throughout the subjects' side-lying recordings. Unlike visual observation, the NIRMS could reliably measure breathing cycles when the subject turned away from the observer.

Due to the very high amplitude of deflections during non-respiratory body movements, we could not measure IBI during and immediately after a body movement. However, similar problems were encountered with visual observation. As stated above, the NIRMS is no worse than visual observation when monitoring breathing cycles during body movements and, in some cases, is better than visual observation.

In summary, the NIRMS represents a safe, minimally intrusive method that can be used for long-term monitoring of either medically frail or cognitively impaired older adults. Unlike current movement-sensitive systems (Alihanka *et al* 1991, Acebo *et al* 1991, Thoman *et al* 1994), the NIRMS is a safe method for monitoring breathing in older adults because it lessens the risk of shock in those who become incontinent, have open wounds or use electronic medical devices (Gardner and Hujcs 1993, Olson 1992). As with other movement-sensitive systems, the NIRMS can characterize breathing patterns based on the frequency and periodicity of breathing.

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