34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Thoraco-Abdominal Asynchrony in Children during Quiet Sleep using Hilbert Transform

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***Abstract*—We present a technique based on the Hilbert transform to quantify the thoraco-abdominal asynchrony (TAA) based on the phase shift between ribcage (RC) and abdomenal (AB) breathing signals acquired using respiratory inductive plethysmography (RIP). We employed this method to investigate RIP during overnight polysomnography (PSG) in 40 healthy children for analysis of their breathing patterns in various stages of sleep (ss 2, 3, 4 and REM) and in two common sleeping positions (supine and lateral). RIP signals free of respiratory or movement artifacts were segmented into 30 second epochs. Those epochs with maximum power in the quiet breathing frequency range and positional invariance throughout were included for further processing. TAA was calculated from corresponding RC and AB excursions. We found a statistically significant influence of sleep position on the level of TAA in all stages of non-REM sleep. In conclusion, the Hilbert transform provides a simple tool for the quantification of thoraco-abdominal asynchrony.**

I. Introduction

During quiet tidal breathing, exertion of the ribcage (RC) and abdomen (AB) tend to exhibit a coincident motion resulting in a high level of synchrony during the respiratory cycle. However this synchronous movement is influenced by upper airway resistance and respiratory muscle activity which are distinctly different between wakefulness and sleep [1]. In normal healthy adults, inspiratory and expiratory airflow resistance has been shown to vary with sleep stages [2] and quantification of thoraco-abdominal asynchrony (TAA) might provide insights into changes in respiratory mechanics across the sleep period. Our objective was to develop a robust method to quantify TAA derived from paediatric polysomnography (PSG) where an elevated asynchrony is considered an indicator of obstructive sleep apnea (OSA) and/or muscle dysfunction [3, 4].

In earlier studies, TAA was assessed based on the width of the loop formed on a Lissajous figure of RC vs. AB signals, called Konno Mead plots [5]. Thereafter, several other methods have been proposed in the literature to estimate the relative phasing during breathing [6-8]. The Labor breathing index (LBI), that reflects both the phasic relationships of RC and AB movement, and the volume displacement due to that movement proved to be effective but is restricted to calibrated inputs which are difficult to perform in children [9, 10]. Waveform independent approaches such as maximum linear correlation yield estimates on uncalibrated signals but are computationally intensive [11]. An automated phase estimation between two time series based on digital phase detection was proposed by Motto *et al.* [12]. This approach has been shown to offer less biased estimates than the Pearson's correlation method and has been successfully implemented in studies evaluating infants at risk of postoperative apnea [13].

In this study, we (a) introduce a TAA estimation method based on the Hilbert transform, (b) employ the technique to study dual band respiratory inductive plethysmography (RIP) signals of normal children during quiet sleep free of movement or respiratory artifacts and (c) observe the effect of different sleep stages (ss 2, 3, 4 and REM) and sleeping positions (supine – lying on back and lateral – lying on sides) on the level of asynchrony. This baseline quantification of asynchrony levels in healthy children during quiet sleep may provide normative data for clinical studies. We hypothesize that TAA is influenced by sleep stages and sleep position.

II. Methods

A. Phase estimation using Hilbert transform

The Hilbert transform gives the instantaneous amplitude and phase of a signal $x(t)$ via construction of an analytical signal $\zeta(t)$ which is a complex function of time [14] defined as

$$\zeta(t) = x(t) + j\tilde{x}(t) = Ae^{j\phi(t)}, \qquad (1)$$

where $\tilde{x}(t)$ is the Hilbert transform of $x(t)$. The instantaneous amplitude and phase are given by

$$A(t) = \sqrt{\left(x^2(t) + \tilde{x}^2(t)\right)} \qquad (2)$$

$$\phi(t) = \arctan\frac{\tilde{x}(t)}{x(t)} \qquad (3)$$

Given two signals $x_1(t)$ and $x_2(t)$, the relative phase between the two signals can be obtained [15] via their Hilbert transforms $\tilde{x}_1(t)$ and $\tilde{x}_2(t)$ as

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$$\phi_1(t) - \phi_2(t) = \arctan\left[\frac{x_1(t)\tilde{x}_2(t) - x_2(t)\tilde{x}_1(t)}{x_1(t)x_2(t) + \tilde{x}_1(t)\tilde{x}_2(t)}\right] \quad (4)$$

To validate the approach, RC and AB breathing patterns were simulated using sinusoidal approximations $r(t)$ and $a(t)$ respectively,

$$r(t) = A_1 \sin(2\pi * \mu * t) \quad (5)$$

$$a(t) = A_2 \sin(2\pi * \mu * t + \phi) \quad (6)$$

where A_1=1, A_2 = 0.8, μ = 0.3 Hz and $\phi \in [0, \pi]$

Further, white Gaussian random noise with different noise levels was added to the simulated signals such that the resultant signal-to-noise ratio (SNR) ranged from [1dB - 30dB]. An example of the breathing patterns simulated using (5) and (6) with ϕ=0.5π and SNR of 10dB is shown in Figure 1. The performance of the phase estimator was evaluated with respect to different levels of phase shift ϕ [0.2π – 0.8π] and SNR [1dB - 40dB] and quantified as the estimation error. It was observed that the error of the estimate were notably higher in noisy patterns (SNR <~12dB) but for higher levels of SNR, the estimates were consistent (Figure 2).

B. Subjects

The study comprised of 40 (20 male) non snoring healthy children with no underlying medical conditions or respiratory disorders. This study has appropriate ethics approval and has been part of previously reported studies [16, 17]. The age and BMI z-score of the subjects ranged between 3.1-12.2 years (mean ± SD: 7.7 ± 2.6 yrs) and -1.7-2.3 (mean ± SD: 0.3 ± 0.8) respectively.

C. Overnight polysomnography

Standard PSG parameters were recorded continuously overnight from the subjects using the S-Series® Sleepwatch system (Compumedics, Australia). Respiratory movements

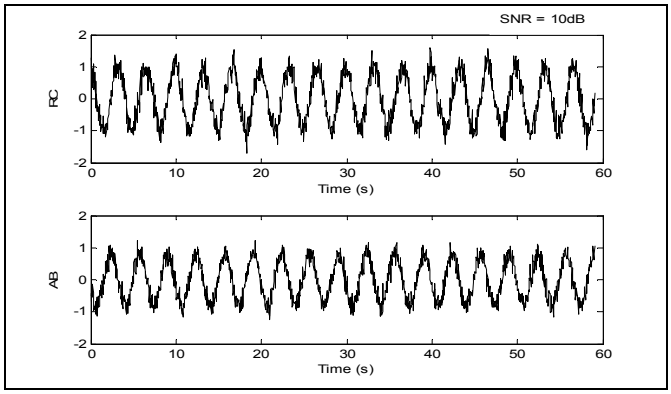

Figure 1. Simulated RC and AB signals corrupted with additive Gaussian noise

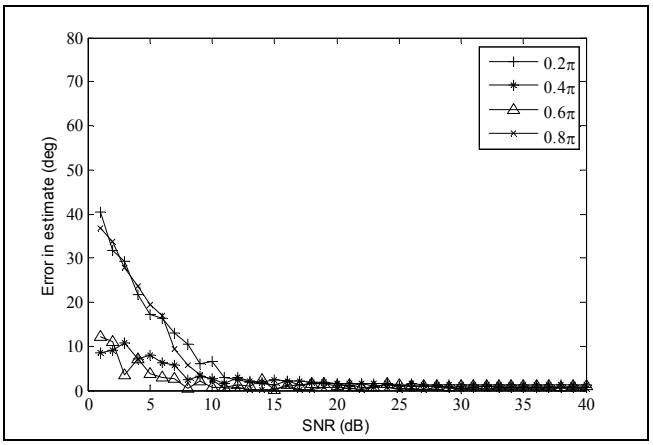

Figure 2. Relative errors of TAA estimated between the two simulated respiratory signals for different levels of SNR.

of the chest and abdominal wall were recorded using uncalibrated RIP. Each child was monitored continuously overnight via infrared camera and by a pediatric sleep technician who also documented observations of sleep behavior including the presence or absence of snoring. More details on the study protocol are published elsewhere [18]. Sleep stages were scored visually in 30 second epochs according to the standardized EEG, EOG and EMG criteria of Rechtschaffen and Kales [19].

D. *Respiratory signal processing*

Respiratory data digitized at 25 Hz were extracted from the thoracic and abdominal RIP channels of the PSG. Custom written computer software was developed under MATLAB® for signal processing. Signal offsets were removed by subtracting the time average. All 30 second epochs distorted by respiratory or movement artifacts as scored by the sleep technicians were excluded from further analysis.

E. Power based filtering

In all 40 children, quiet breathing segments were analyzed breath-by-breath and breathing rate was computed based on inspiratory expiratory timings. The range of breathing frequencies observed was [0.15 to 0.5 Hz] and this was hypothesized as the spectrum of quiet RC and AB signals. Hence to increase reliability of the estimate, a power based segmentation procedure was adopted that categorized each input epoch into either quiet breathing (i.e., 0.15 to 0.5 Hz) or artifact (i.e., outside the QB range) [20]. This was accomplished using a bank of 13 elliptic IIR BP filters each with a pass band ripple of 0.1 dB and a stop band attenuation of 30 dB. These filters span frequencies between 0 Hz and 2 Hz each with a pass-band width of 0.2 Hz allowing an overlap of 0.05 Hz with the neighboring filter. For example, pass band of filter I is (0-0.2 Hz), filter II is (0.15-0.35 Hz), filter III is (0.3-0.5 Hz) and so on. Each epoch entered the set of band pass filters simultaneously. Average signal power of each filter output was computed and values compared. Since it is assumed that QB is between (0.15 to

0.5 Hz), those epochs exhibiting maximum signal power in either the (0.15-0.35 Hz) band or (0.3-0.5 Hz) band were alone selected for further processing. These filtered epochs were checked for sleep position (recorded every second) and were included for analysis only if position unchanged within an epoch. The same process was applied to the AB RIP signal as well.

F. TAA estimation

The processed RC and AB signals were extracted epoch-by-epoch over the entire sleep duration and the phase deviation between the epochs was estimated using the Hilbert approach (eq. (4)). To validate our proposed method against a recent technique that has been shown to perform better than conventional approaches, the XOR-based phase detection [12] was also employed on all epochs. This involves detection of the portion of time that one signal is different from the other using binary convertors and an XOR gate, which in turn provides an indication of the phase relation. Asynchrony was estimated using each method and were averaged for each individual within respective sleep positions in each sleep stage so that each child contributed equally to the group mean.

G. *Statistical Analysis*

Data were analyzed using the statistical software SPSS version 18 and the Graph Pad Prism Inc. version 5.01 for windows. Two-way ANOVA for repeated measures was used to test for differences in TAA measure between sleep stages and between supine and lateral positions. Student's paired t-test and one-way RM-ANOVA were used to compare positions and sleep stages respectively. Associations between TAA measure and age, BMI and gender were determined using Pearson correlation analysis. Data are presented as mean ± SD and p values are 2-tailed with statistical significance determined at $\alpha = 0.05$ unless stated otherwise.

III. Results

Demographic and PSG results have been reported previously [16]. The mean ± SD values of the TAA estimate during quiet tidal breathing in supine and lateral positions of sleep were analyzed. Both the Hilbert and XOR methods yielded results that were highly comparable (Table I). Results obtained using the Hilbert approach is discussed below:

A. Positional effect

The asynchrony estimate (TAA) was found to be significantly higher in the supine sleeping position than the lateral sleeping position. This positional effect was observed in the three stages of non-rapid eye movement (NREM) sleep (ss2: 19.4° vs. 15.9°, $p<0.001$; ss3: 21.25° vs. 16.83°, $p<0.05$; ss4: 22.1° vs. 16.6°, $p<0.01$). However, no significant positional effects were observed in REM sleep (Figure 3).

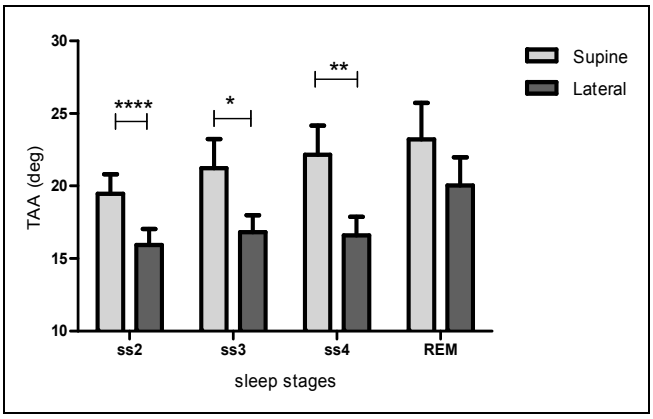

Figure 3. Comparisons of TAA between supine and lateral sleeping positions in each sleep stage. The significant differences due to positional effects are shown (*$p<0.05$, **$p<0.01$ and ****$p<0.001$).

B. Sleep stage Effect

To control for positional changes, only supine asynchrony estimates were analysed to evaluate the effect of sleep stage on TAA. We observed an increasing trend (from 19.4° in ss2 to 23.2° in REM) in the measure. But the differences were not statistically significant. Similar analysis of sleep stages in the lateral sleep position exhibited comparable TAA in the different stages of NREM sleep, but a trend for raised TAA in REM sleep. However, the differences were not statistically significant either (Figure 3).

C. Age, Gender and BMI Effects

The TAA measure was not significantly affected by age, BMI or gender in any stage of sleep or sleep position.

IV. Discussion

In this paper we applied a novel methodology that is based on the Hilbert transform to obtain an estimate of phase asynchrony between thoracic and abdominal breathing movement. The estimates are reliable except for very noisy signals with SNR values lower than 10dB. However, these situations are often overcome in signal processing applications where adequate filtering and artifact removal would increase the SNR sufficiently. The estimate being independent of tidal volume measures makes the method suitable for use with uncalibrated RIP signals. This is specifically advantageous in overnight sleep studies in infants and children where calibration is not feasible. Also the simplicity of computation makes the software implementation less complex than traditional methods like width measurement of Konno-mead loops or the Pearson's correlation. In addition, the results obtained using the Hilbert method were highly comparable with the XOR technique which in turn has been shown to offer less biased estimates than the Pearson's correlation method [12].

Applying the proposed Hilbert approach on a data set of overnight polysomnographic recordings, we have provided

TABLE I. Comparison of TAA between Hilbert and XOR approach in both supine and lateral sleep positions

Sleep stage	Supine		Lateral	
	Hilbert	XOR	Hilbert	XOR
Ss2	19.4 ± 7.9	19.6 ± 7.0	15.7 ± 6.6	15.9 ± 5.8
Ss3	20.3 ± 8.5	21.2 ± 10.1	16.5 ± 6.8	17.3 ± 6.7
Ss4	21.9 ± 9.2	22.9 ± 9.4	16.3 ± 6.7	17.0 ± 6.1
REM	24.3 ± 15.7	23.9 ± 15.6	19.1 ± 11.5	18.7 ± 10.9

normative data of baseline TAA levels in children during different stages of sleep in two common sleeping positions.

Our findings show that normal, healthy children exhibit a certain amount of phase shift (at least 15°) in the coordinated movement of their RC and AB compartments in all stages of sleep. This varies with the sleeping position in NREM sleep and increases when sleeping supine compared to sleeping laterally. Our results are in accordance with previous studies in infants and young children, where the LBI index has shown a TAA increase in supine compared to other sleeping positions using calibrated signals [21, 22].

These positional effects on TAA could be attributed to changes in the distribution of respiratory muscle forces influencing both antero-posterior (AP) and lateral diameters of RC and AB causing changes in their relative excursion [23]. From the lateral to supine position, the AP excursion of AB increases while both AP and lateral excursion of RC decrease. Thus changes in intercostal and abdominal muscle activity altering local compliance in parts of RC and AB might lead to an increased phase shift between the RIP breathing signals when in the supine position as compared to the lateral position. The physiological significance of these fluctuations in TAA and whether they form a part of cyclic regulation of sleep is yet to be understood.

Other factors such as age, gender and obesity have been reported to influence breathing patterns and TAA [6, 24]. However in our group of normal children age, gender and BMI showed no correlation with the TAA measure.

V. Conclusion

The novel Hilbert transform based approach for the quantification of thoraco-abdominal asynchrony provides a simple tool that might be useful in screening for sleep apnea and analysis of associated breathing patterns. This is important in understanding both physiological correlates of OSA as well as indicators of respiratory distress. In our study, the level of asynchronous thoraco-abdominal movement observed in children was affected by sleep position, but not by sleep stage.

References

[1]J. Gallego, S. Benammou, G. Vardon, B. Chambille, A. Denjean, and H. Lorino, "Influence of thoracoabdominal pattern of breathing on respiratory resistance," *Respiration physiology,* vol. 108, pp. 143-152, 1997.

[2]D. W. Hudgel, R. J. Martin, B. Johnson, and P. Hill, "Mechanics of the respiratory system and breathing pattern during sleep in normal humans," *Journal of applied physiology,* vol. 56, pp. 133-137, 1984.

[3]J. Hammer and C. Newth, "Assessment of thoraco-abdominal asynchrony," *Paediatric Respiratory Reviews,* vol. 10, pp. 75-80, 2009.

[4]J. Allen, M. Wolfson, K. McDowell, T. Shaffer, and R. Motley, "Thoracoabdominal asynchrony in infants with airflow obstruction," *The American review of respiratory disease,* vol. 141, pp. 337-342, 1990.

[5]Y. Sivan, T. Deakers, and C. Newth, "Thoracoabdominal asynchrony in acute upper airway obstruction in small children," *The American review of respiratory disease,* vol. 142, pp. 540-544, 1990.

[6]J. A. Verschakelen and M. G. Demedts, "Normal thoracoabdominal motions. Influence of sex, age, posture, and breath size," *American journal of respiratory and critical care medicine,* vol. 151, p. 399, 1995.

[7]R. Warren, S. Horan, and P. Robertson, "Chest wall motion in preterm infants using respiratory inductive plethysmography," *European Respiratory Journal,* vol. 10, pp. 2295-2300, 1997.

[8]A. D. Groote, J. Groswasser, H. Bersini, P. Mathys, and A. Kahn, "Detection of obstructive apnea events in sleeping infants from thoracoabdominal movements," *Journal of sleep research,* vol. 11, pp. 161-168, 2002.

[9]J. Kohyama, T. Shiiki, M. Shimohira, and T. Hasegawa, "Asynchronous breathing during sleep," *Archives of disease in childhood,* vol. 84, pp. 174-177, 2001.

[10]V. Kaplan, J. Zhang, E. Russi, and K. Bloch, "Detection of inspiratory flow limitation during sleep by computer assisted respiratory inductive plethysmography," *European Respiratory Journal,* vol. 15, p. 570, 2000.

[11]G. K. Prisk, J. Hammer, and C. J. L. Newth, "Techniques for measurement of thoracoabdominal asynchrony," *Pediatric pulmonology,* vol. 34, pp. 462-472, 2002.

[12]A. L. Motto, H. L. Galiana, K. A. Brown, and R. E. Kearney, "Automated estimation of the phase between thoracic and abdominal movement signals," *Biomedical Engineering, IEEE Transactions on,* vol. 52, pp. 614-621, 2005.

[13]K. A. Brown, A. A. Aoude, H. L. Galiana, and R. E. Kearney, "Automated respiratory inductive plethysmography to evaluate breathing in infants at risk for postoperative apnea," *Canadian Journal of Anesthesia/Journal canadien d'anesthésie,* vol. 55, pp. 739-747, 2008.

[14]D. Gabor, "Theory of communication. Part 1: The analysis of information," *Electrical Engineers-Part III: Radio and Communication Engineering, Journal of the Institution of,* vol. 93, pp. 429-441, 1946.

[15]M. Rosenblum and J. Kurths, "Analysing synchronization phenomena from bivariate data by means of the Hilbert transform," *Nonlinear analysis of physiological data,* pp. 91-99, 1998.

[16]M. Baumert, M. Kohler, M. Kabir, D. Kennedy, and Y. Pamula, "Cardiorespiratory response to spontaneous cortical arousals during stage 2 and rapid eye movement sleep in healthy children," *Journal of Sleep Research,* vol. 19, pp. 415-424, 2010.

[17]M. M. Kabir, D. A. Saint, E. Nalivaiko, D. Abbott, A. Voss, and M. Baumert, "Quantification of Cardiorespiratory Interactions Based on Joint Symbolic Dynamics," *Annals of Biomedical Engineering,* pp. 1-11, 2011.

[18]M. J. Kohler, K. Lushington, C. J. Van Den Heuvel, J. Martin, Y. Pamula, and D. Kennedy, "Adenotonsillectomy and neurocognitive deficits in children with sleep disordered breathing," *PloS one,* vol. 4, p. e7343, 2009.

[19]E. A. Wolpert, "A manual of standardized terminology, techniques and scoring system for sleep stages of human subjects," *Archives of General Psychiatry,* vol. 20, p. 246, 1969.

[20]A. Aoude, R. Kearney, K. Brown, H. Galiana, and C. Robles-Rubio, "Automated Off-Line Respiratory Event Detection for the Study of Postoperative Apnea in Infants," *Biomedical Engineering, IEEE Transactions on,* pp. 1-1, 2011.

[21]T. G. Oliveira, M. A. S. Rego, N. C. Pereira, L. O. Vaz, D. C. França, D. S. R. Vieira, and V. F. Parreira, "Prone position and reduced thoracoabdominal asynchrony in preterm newborns," *Jornal de Pediatria,* vol. 85, pp. 443-448, 2009.

[22]O. H. Mayer, R. G. Clayton Sr, A. F. Jawad, J. M. McDonough, and J. L. Allen, "Respiratory Inductance Plethysmography in Healthy 3-to 5-Year-Old Children*," *Chest,* vol. 124, pp. 1812-1819, 2003.

[23]V. Vellody, M. Nassery, W. Druz, and J. Sharp, "Effects of body position change on thoracoabdominal motion," *Journal of applied physiology,* vol. 45, pp. 581-589, 1978.

[24]V. F. Parreira, C. J. Bueno, D. C. França, D. S. R. Vieira, D. R. Pereira, and R. R. Britto, "Breathing pattern and thoracoabdominal motion in healthy individuals: influence of age and sex," *Revista Brasileira de Fisioterapia,* vol. 14, pp. 411-416, 2010.