- ORIGINAL ARTICLE -

Practical Design of Flow Meter for Mechanical Ventilation Equipment

Diseño Práctico de Caudalímetro para Equipos de Ventilación Mecánica

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Abstract

This paper introduces a practical technique for the design of an instrument used in air flow measurement or flowmeter. This instrument is an essential component in the hospital medical ventilation equipment functioning, therefore, the parameters design presented in this article focus on this purpose. However, this instrument can be employed to any measurement scale. The technique is based on indirect flow measurement, using a sensor that converts the flow parameter into a differential pressure measurement. An electronic transducer allows the differential pressure values to be obtained as an electrical signal, which is then digitized and analyzed to obtain the original parameter. The experimental procedure presented in this paper utilizes a computational algorithm to perform the signal analysis; however, given the simplicity of the procedure, this could be adapted to any digital processing card or platform, to show the measurement obtained immediately. Preliminary analyses demonstrated instrument efficiency with sensitivity of 0.0681 liters per second (L/s). Accuracy evaluation showed an average measurement error lesser than 1.4%, with a standard deviation of 0.0612 and normal distribution over the set of test measurements.

Keywords: Air flow measurement, Medical instrumentation, Venturi tube, Medical ventilation equipment.

Resumen

En este artículo se presenta una técnica simple pero eficiente para el diseño de un instrumento utilizado en la medición de flujo de aire o caudalímetro. Este instrumento es de uso imprescindible en el funcionamiento de equipos de ventilación médica hospitalaria, por consiguiente, los parámetros de diseño presentados en este artículo se enfocan para este fin. Sin embargo, este desarrollo se puede aplicar a cualquier escala de medición. La técnica se basa en la medición indirecta del caudal, utilizando un sensor que convierte el parámetro de flujo en una medición de presión diferencial. Un transductor electrónico permite obtener los valores de presión diferencial como una señal eléctrica, que luego se digitaliza y se analiza para obtener el parámetro que originalmente se desea medir. El procedimiento experimental que se presenta utiliza un computador para realizar el análisis de la señal, pero se puede adaptar a cualquier tarjeta de procesamiento digital, con el propósito de mostrar la medición obtenida de forma inmediata. Las pruebas preliminares de funcionamiento demuestran la eficiencia del instrumento con una sensibilidad de 0.0681 L/s. La evaluación de exactitud presenta un error promedio en la medición inferior a 1.4 %, con desviación standard de 0.0612 y distribución normal sobre el conjunto de mediciones de prueba.

Palabras claves: Medición de flujo de aire, Instrumentación médica, Tubo Venturi, Equipo de ventilación médica.

1. Introduction

The respiratory system is essential to living systems. The gas exchange depends on this system, which allows the body to obtain oxygen from the air and eliminate the gases generated as waste by-products in cellular reactions [1]. The work of the respiratory system is carried out in a repetitive cycle of two different periods, inspiration and expiration. This cycle repeats 12-18 times per minute for an adult person under normal conditions [1] [2].

Some diseases, such as bronchial asthma, chronic obstructive pulmonary disease (COPD), pulmonary tuberculosis, lung cancer, pneumonia, and more recently severe pulmonary coronavirus disease 2019 (COVID-19), can adversely affect the functioning of the respiratory system [3] [4], necessitating use of mechanical ventilators or artificial respirators [5] [6].

A mechanical medical ventilator is an electromechanical instrument that pumps an oxygen-enriched air mixture to the patient, with the purpose of supplying natural pulmonary ventilation. It is used when the patient suffers from some pathology that seriously affects the functioning of the breathing. Such ventilators can supply air to the patient through special masks, which fit tightly to the face (non-invasive mechanical ventilation), or through a tube that reaches the trachea (invasive mechanical ventilation), either by entering the patient through the mouth (endotracheal or orotracheal tube), or entering the patient through the neck ("tracheostomy") [3] [4] [7]. Fig. 1 presents a simplified diagram of the relationship between the pressure, flow and volume of air supplied to the patient, in a breathing cycle, by a mechanical ventilation equipment for medical use.

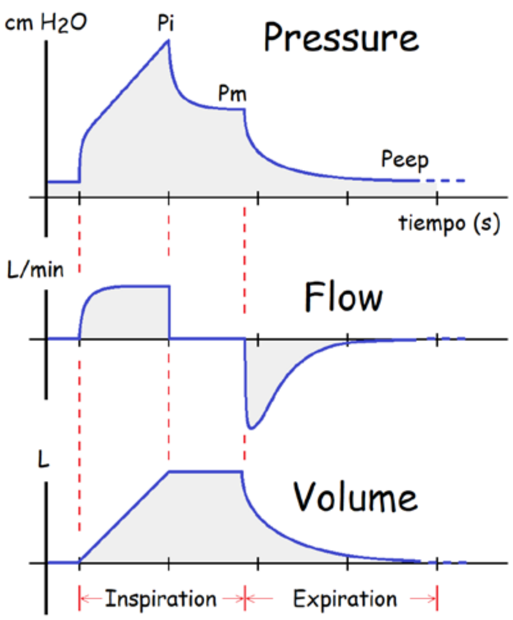

Figure 1. Relationship between pressure, flow and air volume in a mechanical ventilation cycle.

During the inspiration period, ventilators allow air into the lungs with an approximately constant flow rate of less than 2 L/s, without the insufflation pressure (Pi) or peak pressure exceeding an upper threshold of 60 centimeters of water (cmH2O). However, for safety reasons, it is recommended to work with Pi values below 40 cmH2O. The inspiration stage is completed with a pause period, where the plateau pressure (Pm) is maintained at approximately 30 cmH2O, and the flow becomes null. Pm is intended to make gas exchange in the lungs more efficient [7] [8]. In the expiration period, air is let out of the lungs until the cycle is complete, maintaining a positive pressure at the end (Peep), between 5 and 10 cmH2O, to avoid lung collapse [8].

The main parameter to be measured is the volume of air reaching the lungs, since gas exchange depends on this parameter. The required air volume is calculated between 5 and 12 milliliters per Kg of patient weight (mL/Kg), for example, for an 80 Kg patient 0.6 L of air could be supplied in each inspiration, as per the medical prescription [1] [4] [7] [8]. However, it is not easy to measure the volumetric displacement of the patient's rib cage. Therefore, the volume is normally obtained by the direct relationship with the measurement of the air flow, which passes through the inlet duct to the patient [7]. On the other hand, to achieve a constant air flow, as shown in Figure 1, the mechanical ventilation equipment must generate a pressure profile that is constantly adjusted over time. This could be achieved if the control system of the ventilator apparatus is fed back with the measurements of the air flow supply [9] [10].

Given the importance of the air flow measurement in instruments such as ventilators for medical use, this article presents the design and testing of a relatively simple technique in contrast to an instrument for measuring flow or flowmeter, such as piezoelectric plate of variable holes. The proposed technique is easy to manufacture by 3D printing. The instrument thus produced is washable and autoclavable. The current design has been made in accordance of the measurements used in medical mechanical ventilator equipment; however, it can be adjusted to other appropriate device requirements.

Currently there are a significant number of flow measurement devices on the market [11], but few sensors meet the requirements to be implemented in medical instrumentation. It is possible to find a range of specific equipment for medical use oriented to the measurement of fluids, but generally they do not allow control [12]. Some commercial flow sensors adopt characteristics identical to those presented in this article [13] [14], but they are relatively expensive (up to 100 times more expensive than the presented design) and of closed technology, which complicates their implementation in practical designs.

2. Materials and methods

The product presented is a low-cost instrument, which allows the air flow to be measured with high precision, designed for use in mechanical ventilation equipment for medical use. The diagram shown in Figure 2 presents the overall layout of the developed instrument. In general, the process involves hardware manufacturing stage and a software stage for the calibration and implementation of the product.

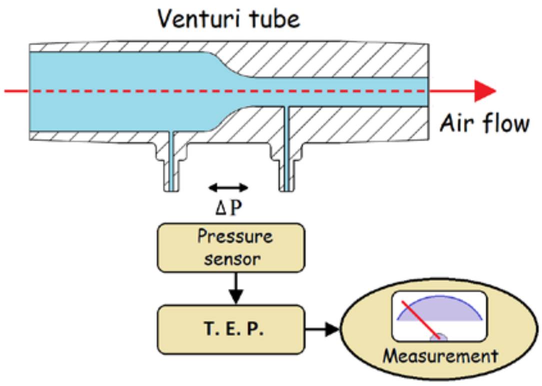

Figure 2. Structure of the air flow meter instrument.

The hardware stage consists of a Venturi tube [15] [16], a differential pressure sensor, and a programmable electronic card (TEP) for digitizing and handling the signal. The software step described in this document was designed to acquire and analyze the signal using a computer, but the procedure can be customized to display the appropriate measurements of medical/clinical interest directly [17] by the use of scanning card.

2.1. Sensor design

The air flow received by the patient is recorded by measuring the response delivered by a Venturi tube [15] [16]. This device, as shown in Figure 3, uses the pressure difference "h" generated when the fluid passes through a section of the duct through which it circulates.

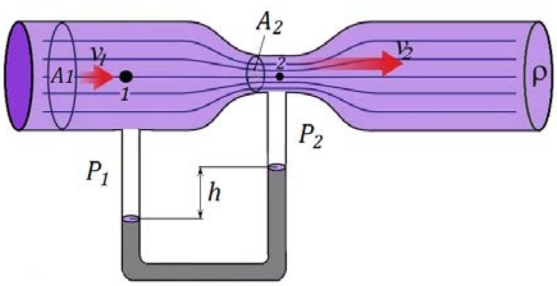

Figure 3. Descriptive diagram of the Venturi tube.

Bernoulli's principle [15] describes the behavior of a fluid that moves along a pipeline. Disregarding friction, Bernoulli's equation Eq. (1) defines that the narrowing in the Venturi tube, given by the section difference between "A1" and "A2", causes a change in the fluid velocity from "V1" to "V2 ". This results in the pressure "P2" being less than the pressure "P1". Therefore, the pressure differential "h" is an indicator proportional to the flow through the pipeline [15] [16].

$$P1 + \rho g y_1 + \frac{1}{2}\rho(V1)^2 = P2 + \rho g y_2 + \frac{1}{2}\rho(V2)^2 \quad \text{(Eq. 1)}$$

If the analysis of equation 1 is carried out taking reference points 1 and 2, indicated in Figure 3, it is observed that the potential energy factor, denoted by "ρgy", is nullified on both sides of the equation because the reference height "y" is the same. On the other hand, taking into account that the velocity of the fluid "V" is defined as the flow "Q" between the area "A" of the duct through which it circulates, equation 2 can be obtained from Eq. (2). The Eq. (2) allows calculating the A2 corresponding to the reduced section of the pipeline.

$$A2 = \sqrt{\frac{\rho\, A1^2\, Q^2}{2\,(P1 - P2)\, A1^2 + \rho\, Q^2}} \quad \text{(Eq. 2)}$$

The design of the dimensions of the Venturi tube starts from the measurement requirements and the physical parameters of the space it occupies. Fig. 4 shows the design of this sensor with the dimensions obtained.

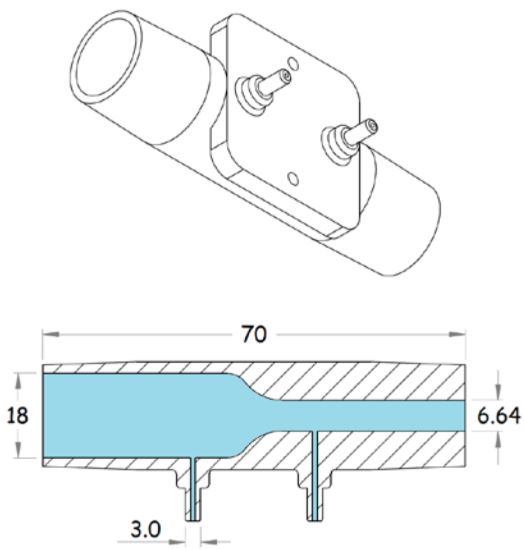

Figure 4. Isometric view and cross section with the dimensions of the designed Venturi tube.

The Venturi tube outlet is connected to a device that transforms the differential pressure into an electrical signal. For this purpose, the electronic component MPXV7002dp was used [18]. It is a positive and negative differential pressure sensor, which allows obtaining an analog output signal between 0.5 to 4.5 Volts (V) with linear response, proportional to the pressure differential in the range of -2 to 2 kilo Pascal (KPa) (equivalent to - 20.39 to 20.39 cmH2O). This device is temperature compensated between +10 and 60° C and has a maximum error of ± 6.25% on the measurements made.

The external diameter of the tube adapts to the dimensions of the standard hoses for medical ventilation equipment. Section A1 is obtained by setting the internal diameter to 18mm. This is done for the purpose of optimizing the strength of the tube walls. The internal diameter of 6.6444 mm is obtained as a result of calculation by applying Equation 2, taking into account that it is desired to obtain a pressure differential of 2 KPa when an air flow of 2 L/s circulates through the tube, given the air density ρ = 1,225 kg/m3. This design was manufactured using a 3D printer. Figure 5 shows the final product.

Figure 5. Finishing of the 3D printed Venturi tube.

2.2. Signal acquisition and processing

An Arduino UNO [19] programmable electronic card (PEC) was used, configured to digitize the signal received from the pressure sensor and transmit the data in real time via USB. The scanning speed was set at 100 samples per second, with a resolution of 10 bits per sample.

The measurement stage was carried out using a personal computer. The technique used by the Arduino system creates a virtual "COM" port (serial port with RS232 protocol) every time USB communication is connected between the Arduino hardware card and the computer. The Matlab programming environment was used to carry out the measurement process. This process is carried out in two stages: information acquisition and analysis of the acquired signal.

To perform the acquisition of the information, the algorithm developed in Matlab configures the active COM port for communication with the Arduino, reading the data that enters during a designated period of time and outputs as graphs in real time from the signal that is acquired. At the end of the acquisition, the data is saved on the hard disk, in a file with the extension ".csv" (Excel file where the data are separated by commas).

The information acquired represents the variations in pressure as a function of time, which originate as a consequence of the variations in air flow through the Venturi tube. Since the fundamental interest of the system is to measure the flow, a second algorithm was carried out to calculate the flow levels per unit of time. Fig. 6 shows the simplified flow diagram of the algorithm developed to analyze the acquired signal.

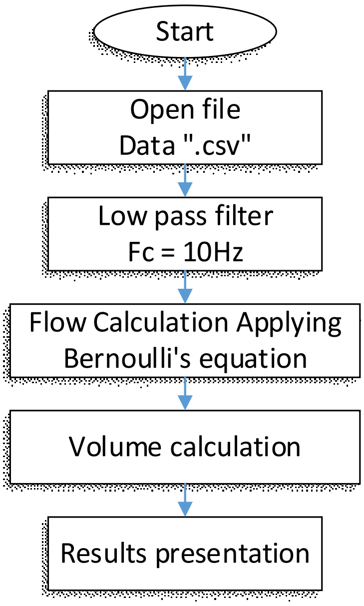

Figure 6. Flowchart of signal analyzer algorithm.

This algorithm: opens the file with the saved data after the acquisition, applies a low-pass filter to the signal with cut-off frequency of 10 Hz to reduce noise, calculates the flow values flowing through the tube and calculates the volume in time function of each respiratory cycle. Eq. (3) is applied for flow calculation, which is obtained identical to Eq. (2) by solving Bernoulli's equation.

$$Q = \sqrt{\frac{2\,(P1 - P2)}{\rho\left(\frac{1}{A2^2} - \frac{1}{A1^2}\right)}} \qquad \text{(Eq. 3)}$$

3. Results and discussions

To test and evaluate the performance of the developed instrument, an experimental procedure was designed by contrast. It consists of passing the total volume of air from a calibrated 3000cc (3L) syringe through the sensor of the designed instrument, while taking the corresponding measurement. The analysis of the obtained data allows to quantify the precision and the degree of accuracy in the measurements. Figure 7 shows the type of calibrated syringe that was used in the experiments.

Figure 7. Calibrated 3L syringe used in performance tests.

In a preliminary test, the signal is acquired by passing the total volume of 3L of air through the sensor in two cycles with different flow rates. To visualize the data processed with the analysis algorithm, the detected results are displayed in graphic form. In Fig. 8, the acquired differential pressure signal is presented in contrast to the calculated flow signal. Figure 9 shows how the volume value increases when integrating the flow signal in each cycle.

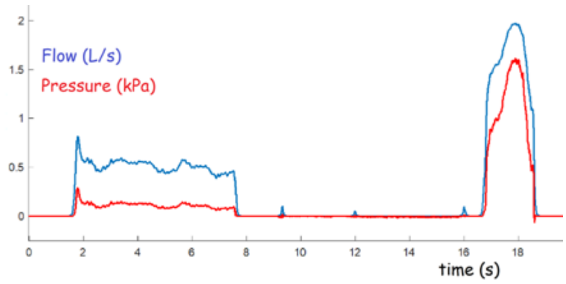

Figure 8. Flow signal calculated (blue) from the acquired differential pressure (red).

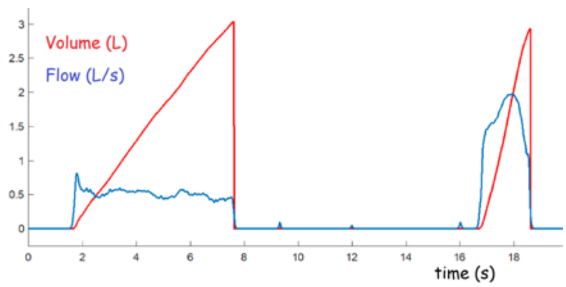

Figure 9. Partial volume (red) when integrating the flow signal (blue) in time function.

Measurements were made exclusively on positive flow (emptying the syringe), since medical ventilation instruments only require measuring the flow corresponding to the inspiratory period as in mechanical ventilation instruments (there are teams that have three flow meters: inspiratory, expiratory and common proximal to the patient). Therefore, all negative flow measurements (syringe refill) were discarded.

3.1. Measurement precision

The appreciation of the system's PET has a sensitivity of 1/1024 in the digitalization of the differential pressure signal. This appreciation is equivalent to a pressure value of 0.0019 KPa, which, on applying the Bernoulli equation, results in a sensitivity of 0.0681 L/s.

As we can be seen in the graphs, this level of sensitivity is almost imperceptible to the naked eye, but represents an error of 1.7% on the total measurement scale.

3.2. Assessment of product accuracy

To assess the accuracy of the instrument, the analysis of a group of measurements was performed using the same reference standard and the level of error was calculated. The evaluation was carried out by analyzing the measurements of 20 cycles of the total emptying of the air volume of the syringe. Each cycle of emptying the syringe was done by hand, with irregularly generated flow rates. We tried to make notable and progressive flow velocity variations between each acquisition file, with values between 0.3 and 2.17 L/s. The data obtained were recorded in Table 1.

Table 1. Data obtained in volume measurements.

File name	Acq. cycles	Total volume in each cycle (L)
P0_dif.csv	2	3.0392 2.9428
P1_7s.csv	2	3.0660 3.0669
P2_5s.csv	3	3.0201 2.9562 2.9450
P3_3s.csv	3	2.9935 2.9681 2.9838
P4_2s.csv	4	2.9741 2.8093 2.8758 2.9624
P5_1s5.csv	6	2.9338 2.9358 2.9162 2.9319 2.9486 2.9131

The absolute error is given directly as the difference between the measured value and the standard. The calculation of percentage error was performed applying Eq. (4), where Em = measured error, E = percentage error, and Standard = 3 L. The average error was obtained by averaging all the measurements.

$$E = \frac{Em * 100}{Patrón} \quad \text{(Eq. 4)}$$

The most relevant values that were obtained as a result of the analysis carried out were the following:

- Reference standard: 3L Syringe.

- Number of measurements: 20.
- Maximum measured volume = 3.0669 L → Em=0.0669 L → E=2.2300%
- Minimum measured volume = 2.8093 L → Em=-0.1907 L → E = 6.3567%
- Average measured volume=2.9591 L → Em=0.0409 L → E=1.3623% with Standard deviation = 0.0612

4. Conclusions

Performance tests showed that the accuracy of the volume measurements shows levels very close to the value of the standard instrument, with an average error of less than 1.4%. The sample of 20 measurements used represents a statistically low population, but the trend recorded with a standard deviation of 0.0612 indicates a normal distribution with a trend centered on the real value. Since medical standards require less than 10% error in the measurement of medical instruments, this system is considered to meet the requirements for clinical application.

The maximum absolute error obtained = 6.3567% is predictable, given that this error occurs on the measurements where the flow has the greatest intensity, which represents a notable influence of the maximum error that can be obtained in the differential pressure sensor used. This benefits low flow measurements.

The sensitivity of 0.0681 L/s, which represents an error of 1.7% over the total measurement scale, could be considered relatively high, but it must be taken into account that flow variations at this level are imperceptible due to the patient's respiratory physiology and therefore they do not represent any risk in the operation of a medical instrument.

From the results obtained in the analysis, it is concluded that the system is efficient in flow measurements and is relatively simple to build. Therefore, it is considered to be a practical instrument to be implemented in medical instrumentation applications. Its efficiency in conditions of maximum humidity with condensations in different environmental temperatures, as well as after sterilization with autoclaves for reuse, will be evaluated in future studies.

Competing interests

The authors have declared that no competing interests exist.

Authors' contribution

All the authors of this document participated in the development and successful completion of project. ND performed the calculations, design, and fabrication of the Venturi tube, and optimized the test-run and error measurement method. AA designed the signal acquisition hardware. ED developed the PC analysis software. NA and SB collected information and medical advice.

Acknowledgements

We would like to thank the Regional Institute of Bioengineering (IRB), affiliated to the Computer and Neuroscience Center (CeReCoN), of the National Technological University of the Argentine Republic, for their scientific and technological collaboration in the overall development of the project. The support from the Biomedical Engineering Group (GIBULA) of the Universidad de Los Andes in Mérida, Venezuela is also acknowledged. We express our gratitude to individuals and respective institutions that have made this project feasible.

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Citation: N. Dugarte, A. Alvarez, E. Dugarte, N. Alvarado and S. Bhaskar. *Practical Design of Flow Meter for Mechanical Ventilation Equipment.* Journal of Computer Science & Technology, vol. 21, no. 1, pp. 42-48, 2021.

DOI: 10.24215/16666038.21.e05

Received: July 22, 2020 **Accepted:** December 15, 2020.