Instrumentation for respiratory flow estimation using tracheal sounds analysis: Design and evaluation in measurements of respiratory cycle periods and airflow amplitude

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Abstract. The increase in the incidence of respiratory diseases resulted in an increased interest in the improvement of the instruments used in research and diagnosis of respiratory dysfunction. We developed a non-invasive instrument able to estimate ventilation parameters by means of tracheal sounds analysis (TSA). The development of the hardware and software used in the instrument is described, as well as initial in vivo results obtained in a normal subject. These tests showed that the TSA was able to provide adequate airflow estimation. Moreover, it was found that accurate respiratory periods can be obtained from tracheal sounds. These results were in close agreement with the physiology, confirming the high scientific and clinical potential of this system.

1. Introduction
There is a consensus in the scientific community that the costs associated with respiratory diseases are increasing and are likely to grow exponentially in the next generations [1]. This resulted in increased interest in the improvement of the methods used in research and diagnosis of respiratory dysfunction.

The Phonospirometry appears to be a very promising method in this regard. This method is based on the estimate of the ventilation parameters by means of measuring breathing sounds in the trachea [2]. From the patient's point of view, obtaining tracheal sounds signal is relatively easy, since the signal is taken non-invasively. These sounds are generated by turbulence mainly within the large and medium airways, being influenced by the flow rate and the dimensions of the airways. As a result, for a given position of the subject and the microphone, the sound amplitude is proportional to the flow velocity, which suggests the possibility of obtaining flow estimates from these measurements. This is currently a topic of great interest because this method can provide a simple alternative to respiratory analysis [3-5]. The areas of interest are the use of tracheal sounds as indicators in the upper airway obstruction and as a source for the qualitative and quantitative evaluation of airflow. In this context, the tracheal sounds analysis (TSA) was applied in measuring the periods related to the different stages of the respiratory cycle [6], to monitor respiration [7], as well as in the detection of the onset of respiratory cycles of inhalation and exhalation phases [8]. Previous studies have reported a connection between the signal in

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the trachea, the underlying geometry of the airways and air flow [9]. Based on this principle, Yonemaru et al. [10] have developed a new method for the detection of stenosis in the trachea, while Shirazi et al., [4] used this principle to detect abnormal swallowing involving silent aspiration. In sleep medicine, intensity changes of tracheal sounds were used in the identification of apnea events [11]. Kraman et al. pointed out that although the TSA is promising as a simple and non-invasive method of assessing airway, more studies are needed to correlate specific anatomical features with the sound characteristics [12]. In this context, this paper describes the design and performance analysis of an instrument capable of evaluating respiratory parameters through TSA.

2. Materials and Methods

2.1. Hardware design

Figure 1 describes a simplified block diagram of the instrument. Tracheal sounds were measured using a Sony microphone (ECM-77B). The balanced outputs are initially processed using an Instrumentation Amplifier (INA118, Texas Instruments Incorporated, Texas, USA) and then analogically high-pass filtered (0.05 Hz, first order) to remove off-set and low-pass filtered (Butterworth, 8th order, 5000 Hz) to reduce external noise. The resulting signal was measured at a sampling rate (fs) of 40 kHz [13] by a data acquisition module (NI 6003, National Instruments, Austin, Texas, USA), which presents a resolution of 16 bits, eight channels and fs maximum of 100 kHz. To simplify the practical use of the system, the hardware platform used was a touch screen notebook (Dell Inspiron series 137000, Intel, Core i5, 4G of RAM, hard disk of 456 GB).

The airflow obtained by TSA was compared with a reference airflow measurement system (Figure 1). It is composed by a fleisch pneumotachometer (PNT) coupled to a differential pressure transducer (176PC, Honeywell Inc. USA), whose signal is amplified by an instrumentation amplifier (INA118, Texas Instruments Incorporated, Texas, USA) and subsequently processed by a low-pass filter (Butterworth, 4th order, 10 Hz) and adapted to the data acquisition system.

2.2. Software design

The software was developed in LabVIEW 2012 environment (National Instruments, Austin, TX). It is composed by two main modules; the first contains the data acquisition subroutines through the USB port, allowing the visualization of PNT and tracheal sounds signals and a visual quality control analysis. At the end of each exam the system allows researchers to save the data in an ASCII file. The second module performs the tracheal sounds signal processing. A simplified flowchart describing the dedicated software and typical signals obtained in the signal processing chain are presented in Figure 2A. It begins with a band-pass filtering (300 Hz to 1 kHz, Butterworth, 5th order) to minimize cardiac effects and to remove noise from the external environment. The signal is then manually segmented to start before the beginning of an inspiration (Figure 2B graphs 1 and 2). The automatic identification of the beginning and end of inspiration and expiration, which allows the estimation of the inspiratory, expiratory and respiratory time, was obtained using the minimum values of Shannon
Entropy [14]. This parameter was calculated at intervals of 20 ms (800 samples) and the resulting signal was low-pass filtered (1 Hz, first order, red points in graph 2B graph 4). As can be seen in Figure 2B graph 3, the respiratory signal obtained from the tracheal sounds is rectified containing only positive values. To perform the reconstruction of the respiratory signal, the segment of the tracheal sounds signal associated with the expiratory phase was multiplied by -1. A typical result of this process is described in Figure 2B graph 5. The final results may be saved in an ASCII file for further clinical/research use. The software also includes the analysis of the spectrogram (or sonogram) of the tracheal sounds.

2.3. In vivo tests in healthy individuals. The ability of the system to identify the duration of the respiratory cycle phases and airflow amplitude was evaluated through a comparative analysis with the results obtained in the reference system (PNT). It was performed analyzing 5 young healthy individuals that do not present history of respiratory diseases or tobacco use (3 male, 24.6 years, 172 cm, 68.7 kg). During the exams, the individuals remained seated, with the head in a neutral position.

2.4. Statistical analysis. The comparison between the results was performed by correlation analysis using ORIGIN 8.0. Values of p<0.05 were considered significant. The RMS error was also calculated.

3. Results

3.1 System development

Figure 3 illustrates the developed system, its typical use and the front panel of the program used in data acquisition. The supporting elements used during the exams, including the biological filter and the microphone are also described.

3.2. System evaluation

3.2.1 Respiratory cycle time in normal individuals. Figure 4 describes the comparative analysis of the PNT and TSA results. It were observed RMS errors of 0.19 s, 0.27 s and 0.03 s in the estimation of expiratory, inspiratory and total respiratory cycle time, respectively.

Figure 2: Simplified flowchart describing the dedicated software (A) and typical signals obtained in the signal processing (B).
Figure 3: Picture describing the system and its use (A) and frontal panel of the data acquisition module subsystem (B). MIC = microphone; BF = bacterial filter.

Figure 4: Associations of inspiratory (A), expiratory (B) and respiratory cycle (C) times measured using the pneumotachometer and tracheal sound analysis in five normal subjects during spontaneous ventilation.

3.2.2 Airflow amplitude in normal individuals. Figure 5A shows a typical result observed in the studied individuals. The correlation coefficient between the PNT and TSA was 0.95±0.02 (range 0.93-0.98) and significant (p<0.0001) in all studied individuals. Figure 5B present a typical spectrogram obtained in a segment of 20 s of tracheal sounds obtained during spontaneous ventilation.

Figure 5: Typical correlation of the airflow measured using the pneumotachometer and tracheal sound analysis observed during spontaneous ventilation (A). Tracheal sound signal spectrogram observed during spontaneous ventilation (B; low) along with the tracheal sound signal (B; middle) and the corresponding typical flow (B; top graph).

4 Discussion

In vivo tests. The morphology of the tracheal sounds signal described in Figure 2B1 is coherent with that presented in a recent study of Elwali and Moussavi [15] and in the studies of Sierra et al. [16] and Que et al. [2], in which the tracheal sounds morphology of normal awake subjects were investigated. In Figure 2B we can also observe that the estimation of airflow from tracheal breath sounds entropy resulted in a signal that follows the airflow variation very closely. These results confirm previous findings [14]. They are also consistent with the study of Yu et al [17] in which the tracheal breath sounds...
entropy was used to detect abnormal reductions in airflow during sleep. Both these earlier results and ours confirmed the high potential of the TSA in physiological and clinical studies. 

**Respiratory cycle time in normal individuals.** Sohrabi et al. [18] recently suggested that TSA was adequate to track the different phases of the breathing cycle. The results presented in Figure 4 clearly provide additional support to this hypothesis. The strong linear correlations observed in Figure 4 are also consistent with previous works investigating normal awake subjects [16]. The observed RMS errors were small, and probably associated with the presence of laminar flow and reduced breathing sounds at low airflow values. Comparing with the mean values of the inspiratory, expiratory and total respiratory cycle periods, these errors were < 18%, < 13% and < 1%, respectively. These values represent a reasonable accuracy, and may be considered appropriate to clinical applications. However, it is important to point out that some measurements were observed far from the regression line (Figure 4: one in the inspiratory, one in the expiratory and two in the respiratory cycle times). The precise definition of the nature of these errors is a work in progress.

**Airflow amplitude in normal individuals.** It was observed that the physiological tracheal sounds signal was absent during low airflow in the respiratory cycle (Figure 5A, near ±0.5V of the PNT output). This is consistent with the study developed by Sohrabi et al. [18], and can be explained considering that below this level there are no breathing sounds due to the presence of laminar flow. The very high correlations observed between the actual measured flow and the acoustically determined flow was also consistent with the cited work. The obtained results corroborate the hypothesis that the band pass filtered tracheal sounds entropy provides adequate estimation of airflow [14] and provides further experimental evidence of the relevance of TSA ability to provide effective noninvasive airflow quantitative measurements.

The spectrogram (Figure 5B, bottom) describes the amplitude spectrum for each time segment of the tracheal sounds signal (Figure 5B, middle). In the reference respiratory signal (Figure 5B, top) the positive values refer to inspiration and the negative values refer to expiration airflow. The spectrogram showed that, in the studied subject, the tracheal sound energy is concentrated in frequencies below 700 Hz and that the energy drops off between 700 and 1500 Hz. It was also observed that there is not much difference comparing the higher frequency components in the inspiratory and expiratory segments of the tracheal sounds. Another interesting finding in the spectrogram is the presence of a silent period separating the inspiratory and expiratory segments, which is consistent with the tracheal sound waveform (Figure 5B, middle). These results give additional support to the behaviour observed recently by Bersain et al. [19]. In agreement with the mentioned work, and with the physiological phenomena, this finding clearly describes the cessation of turbulence. This is characteristic of the of low airflow velocities events in the beginning of inspiration and expiration.

### 5 Conclusion & future work

A novel system for the acquisition and processing of tracheal sounds signals was developed in the current study. The ability of the system to evaluate respiratory cycle time and amplitude was validated by a study in normal subjects. The proposed system may be useful in a wide spectrum of clinical and research studies of subjects with respiratory diseases, enabling simple and noninvasive exams and improving the assistance offered to these patients.

Based on these promising results, future plans include the development of a portable version to help home monitoring and telemedicine applications [20]. They also include studies in well-defined groups of patients with respiratory diseases in order to evaluate the clinical contribution of the proposed instrument, and to integrate the TSA instrumentation described in the present work with machine learning algorithms [21], contributing to improve the diagnostic, understanding and management of respiratory diseases.

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