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Comparative Analysis of Phase Difference Estimation Methods Quantifying Asynchronies between Compartmental Chest Wall Volume Signals

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Abstract— **Asynchronous breathing movements may be observed in the presence of pulmonary disease, such as chronic obstructive pulmonary disease (COPD). This study was undertaken in an attempt to propose a reliable methodology to quantify this asynchrony. Five methods for estimating phase differences between two signals, based on the phase angle of the Fourier Transform (PhD_{FT}), paradoxical motion (PhD_{PM}), the Lissajous figure (PhD_{LF}), maximal linear correlation (PhD_{P}) and least-squares filtering (PhD_{LS}), were compared. Frequency-modulated signals, simulating compartmental chest wall volumes, were used to evaluate the methods. Breathing asynchrony was quantified in two ways; by estimating (a) a single PhD value for the entire recording and (b) time-varying PhDs, representing non-stationarities of human breathing. PhD_{PM} and PhD_{LF} had the lowest average errors (4%), and PhD_{LS} had a slightly higher error. PhD_{FT} had zero error when estimating a single PhD value but a considerable error when estimating time-varying PhDs. PhD_{P} presented the highest errors in all cases. An application of this methodology is proposed in real compartmental chest wall volume signals of normal and COPD subjects. Preliminary results indicate that the methodology is promising in quantifying differences in asynchronous breathing between thoracic volumes of COPD patients and healthy controls.**

I. Introduction

In normal breathing, chest wall compartments, including the rib cage and the abdomen, move in a synchronized fashion during inspiration and expiration. In the presence of pulmonary disease, such as Chronic Obstructive Pulmonary Disease (COPD), asynchronous breathing movements may be observed.

COPD is the fifth leading cause of death worldwide and further increases in its prevalence and mortality are expected in the coming decades [1]. According to the Global Initiative for Chronic Obstructive Lung Disease (GOLD) "COPD is a disease state characterized by airflow limitation that is not fully reversible. The airflow limitation is usually both progressive and associated with an abnormal inflammatory response of the lungs to noxious particles or gases" [2]. Breathing patterns in COPD patients are different than those in normal subjects. Such patterns are characterized by increased respiratory rates, reduced inspiratory times and elevated mean inspiratory flows. In addition to this, asynchrony between rib cage and abdominal movements has been reported in some COPD patients.

A form of discoordination of the respiratory muscles of the upper and lower (abdominal) parts of the rib cage, also known as Hoover's sign, was described as far as 1920. Asynchronous breathing movements in COPD patients were first described in 1975 [3]. In 13 of the 17 COPD patients studied, the abdomen moved inward suddenly near or at end inspiration and then outward during a variable part of expiration. In more recent studies [4], [5], [6], similar phenomena were described for somewhat larger patient populations. According to these studies, phase differences (PhDs) between rib cage and abdominal chest wall volume signals are usually less than 10^0 in normal subjects, but can reach 30^0 in COPD patients. It is pointed out that phase differences between chest wall compartments have also been described in infants after elective herniorapy [7].

Reliable estimation of the PhDs between two respiratory signals requires the use of (a) an appropriate recording system to acquire the investigated signals and (b) an appropriate signal processing methodology. Previous related studies used magnetometers [5], inductive plethysmography [7] and, more recently, opto-electronic plethysmography (OEP) [6], to acquire chest wall volume signals. A number of conventional methods for estimation of phase differences were used in most of the previous studies, including maximal linear correlation (Pearson method) [7], loop analysis [6] and paradoxical motion [6]. Motto et al [7] suggested a novel method based on least-squares filtering, which was demonstrated to perform better than the Pearson method.

The purpose of this work was to investigate PhDs between chest wall volume signals, in an attempt to extend and complement findings of previous related studies. To this end, the performance of five different PhD estimation methods was compared in synthetic signals simulating chest wall volumes. The study was further extended to real signals acquired using OEP.

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II. Materials and Methods

A. Simulated Signals

To estimate the performance of the investigated Phd methods, simulated signals were created inspired by the methodology described in [7] and using MatLab 7.5.0 (The MathWorks, Inc, Natick, MA). Two signals, $s_1(t)$ and $s_2(t)$, corresponding to two different chest wall volumes, were modeled as frequency-modulated sinusoidal signals. Their mathematical expressions are

$$s_1(t) = 1.0 \cdot \cos(\pi \cdot \mu(t) \cdot t^2 + 2 \cdot \pi \cdot \sigma(t) \cdot t)$$

$$s_2(t) = 0.5 \cdot \cos(\pi \cdot \mu(t) \cdot t^2 + 2 \cdot \pi \cdot \sigma(t) \cdot t + \frac{\pi}{4})$$

where

$$[\sigma(t), \mu(t)] = \begin{cases} [0.8,0], \text{if } t \le 80\,\text{sec} \\ [-0.04,4.0], \text{if } 80 \le t \le 90\,\text{sec} \\ [0.06,-5.0], \text{if } 90 \le t \le 100\,\text{sec} \\ [0,1.0], \text{if } t \ge 100\,\text{sec} \end{cases}$$

Noisy versions of the above signals were also created to assess the performance of the interrogated methods in the presence of noise. Two types of additive noise were produced, namely random and Gaussian. Random noise, produced using MatLab's 'rand' command, consists of random numbers between 0 and 1, normalized so as to be within ±30% of the signal. Gaussian noise, produced using MatLab's 'randn' command, consists of numbers following the normal distribution with mean 1 and standard deviation 0, again normalized so as to be within ±30% of the signal. Fig. 1 shows examples of the simulated signals.

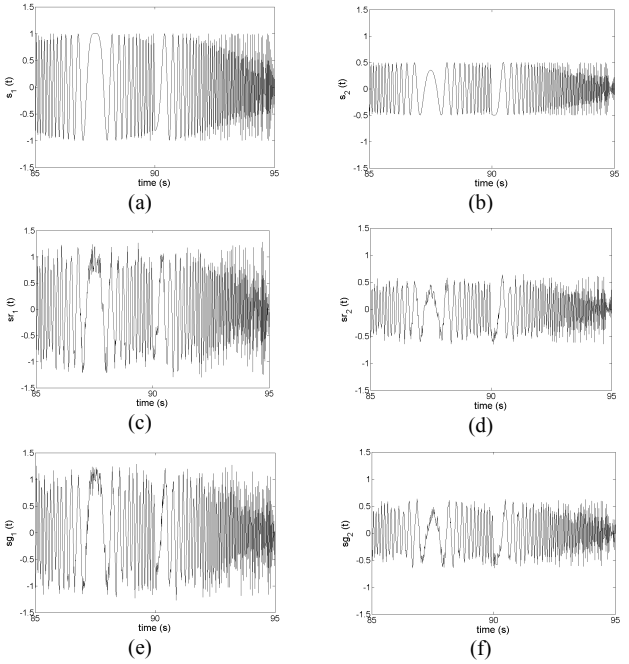


Fig. 1. Examples of simulated frequency-modulated signals without noise (a, b), with random noise (c, d) and with Gaussian noise (e, f).

B. Methods for Estimating Phase Differences between two Signals

In this study, five different methods for PhD estimation were assessed, based on the phase angle of the Fourier Transform (PhD_{FT}), paradoxical motion (PhD_{PM}), the Lissajous figure (PhD_{LF}), maximal linear correlation or Pearson method (PhD_P) and least-squares filtering (PhD_{LS}). Of these, PhD_P is the most commonly used method for estimating PhD, which was shown to have the best performance among a number of methods measuring thoracoabdominal asynchrony [8]. PhD_{LF} and PhD_{PM} have been widely used in estimating asynchronies of respiratory movements. PhD_{FT} is an easily implemented method, not previously used in respiratory signals. Finally, PhD_{LS} was recently suggested and demonstrated to perform better than PhD_P. In the following, the main principles of the five methods are briefly described.

1) Phase Difference based on the Phase Angle of the Fourier Transform (PhD_{FT}): The accuracy of this method depends on the accurate estimation of the signals' frequency. After calculating the signals' main frequency using the Fourier Transform (FT), the phase angles of the FTs of the two signals are estimated. The values of the phase angles corresponding to the signals' main frequency are detected and their difference, which represents the PhD of the two signals, is calculated.

2) Phase Difference based on Paradoxical Motion (PhD_{PM}): Paradoxical motion may be defined as the ratio of time during which the two signals move in opposite directions (i.e., one is increasing while the other is decreasing) over the total signal duration. If two signals are in phase, their paradoxical motion is 0 (or 0%). If two signals are completely out of phase, their paradoxical motion is 1 (or 100%). In all other cases, paradoxical motion has values between 0 and 1.

3) Phase Difference based on the Lissajous Figure (PhD_{LF}): The Lissajous Figure, or loop analysis, is often applied for a single period of the investigated signals. It is produced by plotting one signal against the other. The PhD may be determined by measuring the total left-to-right excursion of the loop and the width of the loop at its midpoint [8].

4) Phase Difference based on Maximal Linear Correlation, or Pearson Method (PhD_P): This method involves determining the phase shift at which two signals are maximally correlated. The correlation coefficient (r) of the two signals is initially estimated. The signals are then shifted relative to one another and the new r is determined. The relative shift at which r is maximal is taken as the PhD.

5) Phase Difference based on Least-Squares Filtering (PhD_{LS}): This method was suggested by Motto et al and is described in [7]. Briefly, the two signals are first filtered using a linear-phase finite-impulse-response (FIR) bandpass filter, then converted to binary, introduced in an exclusive-OR gate and FIR-filtered again.

Using the previously described methods, breathing asynchrony was quantified in two ways; by estimating (a) a single PhD value for the entire recording and (b) time-

varying PhDs, representing non-stationarities of human breathing. In case (a), each method is applied once to the entire signal. In case (b), each method is applied within a window of the signal which is then slided along the signal. In the case of Lissajous figure, the PhD_{LF} is calculated by the characteristics of the loop formed by the data included in the window. The size of the window should be carefully selected. In this study, three window sizes were used, namely 100 samples, corresponding to duration lower than a breath, 250 samples, corresponding to approximately the duration of one breath and 400 samples, corresponding to duration somewhat larger than one breath.

C. Real Chest Wall Volume Signals

Real compartmental chest wall volume signals were acquired using OEP, according to the protocols described in [9] and [10]. The sampling frequency was 60 Hz. In this study, data from seven (7) normal subjects and ten (10) COPD patients were used. All subjects performed an incremental exercise test on a cycle ergometer. The test consisted of measurements during 4 (3 for COPD subjects) min of quiet breathing (QB), followed by 4 (3 for COPD subjects) min of unloaded pedaling and a ramp increase of load to the limit of tolerance (W). Three chest wall volumes were recorded for each exercise stage, namely those of the pulmonary rib cage (Vrcp), the abdominal rib cage (Vrca) and the abdomen (Vab).

PhDs were estimated (a) between Vrcp and Vrca and (b) between Vrcp and Vab. Differences between normal and COPD subjects were assessed using the Wilcoxon statistical test. A p-value of 0.05 was considered significant.

III. Results

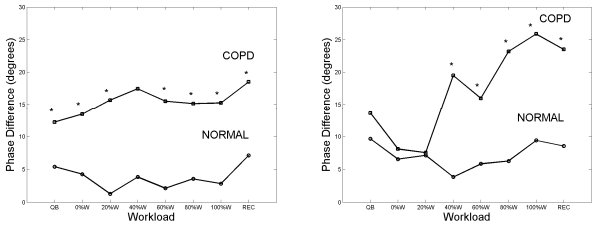
Table I shows error values produced by each interrogated method for the three noise versions of the simulated signals in the case of estimating a single PhD value, as described in section IIB. As we can see, in the absence of noise, PhD_{FT}, PhD_{PM} and PhD_{LF} have zero errors, whereas PhD_P is very inaccurate. In the presence of noise, either random or Gaussian, PhD_{FT} is the only method still producing no error. The other four methods have considerably high errors, rendering them rather inappropriate for the analysis of signals with relatively increased noise levels.

TABLE I
ABSOLUTE ERRORS FOR EACH OF THE FIVE INVESTIGATED PHASE DIFFERENCE ESTIMATION METHODS AND FOR DIFFERENT NOISE TYPES WHEN ESTIMATING A SINGLE PHD VALUE

	PhD_{FT}	PhD_{PM}	PhD_{LF}	PhD_P	PhD_{LS}
Noise-free	0.00π	0.00π	0.00π	0.24π	0.02π
Random	0.00π	0.17π	0.11π	0.24π	0.17π
Gaussian	0.00π	0.17π	0.06π	0.41π	0.26π

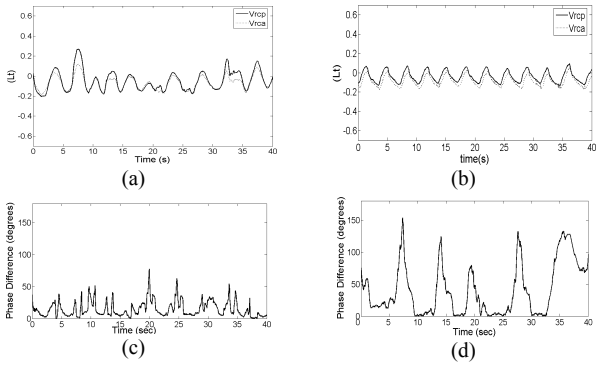
Table II shows mean (±std) error values in the case of estimating time-varying PhDs for three different window sizes. As we can see, PhD_{PM} has the lowest errors, followed by PhD_{LF} and PhD_{LS}. PhD_{FT}, although had no errors when estimating a single PhD value, it has considerable errors in the estimation of time-varying PhDs. PhD_P still presents with maximal errors.

Fig. 2 shows PhD values estimated for real data. These values were calculated using PhD_{FT}, which was shown to have the best performance (see Table I). Fig. 3 shows two examples of time-varying PhDs for a normal and a COPD subject during quiet breathing. These were estimated using a window size of 250 samples and PhD_{PM}, which was shown to have the best relative performance for estimating time-varying PhDs (see Table II). It is pointed out that the corresponding single PhD value in the case of the normal subject was 0^0, whereas that of the COPD subject was 27^0.



(a) (b)

Fig. 2. PhD values between (a) Vrcp and Vrca and (b) Vrcp and Vab, for normal and COPD subjects for different exercise levels. QB: quiet breathing, W: maximal workload, REC: recovery. *: p-value<0.05



(a) (b) (c) (d)

Fig. 3. Examples of time varying PhDs for a normal (c) and a COPD subject (d). (a) and (b) show the corresponding original Vrcp and Vrca signals from which PhD was calculated. The time-varying PhDs in (c) and (d) were estimated using PhD_{PM} and a window of 250 samples. Only a small part of the signals is presented so as to enable visual inspection of the underlying PhD.

IV. Discussion

In this paper, a comparative study between five PhD estimation methods was undertaken in an attempt to suggest a reliable methodology to quantify breathing asynchronies often observed during breathing in COPD patients. Compared to previous related studies, the novelty of this work consists in (a) the use of a PhD estimation method based on the phase angles of the FT, which has not been previously used in chest wall volume signals and (b) the estimation of time-varying PhD, representing non-stationarities of human breathing.

The comparative analysis of the five methods was based on simulated signals especially created for this work. Simulated signals were frequency-modulated so as to approximate real chest wall volume signals. In reality, chest wall volume signals are also amplitude-modulated. However, this type of modulation was not used because PhDs are not affected by the amplitude modulation of the signals.

Real chest wall volume signals, acquired using appropriate measuring devices, are usually corrupted by low- and high-frequency noise. In the simulated signals in this study high-frequency noise was added, usually due to the sensors and electronic parts of the measuring devices. Low-frequency noise was not included, but the real signals were 'detrended', i.e. a linear trend was removed, in an attempt to remove low-frequency components.

The results of the comparative analysis demonstrated that different methods should be used when estimating a single PhD value for the entire recording and time-varying PhDs. In particular, PhD_{FT} had zero error when estimating a single PhD value but a considerable one when estimating time-varying PhDs. This may be due to the fact that the accuracy of this method relies on the accurate estimation of the signals' frequency, which is compromised when a small window is used.

The findings from the application to real signals showed that PhDs in normal subjects were significantly lower than those in COPD patients. This seems to be in agreement with previous studies [3], [4], [5], [6].

V. Conclusion

The work described in this paper is an initial attempt to select a reliable, automated methodology, among existing techniques, to quantify asynchronous phenomena from respiratory signals. Comparative analysis of five PhD estimation methods showed that the methods based on the phase angles of the FT (PhD_{FT}) and paradoxical motion (PhD_{PM}) were optimal for determining a single PhD value and time-varying PhDs, respectively. Further in-depth investigation of computational issues as well as application of the suggested methodologies to large subject population are required to corroborate the findings and determine novel physiological markers for the study of the mechanics of breathing in COPD.

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TABLE II

Average (Std) of the absolute of Errors for Each of the Five Interrogated Phase Difference Estimation Methods and for Different Noise Levels when Estimating Time-Varying PhDs. L: window size in samples

		PhD_{FT}	PhD_{PM}	PhD_{LF}	PhD_{P}	PhD_{LS}
L=100						
	Noise-free	0.28π (0.64π)	0.01π (0.02π)	0.01π (0.04π)	0.29π (0.29π)	0.04π (0.06π)
	Random	0.29π (0.64π)	0.04π (0.06π)	0.07π (0.02π)	0.30π (0.29π)	0.25π (0.24π)
	Gaussian	0.30π (0.64π)	0.04π (0.05π)	0.06π (0.03π)	0.29π (0.29π)	0.21π (0.21π)
L=250						
	Noise-free	0.25π (0.61π)	0.00π (0.01π)	0.01π (0.04π)	0.31π (0.30π)	0.02π (0.04π)
	Random	0.26π (0.62π)	0.03π (0.05π)	0.07π (0.02π)	0.30π (0.30π)	0.20π (0.17π)
	Gaussian	0.27π (0.63π)	0.03π (0.05π)	0.08π (0.03π)	0.31π (0.29π)	0.26π (0.16π)
L=400						
	Noise-free	0.26π (0.62π)	0.00π (0.01π)	0.01π (0.04π)	0.39π (0.34π)	0.02π (0.03π)
	Random	0.27π (0.63π)	0.02π (0.05π)	0.07π (0.03π)	0.38π (0.35π)	0.28π (0.19π)
	Gaussian	0.27π (0.63π)	0.03π (0.05π)	0.06π (0.02π)	0.38π (0.35π)	0.21π (0.12π)