Development of a Handheld Side-Stream Breath Analyser for Point of Care Metabolic Rate Measurement

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Abstract. A novel handheld side-stream breath analyser has been developed. The low-cost device offers breath-by-breath measurements of O_2, CO_2, temperature, relative humidity and gas flow rate. Metabolic rate can be calculated from the inspired and expired gas concentrations; a knowledge of this over a 24 h period can guide calorific intake. The analyser provides easy-to-read results on either a laptop or smart phone. Results for the O_2 and CO_2 sensors demonstrate the device's potential for metabolic rate breath analysis. The O_2 sensor is not able to follow changes in O_2 concentration during the breathing cycle; however, the newly developed affordable and low-power consumption CO_2 sensor performs comparably to a bulky, high power consumption commercial device.

Keywords: Breath analysis · Metabolic rate · Energy expenditure · Smart phone sensing · Handheld breath analyser

1 Introduction

The measurement of energy expenditure (EE) through indirect whole body calorimetry is considered a 'gold standard' for human metabolism measurement [1]. In this system, subjects are studied in a sealed room where the difference in the concentrations of O_2 and CO_2 entering and leaving the room are measured, from which the metabolic rate is determined. Besides the enormous cost involved in constructing and maintaining calorimetry rooms, the measurements are limited to small groups of volunteers and patients. In order to enable a wider study population, a novel handheld analyser that can measure exhaled and inhaled O_2 and CO_2 has been developed. The low cost analyser permits real-time breath-by-breath data logging of the gas contents during an exhalation. The device has been designed considering the needs of patients, where the analysis of metabolic rate in a free-living environment is desired, without the need for a trained practitioner to make the measurements. The low power consumption of the device enables the measured O_2 and CO_2 content of breath samples to be recorded either on a smart phone or on a laptop computer, as required.

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The total EE of a human being can be categorized into three components: resting metabolic rate (RMR), thermal effect of food (TEF) and physical activity [2]. Whilst respiratory chamber EE measurements give accurate and reproducible measurements [3], the subject is confined to a small room with limited ability to undertake activities of daily living (ADL). It has been reported that the energy expended in physical work is the most variable component of daily EE, where it can contribute between 15 and 30 % of the total EE [4]. RMR contributes between 60 and 75 % [4], where the remaining 10 % is attributed to TEF. The device that has been developed offers the capabilities to take multiple metabolic rate measurements throughout a 24 h period.

EE has been shown to vary by the time of day, where an increased EE is observed after eating or exercise [5]. Through monitoring of EE at regular intervals over a period of one day, and then repeating this over many days and weeks, we aim to investigate how EE varies in a free-living environment. The doubly-labelled water technique has been used to research daily EE in free-living human beings previously [6]; however, this technique does not give a dynamic measurement of energy expenditure but merely a measure of total energy over the observation. Additionally, DLW, as whole-body calorimeters, requires specialised knowledge, equipment and a non-trivial protocol. Hand-held calorimetry based on breath analysis, gives a uniquely low cost, non-invasive and low-risk measuring device suitable for use in an EE population monitoring programme. Metabolic carts provide a portable means of investigating a patient's metabolism, only a limited number of subjects can be analysed with such instruments, in part due to limited availability, long warm-up times and necessary calibration procedures [7]. Predictive equations are often used to calculate calorific requirements, in particular in intensive care units where correct energy intake is critical to recovery, however underlying assumptions may result in large errors [8].

Whole body calorimeters and metabolic carts are expensive and complex to use. Whilst domestic devices are available the quality and accuracy of the gas sensors in these are not at the level required for clinical quality measurements. A handheld calorimeter incorporating innovative sensor developments negate the need for long warm-up times associated with traditional sensors and low manufacturing costs means an affordable, simple-to-use precision hand-held indirect calorimeter is now possible.

2 Methods

To enable measurements to be taken periodically over a normal day for a sustained period, the analyser must be compact and portable whilst being able to deliver accurate and reproducible measurements. A side-stream system was selected for its reduced size and to provide the gas sensors with a constant and much reduced gas flow rate when compared to the main-stream flow. In our device, a sample of both exhaled and inhaled gas is extracted from the main-stream exhalation at a rate of 150 ml/min. A block diagram of the system is shown in Fig. 1. Metabolic rate can be determined from the rate at which O_2 is consumed, $\dot{V}_{O_2}$, and CO_2 is produced, $\dot{V}_{CO_2}$, through published equations, e.g. the abbreviated Weir equation (1) [9].

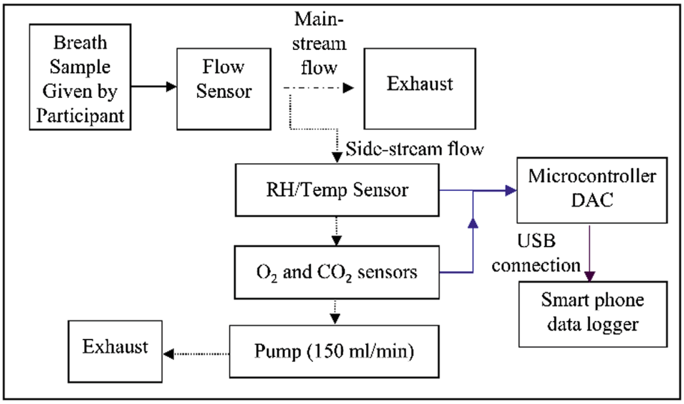

Fig. 1. Functional block diagram of side-stream sampling system, showing sensors included and the data-logging procedure.

$$EEtotal[kcal.] = 3.9\left(\dot{V}_{O_2}\right) - 1.1\left(\dot{V}_{CO_2}\right) \tag{1}$$

Sensors are included for carbon dioxide (CO_2), oxygen (O_2), temperature, relative humidity (RH) and flow. An electrochemical sensor is used for O_2 monitoring (City Technology, MOX-20, UK) and a SFM3000 (Sensirion, Switzerland) for flow rate measurement. A ChipCap2 sensor (GE, USA) provides both temperature and RH measurements in the side-stream tubing. The data are logged via a microcontroller (PJRC, Teensy 3.2, USA) on either an Android smartphone or a Microsoft Windows laptop computer. An acetal shell was used to house the sensors, with 3D printed parts used where necessary, to meet the manufacturing tolerances required for our sensors.

The completed unit is shown in Fig. 2, with data logging performed on a smartphone. A novel non-dispersive infrared (NDIR) MEMS based sensor was developed for the breath analyser. Low cost commercial devices failed to provide the necessary response times and the compact low-power design required for a portable breath-by-breath sensor. The power consumption of the newly developed system was measured to peak at around 100 mW. Experimental data collected previously from respiratory chambers indicated the O_2 content on an exhalation needed to be measured to an accuracy of 0.52 % and the CO_2 content to 1.25 %, in order to obtain EE to an accuracy of 1 %. To put this requirement into perspective the maximum change in energy expenditure associated with the digestion of food (TEF) is between 30 and 70 kJ/h on a resting metabolic rate (RMR) of around 400 kJ/h [10].

A LabVIEW Virtual Instrument (National Instruments, v2014) was developed as an interface to visualise the outputs from the breath analyser on any Windows computer (Windows 7 used in the experiments presented, as shown in Fig. 3). The O_2, RH and temperature sensors were analogue inputs, where a raw voltage was recorded and converted into quantity. The flow sensor provided a digital output, which was converted into a flow value in litres per minute. The CO_2 sensor output is a sinusoidal waveform, the amplitude of which corresponds to the CO_2 concentration in the sensor

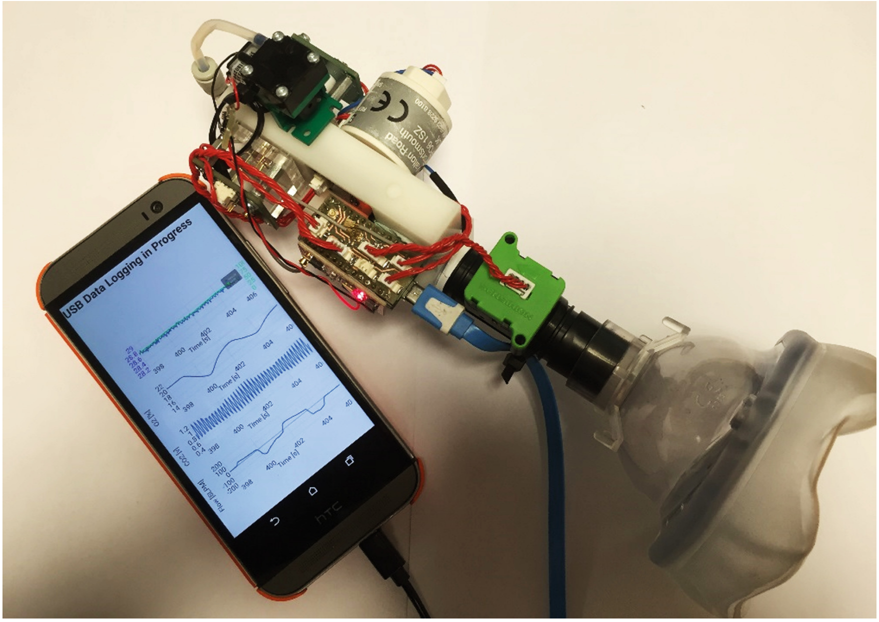

Fig. 2. Side-stream sampling system with smart phone data logging system. Software shown on screen demonstrating ambient conditions, with CO_2, O_2, temperature, RH and flow plots.

chamber. Further post-processing is required to convert the voltage recorded to the gas concentration desired than can currently be performed in real time. Thus only the raw voltage output is plotted on the readout graphs.

The microcontroller was selected to allow up to 16-bit analogue to digital conversion and the necessary I^2C and SPI connections for the gas sensors. The data logging rate is selectable, where a 200 Hz rate provides a suitable balance between file storage size and capture speed. A serial output from the microcontroller, and simple USB connection, enables the output data to also be displayed on a smart phone. The low-power nature of the system enables it to be powered from either a laptop or smart phone alone. Software written in Python (v2.7) permitted displaying and logging of the results on an Android operating system (v5). Future generations of the system will allow the data to be logged wirelessly, to permit Apple iPhone mobile phones to be compatible with our system.

The current Android smartphone application plots the recorded data in real time, including conversion from raw data to a physical quantity, with the exception of the CO_2 readout. A screen print of the application is shown in Fig. 3(a), after a subject has exhaled and inhaled twice through the device. The current generation application allows limited analysis of the recorded data, extracted from the logged data files. Two basic features are extracted, the minimum value and peak value over the course of one measurement, as shown in Fig. 3(b). The flow sensor output provides an indication of when a subject is exhaling or inhaling. Current work includes validation of a complete breath sample, by the flow rate recorded through the flow sensor. The flow rate can be

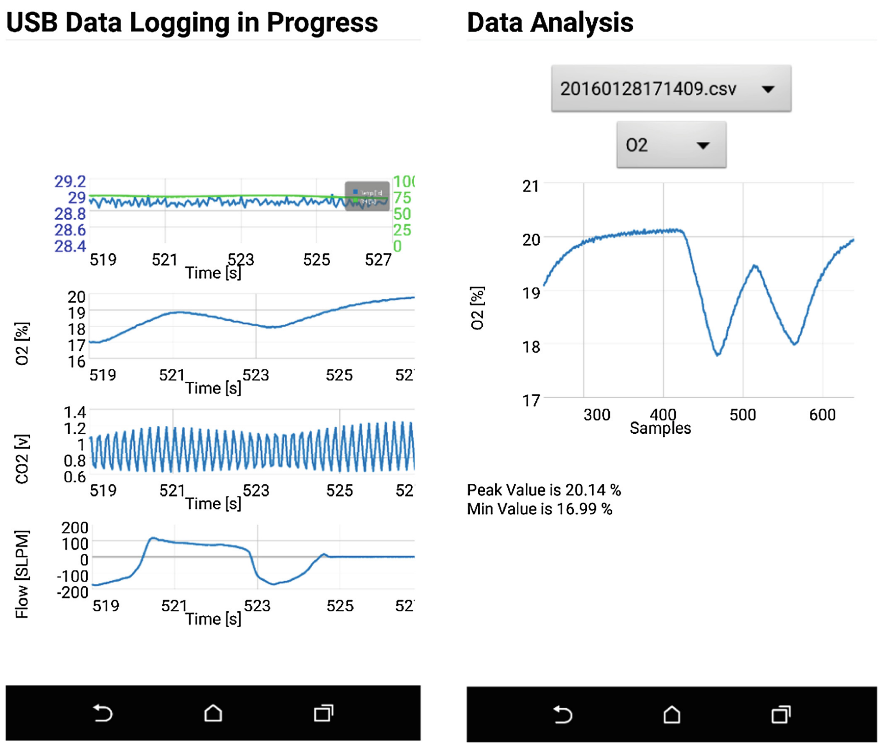

Fig. 3. (a) Screen print showing the data logging application on an Android smartphone, logging temperature, relative humidity, O_2, CO_2 and flow rate; (b) Peak value and minimum value are extracted from the measurement data in a separate data analysis section of the smartphone application.

used to determine the baseline readings and used to indicate the end of an exhalation, when the gas contents of exhaled breath is of interest for our metabolic rate calculations.

Our system has been trialed on three volunteers at the University of Warwick. Data from one subject are presented from experiments where subjects were asked to breathe through the device for a period of 1 min (see Fig. 4). During the experiment, subjects were requested to maintain a fixed, relaxed breathing pattern and regulate their breathing to have a breath-to-breath period of either ten seconds or six seconds, whilst maintaining a ratio of 1:1 between the inhalation period and the exhalation period. The six second cycle provides a similar respiratory rate to that of a resting adult, where 10 to 12 breaths per minute are considered normal for an awake adult [11].

The CO_2 sensor was calibrated against a commercial breath-by-breath sensor, for the 10 s breathing cycle provided by the first subject. The commercial sensor was connected to the output of the side-stream chamber, prior to the pump. The O_2 sensor was calibrated on a gas testing bench, using known values of gas concentrations. The gas concentrations were generated using a mixture of nitrogen and synthetic air. These calibration values were used for subsequent tests.

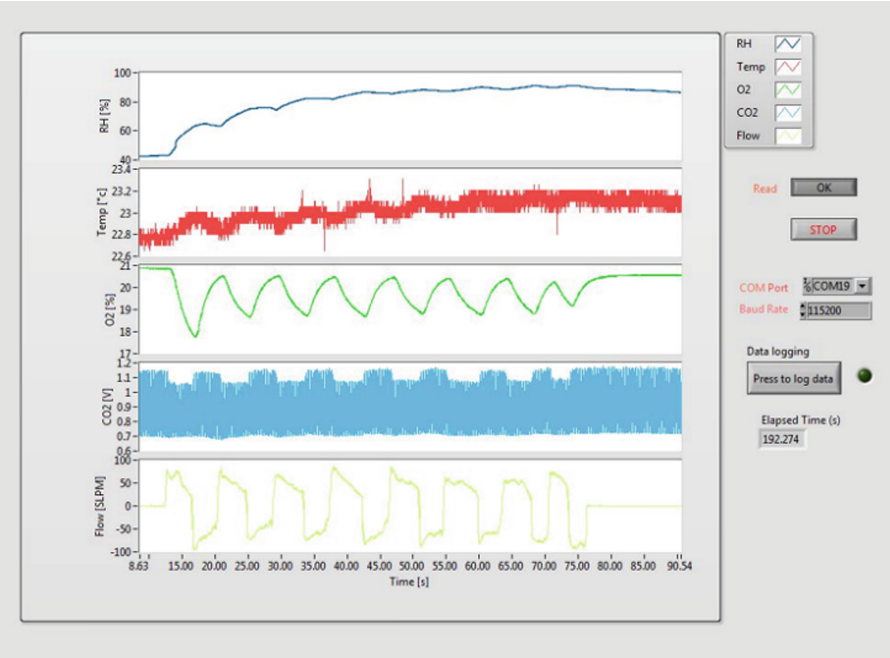

Fig. 4. LabVIEW VI logging system (for Windows computers) showing O_2, CO_2, temperature, RH and flow graphs for a 90 s experimental period with 8 exhalations. Note that during inhalation flow is positive and during exhalation flow is negative.

3 Results

The results obtained were consistent and Figs. 5 and 6 show an example of the results for one subject with a breath period of 6 s. A 90 s section of CO_2 data containing 10 breaths from one subject is shown in Fig. 5. The oxygen sensor output is shown in Fig. 6.

The CO_2 sensor captures the exhalations well, demonstrating a similar performance to that of a commercial device. The t_{90} response time (time for the response to reach 90 % of its final value) was on average ~1.7 s for the research sensor. The commercial device offers a $t_{10} - t_{90}$ response time of around 100 ms. Comparison between the commercial sensor and research device yielded a variance of 0.29 % for a 95 % confidence interval. The mean recorded CO_2 value across the 10 breath samples was 4.70 %, which is within the expected range for normal subjects of 4–5 %.

The research CO_2 sensor is driven at a rate of 5 Hz, which limited the number of measurements that could be taken per breath. A faster drive rate, perhaps up to 100 Hz, would enable a greater number of measurements to be taken per breath. The sensor output does not plateau for all the 10 sample breaths. This outcome is perhaps in part due to the limited measurement rate. The dimensions of the SOI CMOS IR emitter are related to the response time of the device. The larger emitter size chosen in these

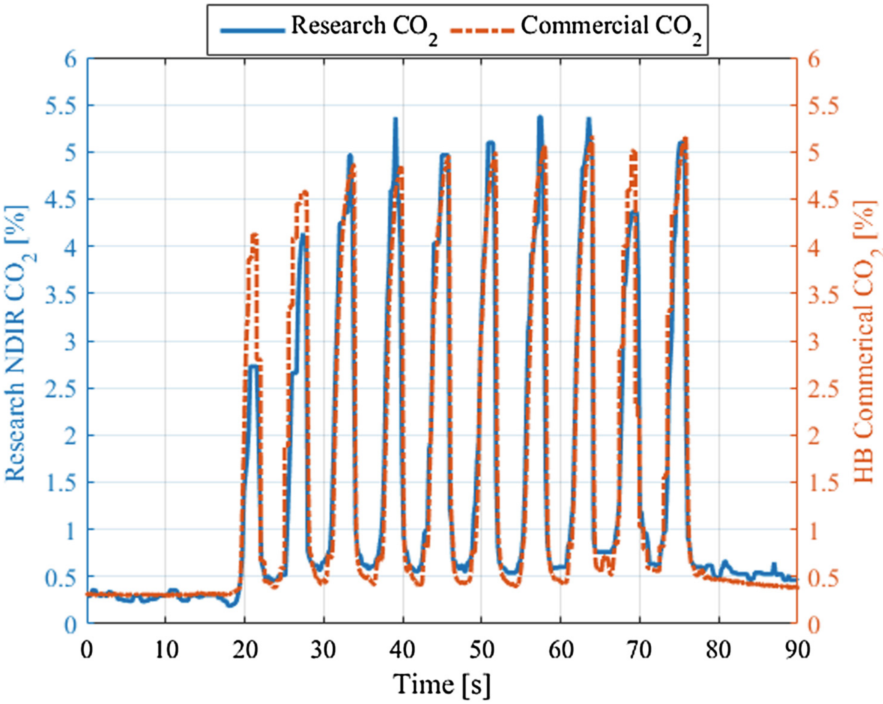

Fig. 5. CO_2 sensor output compared to commercial device, for 10 breaths provided by a subject over a period of 1 min.

experiments provides greater emissivity for a higher detector response, but prevents the faster drive signals desired for reduction in noise and greater measurement rate.

The O_2 sensor does not perform to its specification for breath-by-breath measurements. The baseline for room air is $\sim$21 % O_2. The sensor does not return to this value when the subject is inhaling. The sensor could be affected by the elevated humidity level in an exhalation (peak of measured $\sim$80 % in the side-stream after an exhalation). Furthermore, the minimum O_2 concentration in normal subjects is expected to be in the range of $\sim$16–17 % in exhaled breath; a 4–5 % decrease compared to inhaled air.

A complementary response would be expected from the O_2 and CO_2 sensors to a number of exhales and inhales from a healthy subject. The response of the CO_2 sensor, demonstrating a peak to a maximum concentration (average amplitude of 4.1 %), indicates the exhaled breath gases are reaching the sensors located in the side-stream. The O_2 sensor does not recover to the baseline throughout the inhalation phase of any of the 10 sample breaths. The mean value during inhalation is 19.5 %, where the mean value during exhalation is only 0.9 % lower. The cause of this inadequate response could perhaps be due to the sensors detection principle. The electrochemical device measures O_2 concentration through a reaction inside a gel solution. The reaction rate could become slower as the device ages, due to the layers becoming blocked or from contamination.

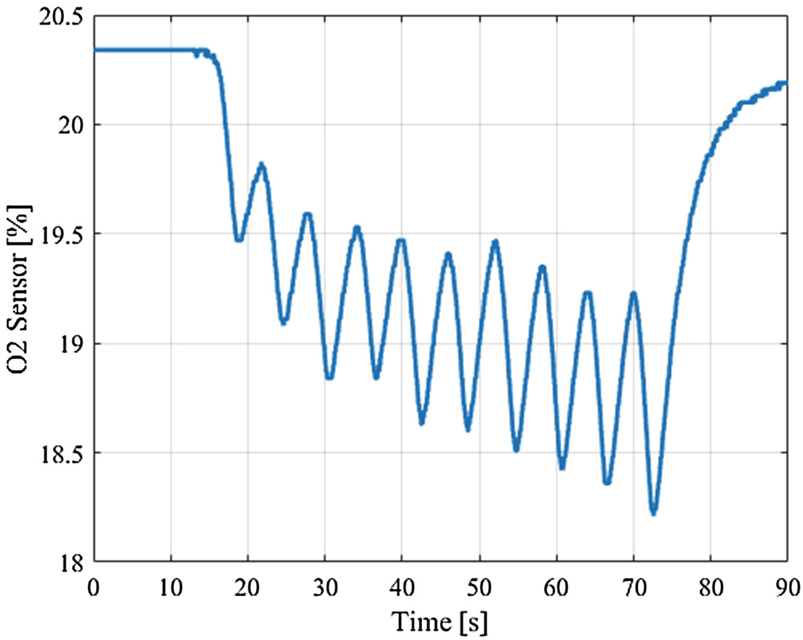

Fig. 6. O_2 sensor output for same subject as Fig. 4, exhaling with a 6 s breath cycle.

The smartphone logging application is a desired feature of a hand-held breath analyser, to enable easy integration into daily living as a portable unit. For the case of a personal breath analyser, when measurements are taken for a matter of minutes over the course of one day, a smartphone is ideal for metabolism analysis. However, in this prototyping phase, a computer was often used to log the measurement data. The current breath analyser unit connects to the data logging device by a USB power, which also serves as the power supply. Smart phones have only one USB port for connections to external devices, thus only one breath analyser can be connected. Furthermore, the phone cannot be charged simultaneously. Perhaps a future device, with wireless integration, will provide a solution to these limitations, which are not present on a traditional computer logging system.

The android application was used to log data at a rate of 200 Hz. The current measurement program is limited in functionality, due to the amount of data recorded. It was found no more than 10 s of data could be displayed on the screen for all five sensors. Therefore, any further data processing was completed after the data had been logged into file storage.

4 Conclusions

A microcontroller based portable breath analyser has been developed. In terms of gas sensors, a novel research CO_2 sensor has been compared to a commercial CO_2 sensor, where it demonstrated similar performance. The performance of the O_2 sensor requires further investigation. Whilst the specification for the device suggests that it should be able to follow breath-by-breath changes in O_2 the measured data did not demonstrate the expected change in O_2 concentration during exhalation (observed maximum $\sim$2.5 %, expected 4–5 %). This could be due to elevated humidity levels in the side-stream section ($\sim$80 %).

The side-stream system demonstrated a promising performance. It was powered by a smart phone alone, with peak consumption $\sim$100 mW. In the future we aim to develop a wireless sensor system, which will allow measurements on other models of smart phone as the current system is compatible only with Android phone technology.

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