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Low-Power Wireless System for Temperature and Humidity Monitoring in Artificial Ventilation

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***Abstract*—Artificial ventilators are commonly used with Passive Heat-Moisture Exchangers (HME) to warm and humidify the inspired air in order to ensure a proper conditioning of inspired gases to the artificially ventilated patients. However, different aspects potentially affect their performances and this change in performance should be analyzed in-vivo during HME operation. In this paper, a wireless measurement system is proposed for the monitoring of air temperature and humidity in-vivo. The system is composed by a measuring device connected to the ventilating tube near the HME and a reading device connected to a Personal Computer (PC). Each device integrates a wireless transmission via low- power Bluetooth module that allows limiting power consumption. For the measuring device, the calculated power consumption when all the on-board components are working is about 15 mA, permitting a continuous monitoring for about 5 days and 16 hours with a rechargeable Li-Ion battery of 2050 mAh. A first prototype was manufactured and tested in the laboratory. Then, this prototype was tested with a setup specially developed to simulate human breath. The tests were conduced changing the respiratory rate and minute volume. Preliminary results are reported showing interesting aspects, such as the warm-up time of the HME. Furthermore, the results shows a direct dependence of humidity loss on frequency-volume ratio requiring future investigations. Clinicians are expected to use this system in-vivo to identify the correlations between clinical issues and HME performances.**

I. Introduction

Artificial ventilators are usually used to assist or replace spontaneous breathing mechanically. Typically an endotracheal tube connected to an artificial ventilator is inserted into the trachea in order to provide air to the lungs. In these con- ditions, inhaled air is no longer conditioned in the upper airways by human tissue. Therefore, relatively cold and dry gases entering the trachea may cause different problems, such as frequent coughing and excessive/thickened mucus production. During surgery, but also in Intensive Care, inspired air is usually humidified. Medical and engineering research in this area has proposed several instruments in order to ensure a proper conditioning of inspired gases to the artificially ventilated patients. These instruments are grouped in three categories, according to their nature: active heaters-humidifiers; passive heaters-humidifiers; mixed heaters-humidifiers. The passive heater-humidifiers are usually called Heat and Moisture Exchanger (HME). They are considered passive conditioners because they do not require an external power supply to warm and humidify the inspired air. Likewise it happens in physiological conditions, these conditioners hold up thermal power and a part of the water content of the exhaled gases (which are at about $32°C$, RH=100%), returning them back through the gas mixture delivered by the ventilator [1], [2]. The HMEs may also play the important role to prevent microbial contamination of ventilators circuits. The use of the HMEs turns out to be convenient also from an economic point of view; unlike to the active conditioners, they do not require external sources of energy, they are not bulky and do not require the aid of specialized personnel. However, because these are completely passive devices, the HMEs cannot accommodate to the surrounding environment conditions, whereas it happens in the physiological conditioning. Their performances are potentially affected by environmental temperature [3], patient's body temperature and distance from the airways. Some manufacturers provide values of water loss of the HME according to the International Organization for Standardization (ISO) standard 9360 -2: 2001, but this specification is not easily applicable in the clinical setting. Moreover, a minority of the HMEs available was analyzed in- vivo and only few studies are reported for ex-vivo analyses [4], [5]. Additionally, in ex-vivo studies, the values of water loss are not easily and intuitively translated into clinical benefits. Clinicians are expected to use these measurements to identify the correlations between clinical problematics and HME performances. To facilitate the clinical analysis, the change in performance should be analyzed by measurements in-vivo during the HME operation. However, intra-tracheal moisture measurements are technically complex. In the literature, different studies propose to sample the air from the trachea and to analyze this sample with an external humidity sensor [6]. In this procedure, it is necessary to pay close attention to prevent loss of water vapors due to condensation in the tube. On the other hand, the requirements for in-vivo monitoring are challenging. One critical aspect is the fast response in the humidity measurements. Fast humidity measurement systems for human breath are reported in the literature; these systems can regard relative [7] or absolute humidity methods [8]. In [9], a method for measuring temperature and humidity within the nasal cavity is proposed. Authors presented the sensor and a general idea of the dew point temperature measurement

system, but we do not found the realized instrument. Different other systems and methods are reported in the literature for similar applications [10], [11]. However, several aspects are not completely covered, such as long monitoring, wireless transmission, low-power consumption and user-friendly interface. Nowadays, there are no commercial devices fitting these monitoring requirements. Our aim is to develop such a measurement system for monitoring the HME performance in-vivo. In this research project, we designed a user-friendly low-power wireless system that can measure air temperature and humidity for in-vivo HME performance evaluation. The measurement system is placed on the tracheal tube, closer to the HME and, therefore, closer to the patient's mouth. In this application, a wireless transmission is preferred instead of a cabled solution for safety reasons. Critically ill patients can be agitated because of delirium, a common complication affecting up to 70% of patients, pain due to surgical procedures or trauma and other conditions [12]. In such cases, sudden movements can cause a traction on the connecting cables. This traction could damage the device and, more importantly, the ventilation circuit and the tracheal tube causing serious adverse events. A wireless system leaves much more freedom of movement to the patient and the operator, reducing the risk of tracheal tube removal or its dislocation into the right main bronchus. The system can immediately show to the clinicians the status of the airflow sent to the patient permitting to identify unwanted situations and/or a deterioration of the HME conditions. A first prototype has been manufactured and tested in the laboratory. Furthermore, the developed prototype was tested with a setup specially developed to simulate human breath. Preliminary experimental results demonstrate that the proposed system could be a starting point for in-vivo long monitoring of HME performance.

II. System Design

The measurement system topology is shown in Fig. 1. There

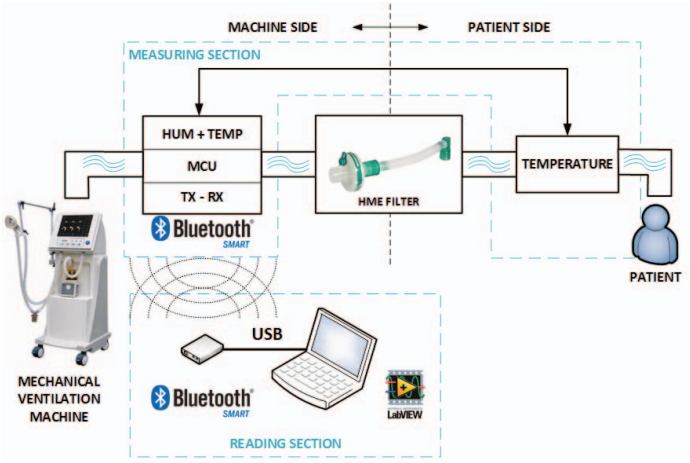

Fig. 1. Schematic system representation

are two main sections, called *Measuring Section* and *Reading Section*. The first is the core of the project, it executes the main operations: reads the sensors, builds the data block and manages the Bluetooth Low Energy (BLE) connection; this is also subdivided in two parts: machine side and patient side. Project specifications require two different measurement points, before and after the HME filter. Before the HME (machine side) temperature and humidity are measured, after that (patient side) only temperature. The Reading Section reads the data sent by Measuring Section and it provides to the user an easy-to-understand set of data.

A. Measuring Section

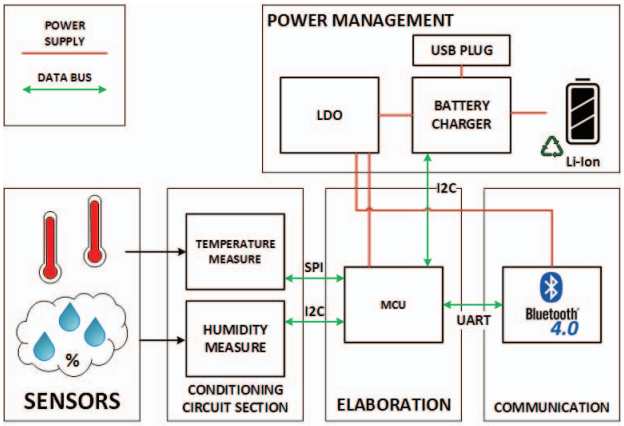

Fig. 2. Block diagram of the measuring section

The block diagram of the measuring section is reported in Fig. 2. It is divided in five subsections:

- Sensors: it contains two temperature sensors (Pt1000) and a humidity sensor (P14 FemtoCap). The two temperature sensors are classical thermistors suitable for accuracy and small dimensions; the humidity sensor P14 is chosen because it is a fast response sensors, the data-sheet reports a recover-speed from 50 %RH to 0%RH in 3 seconds; P14 returns relative humidity value which must be converted in absolute humidity.
- Conditioning Circuit Section: is based on two different integrated circuits, one for the temperature (MAX31865) and one for the humidity (ZSSC3123). MAX31865 (by Maxim) is an easy-to-use resistance to digital converter, optimized for platinum resistance temperature detectors. ZSSC3123 (by ZMDI) is an accurate capacitance-to-digital converter. An integrated circuit (IC) solution has been chosen for small dimensions.
- Elaboration: for managing the whole board a PIC24 microcontroller unit (MCU) (by Microchip) has been chosen.
- Communication: a BLE integrated transceiver with an on-board antenna has been implemented for low-power characteristics. The Bluetooth module is RN4020 commercialized by Microchip.
- Power management: this section contains a Li-Ion battery pack, with its own charger BQ24251 (by TI) which integrates a power path management; after that, to ensure

the correct voltage to the components on the board, there is a Low DropOut (LDO) voltage regulator, TPS7133 (by TI), assuring a reference voltage of 3.3 V. The BQ24251 integrates a Universal Serial Bus (USB) compliant interface. It detects which device is supplying the power and automatically scales its own drawn current.

The implemented firmware in the MCU is a non-ending loop designed to read the signals from sensors and to transmit the data every 250 ms. This interval could be reduced up to 60 ms.

There is also a fault detection procedure, which tells the operator if everything is OK or, if something is gone wrong and from where the error comes. The role of this section is to retrieve data from the sensors and to send a periodic update to the Reading Section. The transmitted data are raw; hence, conversions are made by Reading Section, because usually there are no practical limitations in the computing power of a PC (Personal Computer) or a mobile/tablet. The circuit board was manufactured in Printed Circuit Board (PCB) technology, and its dimensions are 92.5 x 77 mm.

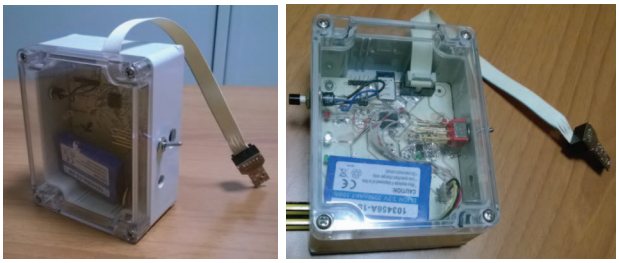

Fig. 3. Picture of the manufactured measuring section

B. Reading Section

Reading Section permits to receive measurement points by a BLE, and then to send all the data to the PC. The use of a BLE communication makes Measuring Section virtually discoverable and readable by every new generation smartphone or tablet with BLE compliant hardware. In the world of laptops and desktop computers, BLE is not as widespread as in the mobile, this means an absence in the Operating System (OS) Application Programming Interface (API) for handling this kind of devices. Therefore, in this preliminary prototype, we used LabVIEW in association with a specific manufactured module. This module is required because also LabVIEW founds its functionalities on the APIs of the OS, so it is necessary to pass through an adapter. This module translates the Universal Serial Bus (USB) serial bus to a Transistor-Transistor Logic (TTL) compatible Universal Asynchronous Receiver-Transmitter (UART) protocol, and then it is connected to the BLE. The LabVIEW VI (Virtual Instrument) is shown in Fig. 4. The schematic of the adapter is represented in Fig. 5, and the manufactured board is reported in Fig. 6. Its dimensions are as quite as possible similar to a standard USB pen drive. As introduced before, this section converts the raw data to temperature and humidity measurement values. The

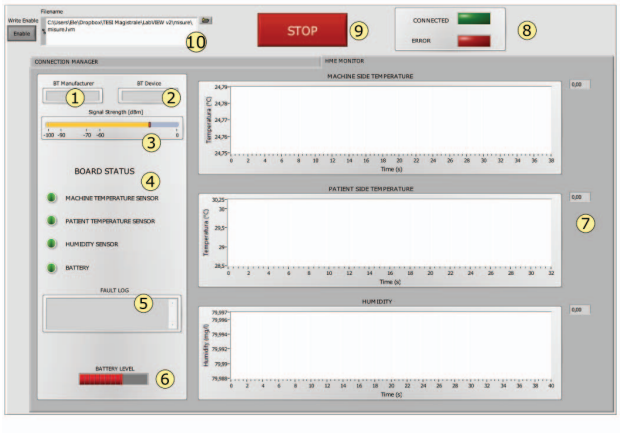

Fig. 4. LabVIEW measure monitor: (1) BT Manufacturer, (2) BT Device, (3) Signal Strength in dBm, (4) Board Status: green LEDs if OK, red LEDs if a fault is present, (5) Fault Log: a verbose description of the Fault Type and history, (6) Battery Level, (7) Measures: in this panel are represented, on a temporal chart and on a digital meter, absolute humidity and temperatures, (8) BLE connection status, (9) VI main stop, (10) Saving path of the measure file

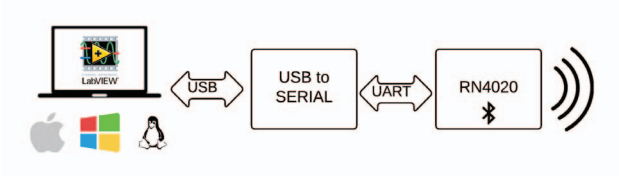

Fig. 5. USB-to-BLE module schematic

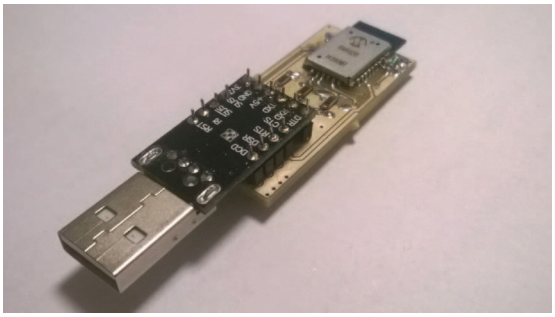

Fig. 6. Picture of the manufactured USB-to-BLE module

conversion of temperature values is performed as explained by MAX31865 data-sheet. For the absolute humidity value two steps are required in order to convert P14 capacitance raw data to %RH, and then, using temperature values, to an absolute humidity value evaluated by mg/l. The adopted conversion formulas are reported in [13].

III. Laboratory Tests

The overall system was tested in a climatic chamber (Perani UC 150/70) comparing its output data with the measurements from two reference sensors, linked to two digital multimeters (Tektronix DM2510G); these multimeters are connected to a PC by a General Purpose Interface Bus (GPIB), IEEE 488.

The whole measurement procedure is controlled by an own-written LabVIEW VI. This setup is shown in Fig. 7. The two

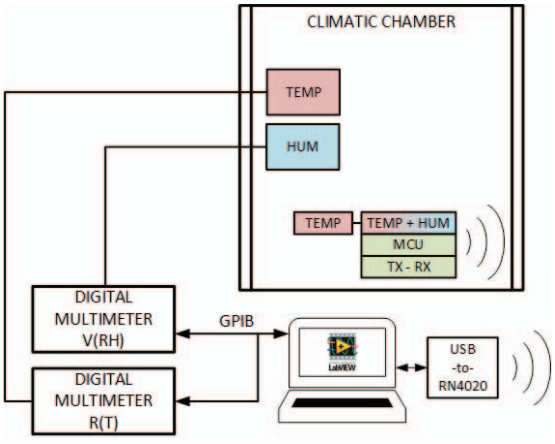

Fig. 7. Preliminary tests setup

reference sensors are a Pt100 for temperature measurements and a HIH3610 (commercialized by Honeywell) for humidity measurements. The LabVIEW VI software is structured to send a data request to each multimeter every time it receives new data from the system. Initially, temperature was changed maintaining humidity inside the climate chamber constant, and then humidity was changed maintaining a constant temperature. Each test has a duration of about 35 minutes. In the literature [14]–[17], Authors agree that the inhaled air should, for the purpose of a proper conditioning, leave the conditioner at a temperature of about 30-32 $°C$, with an absolute humidity of about 30-33 mg/l. The term "conditioning" refers to the process of ensuring proper body homeostasis through the humidification and warming of inspired gases. Therefore, in the experimental test, we used a temperature range with a central point at $30°C$ and a span of ± 20 $°C$. Preliminary experimental results are shown in Fig. 8 and 9.

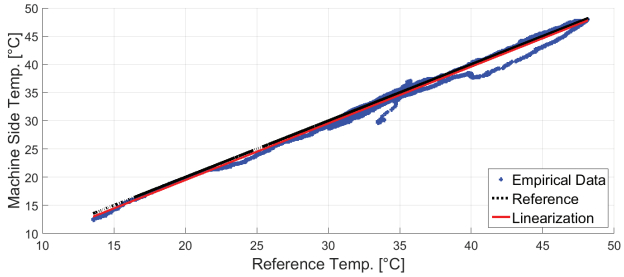

Fig. 8. Laboratory test of machine side temperature

These figures show an inherent hysteresis of the temperature sensors starting from 30 $°C$. The least square approximation reports these equations:

$$PT_T = 1.035 \cdot (REF_T) - 1.74 \quad (1)$$

$$MC_T = 1.006 \cdot (REF_T) - 0.64 \quad (2)$$

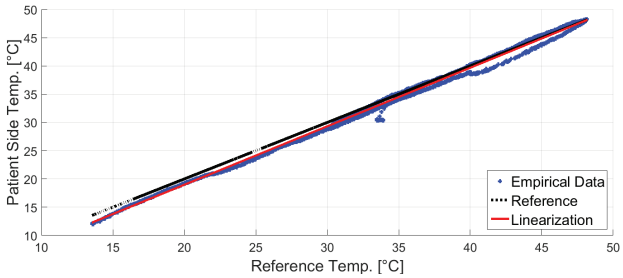

Fig. 9. Laboratory test of patient side temperature

Where PT_T is the patient side temperature and MC_T is the machine side's. The data retrieved let us estimate linearity, hysteresis, the coefficient of determination (R^2) of the sensors and the average deviation from the reference, reported in Table I. The same data analysis was done for the humidity measurements (Fig. 10); the parameters reported in Table I were calculated also for humidity measurements. P14 least square approximation equation is:

$$RH_{P14} = 0.943 \cdot RH_{HIH} + 0.97 \quad (3)$$

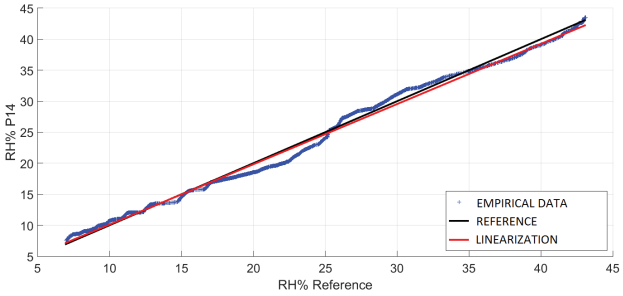

Fig. 10. Laboratory test of machine side relative humidity

Table I
LABORATORY MEASURED SENSORS CHARACTERISTICS

	PT Side Pt1000	MC Side Pt1000	MC Side P14
Linearity	$-3.01°C$	$-3.39°C$	$4.66\%RH$
Hysteresis [%FSO]	4.15	6.47	3.6
R^2	0.9940	0.9891	0.9338
ΔT [°C]	$\approx \pm 0.54$	$\approx \pm 0.44$	–
$\Delta\%RH$	–	–	$\approx \pm 1.62$

IV. CLINICAL TESTS

Preliminary clinic tests were executed using a respiratory circuit with inhalation and exhalation branches, an interposed plastic balloon simulating the lungs and a mechanical ventilator allowing the possibility to modify the respiratory rate and volume. The volume of inhaled and exhaled gas with

each respiratory cycle is defined as the tidal volume. The total volume in the respiratory cycle per minute is the minute volume, which is influenced by both the respiratory rate and the tidal volume. The schematic diagram of the test equipment is depicted in Fig. 11. The laboratory tests were conduced

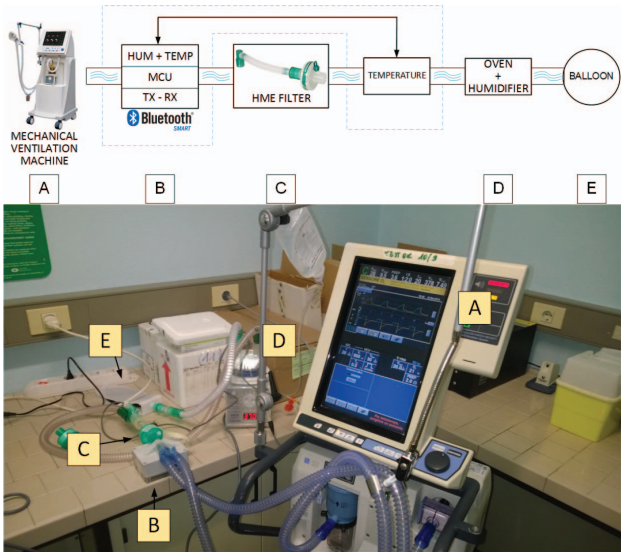

Fig. 11. Test Equipment:(A) Mechanical ventilator, (B) the system under test, (C) HME filter, (D) Oven and air humidifier, it conditions the air as the lungs does,(E) a lungs simulator; it is a plastic balloon which simulate the lungs elastic air emission by relaxation

in a safe environment, only using mechanical simulators and machines; these tests emulate breathing acts. They are composed by two different phases:

- inhalation: air flows from the mechanical ventilator through the sensors, the HME and an oven/humidifier, inflating the balloon which acts as lungs simulator;
- exhalation: air gets back to the mechanical ventilator by elastic relaxation of the balloon, with temperature and humidity according to physiological levels of the human breath.

The system has been tested with a new HME filter trying to highlight the warm-up time. The test was conducted, as described before, changing respiratory rate and minute volume, starting from a respiratory rate of 10 with a minute volume of 15 l/min, ending with a respiratory rate of 30 and a minute volume of 45 l/min and vice-versa, for four times. A detail of the test at a respiratory rate of 10 with a minute volume of 15 l/min is shown in Fig. 12. Observing the temperature curve of the patient side, the two phases (inhalation and exhalation) are clearly shown.

These tests were done aiming at monitoring the amount of water vapor loss by the HME filter and the difference between temperature from patient and machine side. Depicted in Fig. 13, an increasing in the absolute humidity with an increasing in respiratory rate and minute volume is shown. Furthermore, absolute humidity drift at every test round is evident. This denotes a sort of hysteresis of the filter during its warm-up time. Also for the temperature, the behavior of the filter

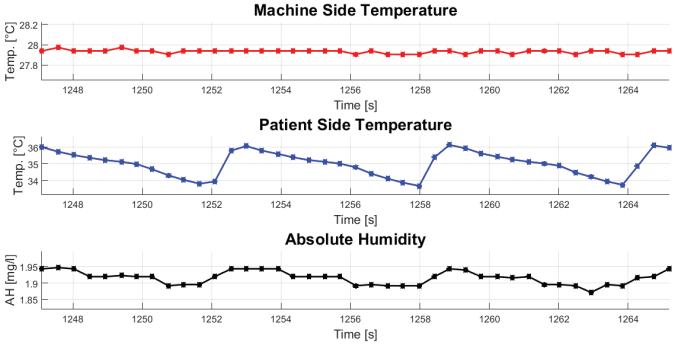

Fig. 12. Detail of a test with a respiratory rate of 10 act/min and a minute volume of 15 l/min.

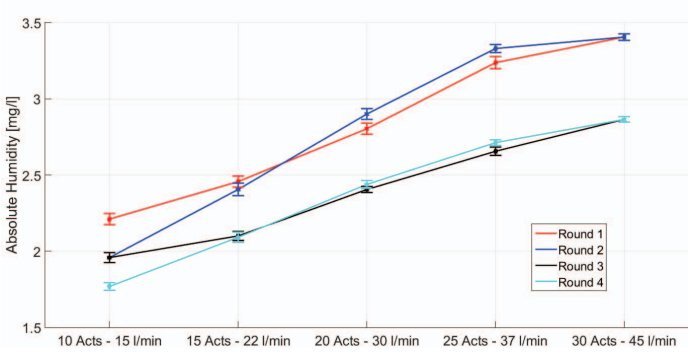

Fig. 13. Absolute humidity calculated with the proposed system for different respiratory rates and minute volumes.

differs from the beginning to the steady state. The machine side temperature has remained for the whole test into the range of 27.65 - 29.15 $°C$, as shown in Fig. 14. The ΔT between the two sides grows up to a difference of about 7 $°C$, and then it remains more or less constant. These preliminary results show a direct dependence of the absolute humidity with respiration rate and minute volume and the necessity of a certain amount of time to permit to the HME filter to maintain the correct temperature (around 36 $°C$) of the patient side airflow. These aspects will require future investigations.

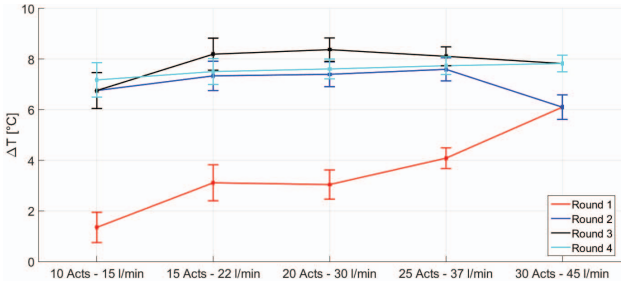

Fig. 14. ΔT between machine side and patient side for different respiratory rates and minute volumes.

V. Conclusion

A preliminary measurement system for in-vivo monitoring of HME performance was developed. This system can measure

air temperature and humidity transmitting the data wirelessly to a reading unit connected to a PC. A Bluetooth low-energy module permits low-power consumption assuring long-time monitoring. A specific virtual instrument interface was developed permitting user-friendly operations. Tests was conducted in the laboratory to analyze the behavior of the sensors and comparing the output data with two reference sensors. Furthermore, the developed prototype was tested with a setup specially developed to simulate human artificial ventilation as it is currently used in Anesthesia and Intensive Care. The preliminary results permitted to calculate the warm-up time of the HME. In addition, the results showed a direct dependence of the absolute humidity with respiration rate and minute volume requiring future investigations. Future studies should evaluate the proposed system in-vivo to identify changes in HME performances and possible correlations with clinically relevant outcome measures, helping in the prevention of lung diseases induced by inadequate humidification of inhaled gases during mechanical ventilation.

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