BREATH-ANALYSIS: A NON INVASIVE METHOD FOR DETECTION OF DISEASES AND HEALTH CONDITIONS USING GAS SENSORS

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Abstract— Studies have shown a strong and essential correlation between volatile organic compounds and certain other gasesin exhaled breath and incidence of specific diseases thus offering strong potential for clinical diagnostic application using exhaled breath gas sensing and analysis. Breath analysis provides a non-invasive technique making it more agreeable and efficient compared to current invasive techniques such as blood or urine sampling. A handheld device which is cost effective, reliable and capable of early diagnosis of diseasesand health conditions is needed. This technology should also be able tobe applicable in the diet and fitness domain by detecting Ketogenesis (fat burn). With the advent of MEMS technology, solid state sensors have become more and more common in sensor modules used to detect various gases. The readings obtained from the sensors on detection of the biomarkers can also be made more accurate by employing an Artificial Neural Network (ANN).Compact hardware housing, user friendly presentation of dataand server data storage should be developed in such a handheld device. These devices would make clinical diagnosis a rapid, non-invasive method adopted at the triage station in hospitals. In this paper the design, working and utility of a prototype handheld device is discussed.

Keywords— Artificial Neural Network, Breath Analyser, Fat-burn, Ketogenesis, non-invasive, solid state sensors.

I. INTRODUCTION

Human breath constitutes a wide array of gases including volatile organic compounds (VOCs) and biomarkers in it are closely associated with incidence of certain diseases. Hydrogen sulphide, Acetone, Ammonia, and Pentane are known to be associated with Halitosis, Diabetesand various other diseases. Hence, a handheld device capable of detecting Diabetes [1], Halitosis [2], CO poisoning [3], alcohol intoxication[4][5], breath temperature [6] and Ketogenesis (fat burn) [7]was fabricated as a prototypeand tested. The device measured Acetone, Carbon Monoxide (CO), Hydrogen Sulphide (H_2S) and Ethyl Alcohol concentrations in the exhaled breath using solid state gas sensors [8][9]. The prototype designed used an array of solid state and electrochemical gas sensors, a microcontrollerfor analysis along with power and display circuitry. The prototype was fabricated by 3D printing rapid prototyping method. The gas sensors were calibrated using a mass flow controller calibration setup and calibration gas cylinders.The data obtained from the sensors on detection of biomarkers was made accurate by employing an Artificial Neural Network (ANN) [10]. Wireless data transmission of user data to server, data processing and data presentation on website was also provided.

II. DISEASESAND FITNESS CONDITION OVERVIEW

Elevated blood sugar level and disorder in metabolism due to Diabetes Mellitus (DM) may lead to heart diseases and limb amputation or even premature death. Breath acetone (CH_3COCH_3) levels are found to have a strong relation with the Type-1 Diabetes. In Diabetic people higher levels of ketones like acetone is found due to decarboxylation of Acetoacetate which derives from lipid peroxidation or lipolysis. Studies have shown patients diagnosed with diabetes havingbreath acetone concentration of over 1.8 ppm. Healthy individuals recorded nearly 2-6 times lesser levels approximately 300-900 ppb of acetone concentration.

Carbon monoxide (CO) is a colourless, odourless gas produced from incomplete combustion of fossil fuels. Haemoglobin has more affinity for Carbon Monoxide than oxygen which makes it toxic by forming the compound Carboxyl-Haemoglobin (COHb). Light headedness, headache, nausea, coma are a few symptoms of COHb formation with the initial effects not being easily detected. It has been shown that COHb levels are directly proportional to exhaled breath CO concentration. The range of carbon monoxide concentrations obtained among non-smokers was 0-6 ppm with a mean of 1.26 ppm and among smokers was 1-68 ppm with a mean of 16.4 ppm.

Halitosis (oral malodour) is a symptom where exhaled breath constitutes an unpleasant odour. A bad odour is composed of hydrogen sulphide (H_2S) and other volatile sulphur compounds (VSCs) whose concentrations exceed 1 ppm for people with halitosis whereas their concentrations are less than 150 ppb for healthy people. The breath alcohol level is directly related to the degree of intoxication.Alcohol intoxication can be classified into 3 main ranges [11]: Haziness from 130 to 260 ppm, slight drunkenness from 260 to 390 ppm and drunkenness from 390 to 650 ppm.

The prototype breath analyser can also be used as a Ketogenesis detector which analyses the exhaled breath for high acetone levels and gives an estimate

of whether the diet plan employed by a person is working and if he/she is burning fat cells. "Fat-burning" happens due to ketosis when carbohydrates and other nutrients fail to give energy to the body. If the sensor detects elevated amount of acetone or other ketones in the exhaled breath, it implies a strong correlation to fat burning. It is also shown in different studies that the heart rate also affects the rate of fat burning in humans. The workout regime has to be maintained in a particular range of heart rate specific to the age of the person for maximum fat burn. One such study finds the maximal fat oxidation range lies between 60.2% to 80% of the person's maximal heart rate[12]. All the gas sensors are calibrated to achieve accuracy for the aforementioned concentration ranges.

III. SENSORS SELECTED

In the prototype, for H_2S sensing, TGS 2602 [13]odorous gas sensor was chosen.A TGS 2620 [14] solvent vapour alcohol sensor was used to measure breath ethanol levels. TGS 822 [15] being sensitive to organic vapour solvents was chosen for acetone sensing. The above sensors are all solid state sensors, show very good response time (<1 min) and sensitivity characteristics. An electrochemical sensor, TGS 5342 [16] was chosen for Carbon Monoxide (CO) sensing. The sensor has a compact size when compared to other electrochemical sensors and the electrolyte in it is environment friendly. This sensor has very good sensitivity, can be calibrated easily, has a linear relation with CO concentration and is not affected by interfering gases. However, the response time of this sensor is almost double that of MOS type sensors.A highly accurate surface mount temperature-humidity sensor SHT 25 [17] was used. It uses I2C protocol to transmit data to the microcontroller and is pre calibrated with 1.8% RH accuracy. A standard fingertip held heart rate sensor[18] was used to record the pulse of the user to supplement the "fat burn" detection.

IV. DESIGN

A Raspberry Pi Model 3B was used as a versatile microcontroller to handle the array of sensor data values, perform real time analysis while controlling various input and output devices like LCD touchscreen, Diaphragm pump, etc in our prototype. The Raspberry Pi generates the necessary control and input/output signals for the communication, sensor board and DC diaphragm pump using its general purpose input output(GPIO) pins. A 3.5" TFT touchscreen was integrated for our prototype which uses the SPI (Serial Peripheral Interface) bus for communication with the microcontroller. To meet the power requirements of the device a set of six 3.7V 4000 mAh lithium-polymer batteries was used.The batteries were connected in 3S-2P configuration (11.1V, 8Ah). The battery pack capacity of this arrangement can run theprototype continuously for about 12 hours drawing up to 2A current constantly and with judicial use the prototype can be used for around 4-5 days when discharging from full capacity. The Li-Po battery pack takes approximately 10 hours to charge to full capacity again. The batteries are thin, light and powerful and ideal for the handheld device. Overcurrent protection circuitry for the battery was included. A balance battery charger is used to recharge the batteries thus ensuring rapid charging. A monolithic integrated circuit, LM2596 was used as a step-down switching regulator (Buck converter)which was capable of driving up to3A load with good line and load regulation. The Buck converter was set to give a constant output of 5V (to avoid fall in output voltage due to fall in battery voltage) so as to power the different components of the device. LM1117 IC was used as a low dropout voltage regulator with the dropout of 1.2V at 800mA of load current. It was used to power up the ADS 1115 ADC, SHT25 sensor and the pulse sensor by supplying 3.3V. ADS1115 was used as a precision analog-to-digital (ADC) with 16 bits of resolution. It features an on board reference and oscillator and operates from a single 3.3V power supply which is provided by LM1117 IC. The ADC sampling rate and gain was programmed to increase the resolution of the readings. A high input impedance amplifier was used to amplify the minute current output of the electrochemical sensor and give workable voltage reading at the ADC. A H-bridge motor driver was used to power a DC diaphragm pump as the microcontroller has nowhere enough power/current supply to drive a DC pump which draws 200mA (160mA more than what RPi can provide at most). The motor driver takes logic power supply (5V) as well as motor power supply (11.1V) while maintaining a common ground. The DC diaphragm air pump was used to pump out the air from the sensing chamber after sampling process. The pump is airtight and pumps out air at the rate of 0.8L/min. The values taken from the sensors were transmitted to the server system in order to implement real time graph plotting and analysis on the website using*Python HTTP requests* module. A printed circuit board was fabricated for the sensor array and other components and placed into the sensing unit.

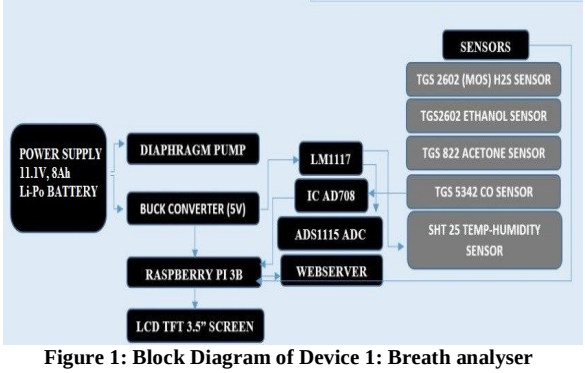

Figure 1: Block Diagram of Device 1: Breath analyser

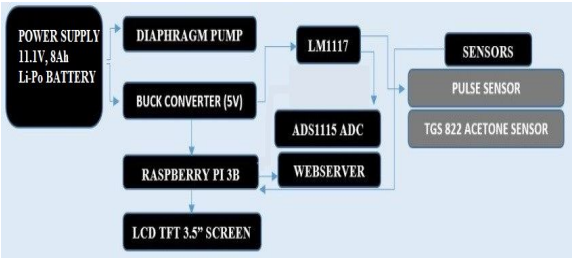

Figure 2: Block Diagram of Device 2: Diet Monitor

Rapid Prototyping 3D Printing was adopted to fabricate the device with Fused Filament Fabrication (FFF) Technology. The material used was Polylactic Acid Plastic (PLA). The 3D printer used a single all metal hot end extruder with nozzle diameter 0.4mm to print the prototype parts by G-codes on a 3-way adjustable bed with layer height approximately 0.3mm and infill density approximately 30% (Sensing unit infill was increased to 80% to maintain airtight conditions).Minimum backflow, gas sampling and drawing in of fresh air during exhaust (post sampling) were the criteria considered for the design. A brass non-return valve (NRV) was placed in the inlet pipe. The NRV stops the inlet air to flow out through it and only allows fresh charge to enter when the sampling process is over and the exhaust valve is opened. When the user breathes into the device, pressure is created after some time which shuts the NRV and blocks the passage of air in both the directions. This is the maximum capacity stage. The dimensions of the sensing unit were such that the maximum capacity of the sensing chamber was equal to the average human breath volume during forced exhalation. The diaphragm pump [19] acts like an exhaust valvealso (when the suction nozzle is perfectly sealed) thus catering to the need of airtight sampling chamber. The full assembly of the prototype is shown in Fig 3,4,5 and 6.

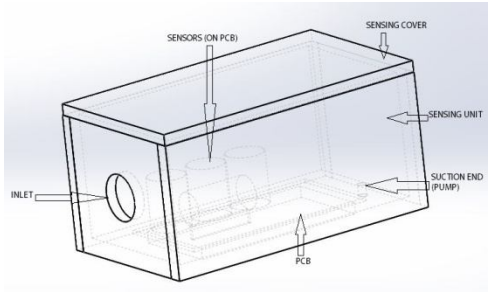

Figure3: Sensing Chamber Assembly

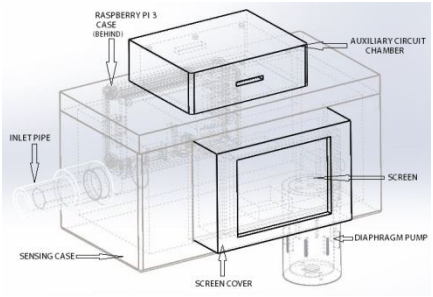

Figure 4: Sensing Case Assembly

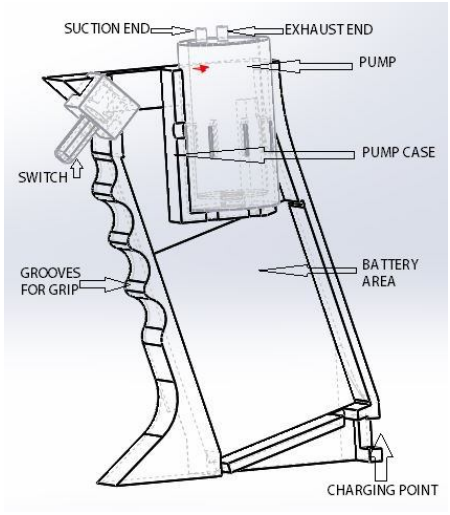

Figure 5: Handle Assembly

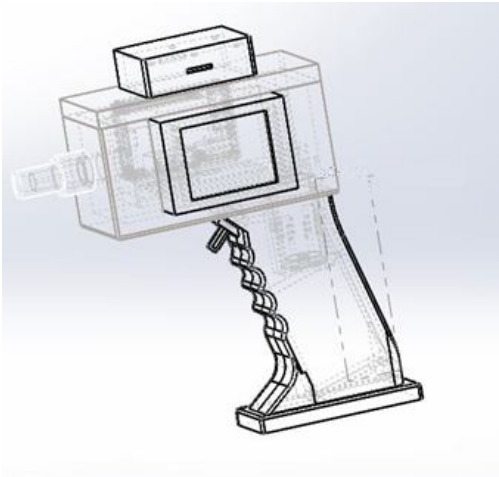

Figure 6: Full Assembly of Breathanalyzer

Figure 8: Block Diagram of breath flow

V. CALIBRATION

The sensors were calibrated to obtain a relationship between the voltage measured by the ADC and the concentration of test gas. A mass flow controller setup was used to regulate test gas and synthetic air concentrations with respect to time. At regular intervals of time increasing concentrations of test gas is passed to the sensor with periods of synthetic air being passed for the sensor to stabilize. The sensor readings are stored after which a graph was plotted as shown in Fig. 10, 12, 14 and 16. The best fit curve was chosen from the scatter plot according to regression analysis and mathematical modelling after which the equation of the curve was noted which was used to determine concentration from the voltage readings as shown in Fig 11, 13, 15 and 17.The calibration setup included synthetic air and test gas calibration cylinders, mass flow controllers to obtain required concentration, a test chamber to place the sensor and a mixing chamber as shown in Fig. 9. The

calibration was carried out inside a temperature-humidity chamber to simulate breath conditions of 90% relative humidity as the sensitivity MOS type sensors changes with change in ambient humidity. Accuracy was achievedfor the calibration by employing an Artificial Neural Network (ANN). The neural net was made to train with the sensor dataset. The weights were obtained by training the algorithm on the PC which was appended to the Graphical User Interface (GUI) code on the prototype which increased the pace of training and testing of the data. The peak voltage value formed the input node and the ppm levels formed the output node and on providing the test data to the algorithm accurate ppm levels were obtained. Fig 18 shows the impact of using ANN in error minimization.

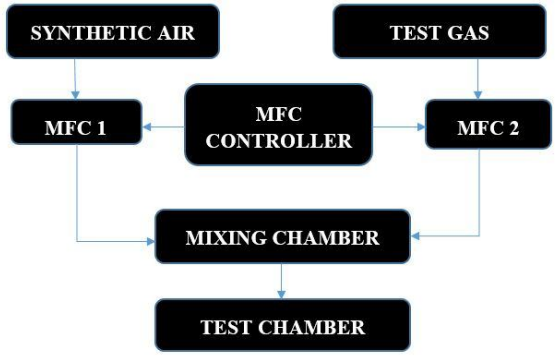

Figure 9: Schematic diagram of sensor calibration setup

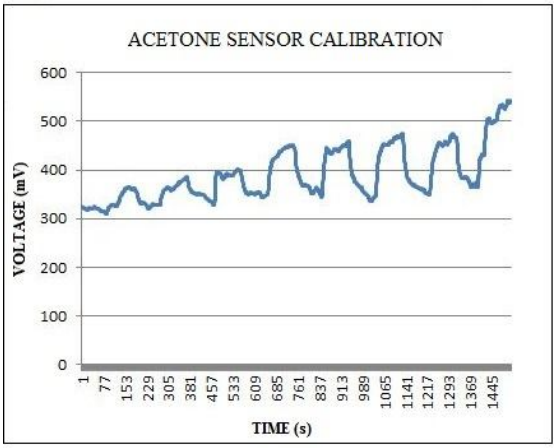

Figure 10: Voltage vs Time graph of Acetone sensor calibration

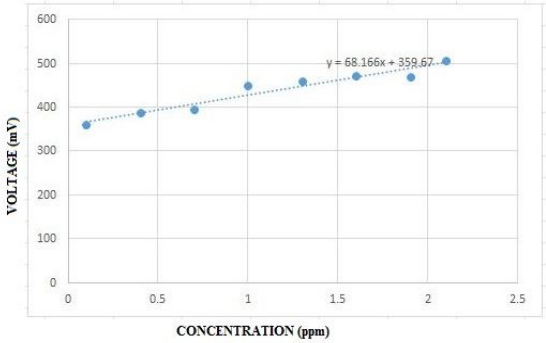

Figure 11: Linear equation generated from calibration of Acetone sensor

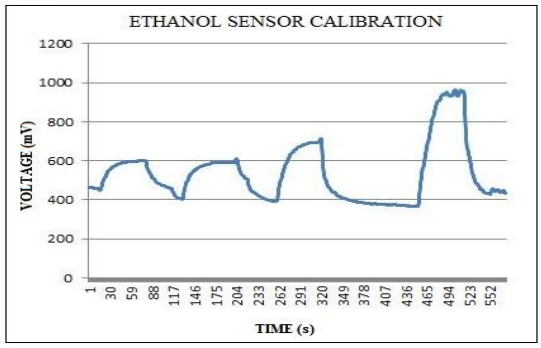

Figure 12: Voltage vs Time graph of Alcohol sensor calibration

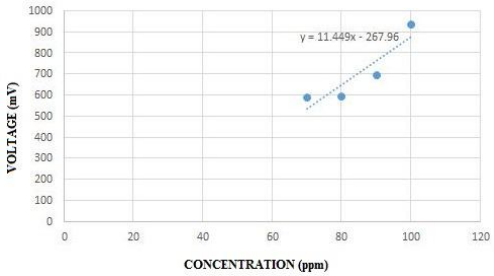

Figure 13: Linear equation generated from calibration of Alcohol sensor

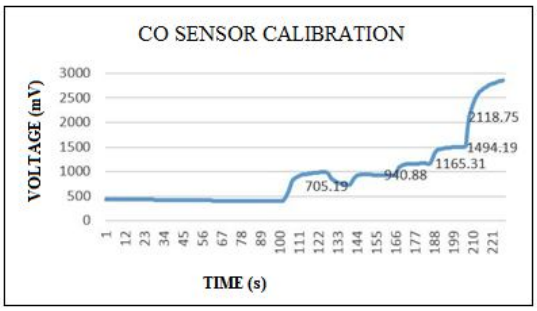

Figure 14:Voltage vs Time graph of Carbon Monoxide sensor calibration

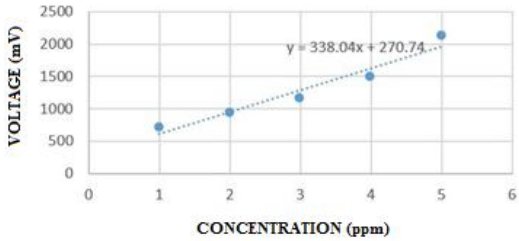

Figure 15: Linear equation generated from calibration of Carbon Monoxide sensor

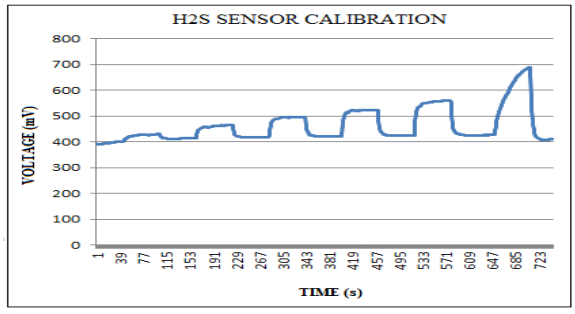

Figure 16:Voltage vs Time graph of Hydrogen Sulphide sensor calibration

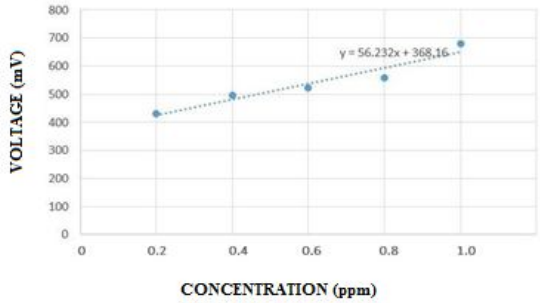

Figure 17: Linear equation generated from calibration of Hydrogen Sulphide sensor

GAS	DESIRED OUTPUT (PPM)	CALCULATED-LINEAR REGRESSION	CALCULATED-ARTIFICIAL NEURAL NETWORK	CALCULATED ERROR USING LINEAR REGRESSION	CALCULATED ERROR USING ARTIFICIAL NEURAL NETWORK
ACETONE	0.8	0.6896	0.7999	0.1161	0.0001
CO	2.3	2.06345	2.2999	0.23655	0.0001
ALCOHOL	120	118.4967	119.899	1.5033	0.101
H_2S	0.66	0.4783	0.6599	0.1817	0.0001

Figure 18: Error minimization using ANN

VI. WORKING

The schematic diagram of the flow of breath is shown in Figure 8. The microcontroller and the sensor board receive their power via a Buck converter which regulates the battery voltage to 5V. The user clicks on "Start Test" in the user interface and exhales in the inlet pipe of the device. The exhaled breath passes through the non-return valve and enters the sensing unit. The calibrated sensors take the readings which are plotted on the screen in real time. Metal oxide type gas sensors work on the principle of change in resistivity with change in gas concentration. The sensing mechanism generally includes oxidation or reduction of thesemiconductor and adsorption of the chemical species on the semiconductor which results in the electronic transfer of delocalized conduction band electrons to localized surface states. The effect of the surface phenomena is reversible one and causes significant change in the electrical resistance.With increase in gas concentration due to adsorption the conductivity of the sensor increases. This change in conductivity can be converted to voltage and measured with the help of a voltage divider circuit. For the CO sensor which is an electrochemical sensor a different approach was used to measure concentration. This sensor acts as a cell which gives different values of current for different ppm of CO gas. Since the order of current output is in nano-amperes, a high input impedance amplifier circuit was used which amplifies this output. The pulse sensor measures peaks in the infrared light bounced off the blood flow in the finger. The pulse is sampled for 20 seconds and number of peaks is calculated giving an accurate measure of the heart rate. Once the data has been sampled and results displayed the diaphragm pump is switched on via the motor driver and the sampled gas is removed while drawing in fresh air through the inlet pipe and non-return valve by generated suction. During the real time plotting process (sampling process) sensor readings are transmitted to a server via a data transmission system (Python requests). The diaphragm pump stops after a predetermined time and the sensors return back to the normal states and are ready for the next cycle.

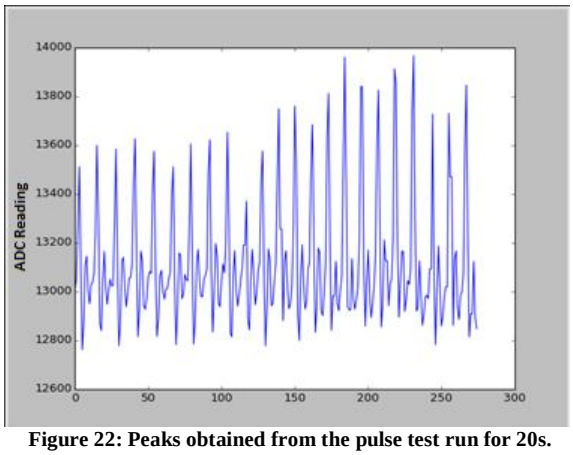

Figure 22: Peaks obtained from the pulse test run for 20s.

VII. RESULTS AND CONCLUSIONS

Results of the breath analysis test on five individuals are shown below.From the results we can conclude that our breath analysing prototype using gas sensors can be a legitimate method for disease detection, health condition and diet monitoring and can be used as a precursor to actual laboratory tests.Furthermore the application of ANN increases the accuracy of testing.

Table 1: Results for Diabetes test (Device 1)

	Reading (ppm)	Safe Value	Inference
Person 1	1.7	0.15-0.9ppm	Check for Diabetes
Person 2	0.80		Healthy
Person 3	0.47		Healthy
Person 4	0.60		Healthy
Person 5	0.62		Healthy

Table 2: Results for CO poisoning test (Device 1)

	Reading (ppm)	Safe Value	Inference
Person 1	2.3	Below 6ppm	Normal
Person 2	2.1		Normal
Person 3	1.87		Normal
Person 4	2.43		Normal
Person 5	3.0		Normal

Table 3: Results for alcohol test (Device 1)

	Reading (ppm)	Safe Value	Inference
Person 1	38	Less than 150ppm	Normal
Person 2	59		Normal
Person 3	183		Slightly Intoxicated
Person 4	77		Normal
Person 5	133		Normal

Table 4: Results for Halitosis test (Device 1)

	Reading (ppm)	Safe Value	Inference
Person 1	0.31	Less than 1ppm	Healthy
Person 2	0.22		Healthy
Person 3	0.73		Healthy
Person 4	1.12		Halitosis
Person 5	0.66		Healthy

Table 5: Results for Pulse test (Device 2)

	Age	Pulse Reading	Reference Value	Inference
Person 1	37	100.03	Check table below	Normal
Person 2	22	106.34		Normal
Person 3	35	102.75		Normal
Person 4	20	113.8		Normal
Person 5	21	108.21		Normal

Table 6: Reference heart rate range for maximum "fat burn"

AGE	HEART RATE
0-10	126-147
10-20	120-140
20-30	114-133
30-40	108-126
40-50	102-119
50-60	96-112
60-70	90-105

Table 7: Results for Fat burn test (Device 2)

	Reading (ppm)	Threshold Value	Inference
Person 1	1.8	2ppm	No fat burn
Person 2	1.66		No fat burn
Person 3	2.8		Fat Burn
Person 4	3.0		Fat Burn
Person 5	2.5		Fat Burn

Future scope: The prototype builthas a limitation of the sensors being cross sensitive to other gases like ethanol. Thus even small intake of alcohol before testing can make the results for acetone and H_2S inaccurate. This breath analyser prototype is specific to only certain diseases and conditions. New sensors capable of detecting other biomarkers and being insensitive to interfering gases can be included in the sensor array if they are capable of detecting concentrations present in human breath. For example, MOS type NO sensors, currently not in market can be developed for the purpose of detecting diseases like asthma [20] and Hydrogen gas sensorsfor the detection of gastrointestinal diseases and specific food intolerances. Currently available MOS type NO and hydrogen sensors are applicable only for industrial use and cannot detect the low concentrations of gases present in human breath.

ACKNOWLEDGEMENTS

This work was done as part of internship at Centre for Nanoscience and Engineering, Indian Institute of Science, Bangalore during the summer of 2016.

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