Biomedical Signal Processing and Control 36 (2017) 57–62

Contents lists available at ScienceDirect

Biomedical Signal Processing and Control

journal homepage: www.elsevier.com/locate/bspc

Short communication

Monitoring respiratory impedance by wearable sensor device: Protocol and methodology

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ARTICLE INFO

Article history:
Received 10 November 2016
Received in revised form 9 March 2017
Accepted 29 March 2017

Keywords:
Respiratory impedance
Mobile monitoring
Healthcare
Protocol
Forced oscillation technique
Online identification
Frequency response
Closed loop control
Wearable sensor
Respiratory disease
Cyber-physical systems
Fractional order impedance

ABSTRACT

This paper presents an original protocol, method and deploying solution for monitoring respiratory impedance in mobile individuals. The purpose of the follow up study may be either short-term, or long-term. The proposed steps take into account effects of exercise, rest and sleep on the mechanical properties of the respiratory signals and their computed impedance as a function of time and frequency. The end-product is a wearable sensor, which may be linked in a network of sensors, cyber-physical systems containing mobile devices, central data logging archives and possible alarms to medical supervisor. The added value is that given a treatment profile, its efficacy can be measured by evaluation of respiratory impedance on a daily (regular) basis, without the necessity for the patient to come to healthcare units provided with lung function test facilities. The visit to specialist can be then decided based on the follow-up online by the responsible person and treatment adaptation can be performed. Reduction of visits is thus an important factor in reducing healthcare costs and improving patient wellbeing while delivering highly qualitative services.

1. Introduction

In today's healthcare framework, monitoring vital signs on-the-go is an efficient and pragmatic way to maintain contact between patient and general practitioner (GP). Enabled by mobile technologies such as mobile apps in smartphones, a variety of health monitoring devices and gadgets provide a revolutionary lifestyle in modern society.

A well-established problem of modern society and urban development is the presence of alarming smog and allergens levels, which play a triggering role in asthma and other obstructive diseases in the airways. An equally important societal challenge is ageing of the population and limited hospital facilities, with prolonged waiting times for scheduling consultation visits. This paper addresses this problem by offering a solution for a wearable sensor monitoring device to establish at regular times the respiratory condition of the patient under observation and contact the GP whenever necessary. The data logging system may be linked through mobile services to the personal patient file in a local server and made available to the GP through the cloud. If necessary, alarm procedures may notify the patient to take action (e.g. asthma patient to inhale specific medication) and connect to the GP's mobile app to establish a patient visit, or, in more critical cases, a connection to the emergency services (e.g. asthma attack).

For instance, in patients undergoing specific medication (e.g. asthma or chronic obstructive pulmonary disease), monitoring the respiratory impedance at regularly defined times can give insight into the evolution of the medical treatment and allow follow up. This may be more beneficial and cost effective than intermittent visits to the GP or the medical specialist. Events which may reveal symptomatic problems occurring at sporadic times may be detected and acted upon in effective time to diminish risk for the patient and increase specificity and effectiveness of the treatment. A conceptual visualization of the cyber-physical system is given in Fig. 1.

To enable wearable lung function devices, the most crucial property of the protocol is patient autonomy. The only lung function test available in clinical practice which does not require patient cooperation in special manoeuvres is the forced oscillation technique (FOT) [1–3]. Surprisingly, hitherto this test has not been

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http://dx.doi.org/10.1016/j.bspc.2017.03.018
1746-8094/

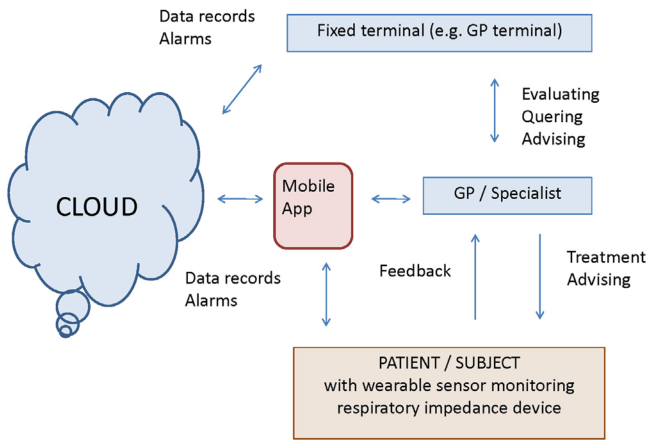

Fig. 1. A conceptual overview of the cyber-physical system with integrated wearable sensor FOT device.

standardized, unlike its more established tests such as spirometry and body plethysmography [4]. The main reason is a lack of acceptance from clinical practice of its added value, despite repeatedly efforts to show its benefits [5–7].

The paper here exploits the advantage of FOT and a monitoring device for respiratory impedance is presented along with its measurement protocol and methodology.

The paper is organized as follows. A brief summary on FOT and proposed methodology for online estimating the impedance is given in the second section, along with the protocol for applying it. Section 3 describes the prototype and its validation, followed by discussion of results in Section 4.

2. Methodology

2.1. Principles of FOT

FOT is a non-standardized lung function test based on the action-reaction principle applied to the lungs during normal breathing in a non-invasive manner [1]. The patient is advised to breath normally, i.e. no force manoeuvres are required, a non-condition which enlarges significantly the applicability scope to marginal groups such as infants, children and the elderly [8].

Developed half century ago [9,10], FOT implies the generation of a signal $U_g(t)$ (sinusoidal, or combination hereof) within 0.3 kPa peak to peak amplitude [1,2]. Signal generation has varied in the past decades from loudspeakers, mechanical pistons, to fan-based ventilators, depending on the envisaged range of frequencies and power to be applied to the patient's lungs [11–14]. For instance, loudspeaker based mechanisms are commonly used for frequencies well above the breathing frequency; 5–250 Hz, revealing properties of the proximal airways and useful for aerosol deposition studies [15]. Mechanical actuators and ventilators works at lower frequencies (<5 Hz) enabling information on lung tissue properties such as viscoelasticity [16–18].

A reactive force following the excitation signal $U_g(t)$ is opposed by the respiratory tract which gives the possibility to compute the resistance and compliance of the system as a whole [7,10]. Obviously, athletic and healthy persons will exhibit lower resistance and high compliance, implying a smoother movement of airflow with less pressure difference between the mouth and the pleura [2].

By contrast, patients with chronic obstructive pulmonary disease will exhibit high resistance, low compliance, implying that a higher pressure difference is necessary to obtain a similar airflow as a healthy person [5,19,20]. These mechanical effects of work of breathing evaluated as pressure volume curves are also visible in the values of the calculated respiratory impedance [3].

The breathing of the person performing the FOT test is regarded as noise in this context, and its biasing effect must be eliminated by signal processing techniques [21,22].

2.2. Proposed online respiratory impedance estimation

FOT devices record pressure $P(t)$ (kPa) and flow $Q(t)$ (L/s) signals measured at the mouth of the patient in the following relation:

$$P(s) = Z_r(s) * Q(s) + U_r(s) \tag{1}$$

where s denotes the Laplace operator, Z_r the respiratory impedance and U_r the breathing signal.

With these signals available at every sampling period (e.g. every 1 ms), non-parametric identification methods can be applied to extract the cross-spectral power density of the input and the output of the system. Since the output amplitude is dependent on the input amplitude, the input signal is also used to reference the magnitude of the respiratory impedance [23]. Given the input is of sinusoidal type ($A\sin(\omega t)$), the impedance is a frequency dependent complex variable evaluated as:

$$Z_r(j\omega) = \frac{S_{PU_g}(j\omega)}{S_{QU_g}(j\omega)} \tag{2}$$

where $S_{XY}(j\omega)$ denotes the cross-correlation spectra between the various input–output signals, $\omega = 2\pi f$ is the angular frequency and $j = \sqrt{-1}$, the result being a complex variable. The derivation of (2) from the measured signals is detailed in [21].

Notice that in this case two signals need to be recorded from the patient: air-pressure and air-flow. In order to reduce costs and weight of the device, air-flow measurement instrumentation may be eliminated (i.e. pneumotachograph). This is possible as proposed in [24]. The alternative measurement using only pressure signal measurement exploits the use of a calibration tube as a pre-requisite to the online monitoring procedure. In practice this is possible, since the patient needs to calibrate the system only once with a tube whose impedance is known and can be delivered together with the device. Briefly, the methodology can be described as follows:

- measure pressure with a sealed system (flow is zero) – delivers an impedance Z_p and
- measure pressure with a known load – delivers an impedance Z_q.

As described in [24], this provides enough necessary information to estimate the respiratory impedance of the subject in the absence of flow measurement. In that work, the authors have proven that it delivers the same impedance values as with the standard impedance estimation form from (2). The form of the impedance is then given by:

$$Z_r(j\omega) = \frac{Z_q(j\omega)(S_{PU_g}(j\omega)/S_{U_gU_g}(j\omega))}{1 - Z_p(j\omega)(S_{PU_g}(j\omega)/S_{U_gU_g}(j\omega))} \tag{3}$$

with Z_p and Z_q known, as obtained from the calibration procedure.

For the online estimation method, one needs to measure pressure at the mouth of the patient continuously and introduce a window interval for data processing. This introduces latency in the information on respiratory impedance properties evolution in time, but its value may be neglected since the process is in open loop evaluation. For instance, given an input signal generated at 2 Hz, the impedance may be evaluated every 5 s and plotted as a point in frequency (e.g. real and imaginary parts of Z_r evaluated at 2 Hz) as a function of time. Variability of the real and imaginary values of the impedance in time on a regularly basis (e.g. daily, at same hour