

Design and Development of a Smart Mobile-based Optical pH Sensor Using Paper Microfluidics for Oral Health

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Abstract: We have developed and validated a handheld optical pH sensor for monitoring saliva suitable for biomedical diagnostics. The system utilizes a cost-effective paper-based test strip, where a pH responsive dye (phenol red) was immobilized on grade one cellulose filter paper. A pH change leads to a visible colour change on the strip. This change is quantitatively detected by a custom electronic meter that minimizes interference from ambient light. The sensor accurately detects pH values ranging from 6.0 to 7.8, utilizing a 570 nm light-emitting diode that corresponds with the peak absorption of phenol red. It has 2371.4 counts per pH sensitivity, and the calibration slopes indicate a limit of detection of 0.166 pH units for the green channel and 0.244 pH units for the red channel. The data read by the electronic circuit was sent to the Android-based smartphone for the display of the pH values using the USB protocol. The results are rapid (60 seconds) and a strong linear relationship of 0.96% accuracy between actual and predicted values. Field validation against the laboratory pH meters demonstrated a strong correlation. This simple, low-cost, eco-friendly platform is ideal for mass screening and self-monitoring, as a quick, non-invasive pH assessment is required.

Keywords: Handheld optical detection, pH sensor, Paper microfluidics, Point of care diagnostics, Smart mobile.

Introduction

With the current trend of changing lifestyle and diet patterns, issues related to oral health have risen significantly worldwide. According to the World Health Organisation (WHO), it is estimated that more than 3.7 billion individuals are affected by oral diseases [18, 24]. Among these, the major concerns are dental caries, the most prevalent condition worldwide [17]. The 90-95% of the population is affected by periodontal disease, whereas dental caries impacts 60-80% of children in India. [12]. Periodontitis, gingivitis, and other conditions, such as acid reflux or gastroesophageal reflux disease, can affect the pH of saliva, regardless of age or gender. Oral health management in developing countries is poorly established or

perceived as a burden, and access to preventive and therapeutic dental services remains limited [24]. Recent data reveals troubling patterns: 71% of Indians brush their teeth only once a day, while 15% brush less than once a day, rising to 23% in rural areas [2, 22]. These basic oral hygiene gaps translate into substantial health consequences. Oral health has often been viewed as separate from overall health, but this perception is shifting. While dental professionals once focused on treating oral diseases, modern dentistry now emphasises prevention and highlights the vital connection between oral hygiene and general health. Recognising this connection is critical in achieving a healthy smile and a healthier body.

The healthy pH range is from 6.7 to 7.4 pH, but can vary from 6.2 to 7.6 pH, making it relatively neutral [11]. Individuals suffering from oral diseases, patients with radiotherapy-related caries, and those with gastroesophageal reflux disease have a lower pH over long periods [5, 7, 14]. So, regular saliva pH evaluation is important. Studies have shown that different bacteria thrive at specific pH levels. *Porphyromonas gingivalis* grows optimally in environments with *Prevotella intermedia*, which flourishes at a pH of 6.5 to 7.0, while *Fusobacterium nucleatum* favours a pH of 5.5 to 7.0 [3, 20]. Both organisms thrive within these specific pH ranges, highlighting the importance of maintaining a pH balance in promoting oral health. Traditional methods of measuring saliva pH, such as pH strips, are ineffective because their accuracy can be less precise than digital meters, and colour differences may lead to reading errors. Digital pH meters require more equipment and expertise than pH strips [21]. Clinical evaluation of saliva pH is expensive and time-consuming.

Multiple techniques are suggested for quantifying saliva, utilising a flexible, laser-fabricated nanographene-based sensor to assess pH levels in saliva. The graphene oxide (GO) sensing layer on a laser-induced graphene (LIG) electrode in this capacitive pH sensor is composed of a polyimide (PI) sheet, which serves as both the substrate and precursor for LIG [1]. Additionally, quick, colour-based salivary pH detection is made possible via a 3D-printed hydrogel sensor. A non-toxic hydrogel ink is used to create the sensor, which is then 3D printed into a matrix of $5 \times 5 \times 1$ mm sensing pads [13]. Using a smartphone, the portable device can quickly determine oral pH using a non-toxic vegetable-based colourimetric indicator [2]. All these methods compromise on both cost and accuracy and are not user-friendly; therefore, any patient cannot use these techniques without prior knowledge and expertise [16]. Technology is an emerging method of diagnostic and pre-diagnostic techniques. This work involves using a pH-sensitive dye embedded in a paper-based strip and scanning the dye's colour variation using a colour sensor and a self-developed mobile application to identify the pH.

Materials and methods

Filter paper (WhatmanTM standard grade 1 paper), polyvinyl alcohol (Cat. No 363138-500G), sodium dihydrogen phosphate monohydrate (Cat. No 1.93224.0561-500G), and dibasic monohydrogen phosphate (Cat. No 1.19753) were purchased from Sigma Aldrich Ltd, India. Phenol red ($C_{19}H_{14}O_5S$) (Cat. No 143-74-8) is purchased from Fisher Scientific Ltd. Lamination sheets with two and three layers were purchased from Amazon India Ltd. Distilled water was provided by Merck Millipore Milli-QTM Ltd direct water purification equipment. We bought a digital pH meter (Orion STARTM A214) from Thermo ScientificTM Ltd. The pH sensor strip was cut using a Cricut (model Explore One), and the strip's base was cut using an Epilogue Laser cutter. A thermal printer (Mt800) was sourced from Amazon India Ltd, and an Ultimaker 2+ was used. Siemens Digital Industries Software's Solid Edge 2024 was used to create a 3D CAD model of the mechanical chassis. The mechanical chassis was printed using Ultimaker 2+. Easy EDA was used for circuit and printed circuit

board design, and CorelDRAW was used to create the cut lines for a laser cutter. The strip design was created using the Cricut Design Space program.

Block diagram of the developed optical pH system

The block diagram of the developed pH system is shown in Fig. 1. We have used a light-emitting diode (LED) with a specific wavelength that matches the peak absorption of phenol red to increase accuracy. The LED is directly on top of the strip's reference and detection zones, and the colour sensor is directly at the bottom of the zones. This setup would increase accuracy by reducing interference. Then the colour sensors' raw data is fed into the microcontroller. Here, we have used an ESP32-C3 [23] to process and optimise data for transfer through USB to a mobile device. Another advantage of the system is that the required power is drawn from the mobile. Hence, the entire system is compact and easily accessible. From the mobile application, data can be sent or saved to Google Sheets or on the internet, making it a more cost-effective solution. It can detect even minute pH variations [9], crucial for identifying early signs of oral health issues. Additionally, using a colour sensor in a constant light source yields consistent results, making it more reliable [7, 9, 19].

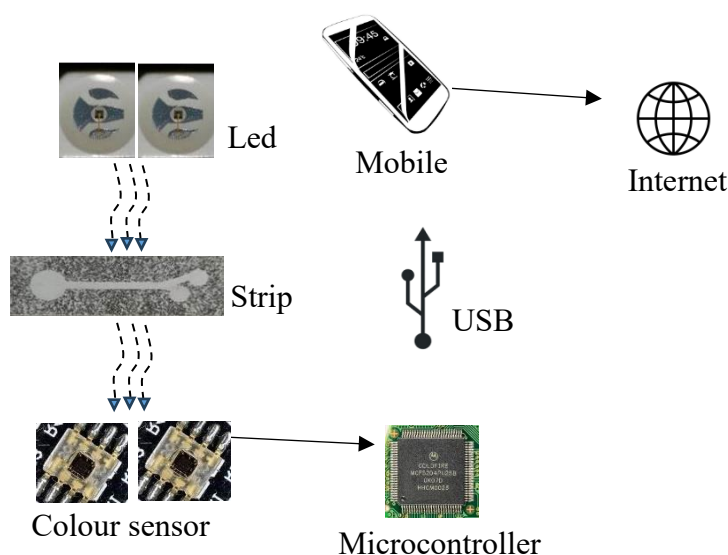


Fig. 1 Block diagram of the developed point of care testing detection system

Strip design of chemical sensor

The chemical sensor strip consists of 4 layers, as shown in Fig. 2B. The first layer is cut using the programmable paper cutter (Cricut Explore One) according to the design that has been developed. The two holes where the LED would project light are cut, as we want to reduce interference as much as possible. An additional hole is also created at the bottom for users to insert the sample. This layer is made of a lamination sheet (one-sided), which is used to protect the strip from external dust.

The second layer is the test strip, which consists of Whatman™ standard grade 1 paper, where the sample and analyte are placed and measured. The reason for choosing Whatman™ standard grade 1 paper is that it is generally used and available, and the retention size of the paper is 11 µm, which can comfortably clear all the debris and waste from the sample. Firstly, we printed the thermal resin on the Whatman™ standard grade 1 paper using a thermal printer with the specific design (2 mm and 3 mm circles for reference and detection zone, an 8 mm circle for the sample and a 2 × 45 mm rectangle for the sample to flow). The printed pattern paper was

placed on a hotplate at 100°C for 15 minutes, causing the ink to melt and sink into the paper, creating hydrophobic regions, as shown in Fig. 2A. Consequently, the sample only flows through the specifically designed path. The third layer is a double-sided lamination sheet. Its main purpose is to attach the paper to the base, so it has cuts only at the detection and reference zone (2 mm and 2.5 mm circle cut). The 4th or base layer is a 2 mm thick acrylic sheet. This specific material is used as it is transparent and easily available at a low cost. The transparency would allow the sensor to read data freely, rather than being restricted, which would be more suitable for this kind of application, as the whole area of colour changes must be captured. The base also has two holes at the reference and detection zone (2 mm circle cuts). The acrylic base was cut using a laser printer.

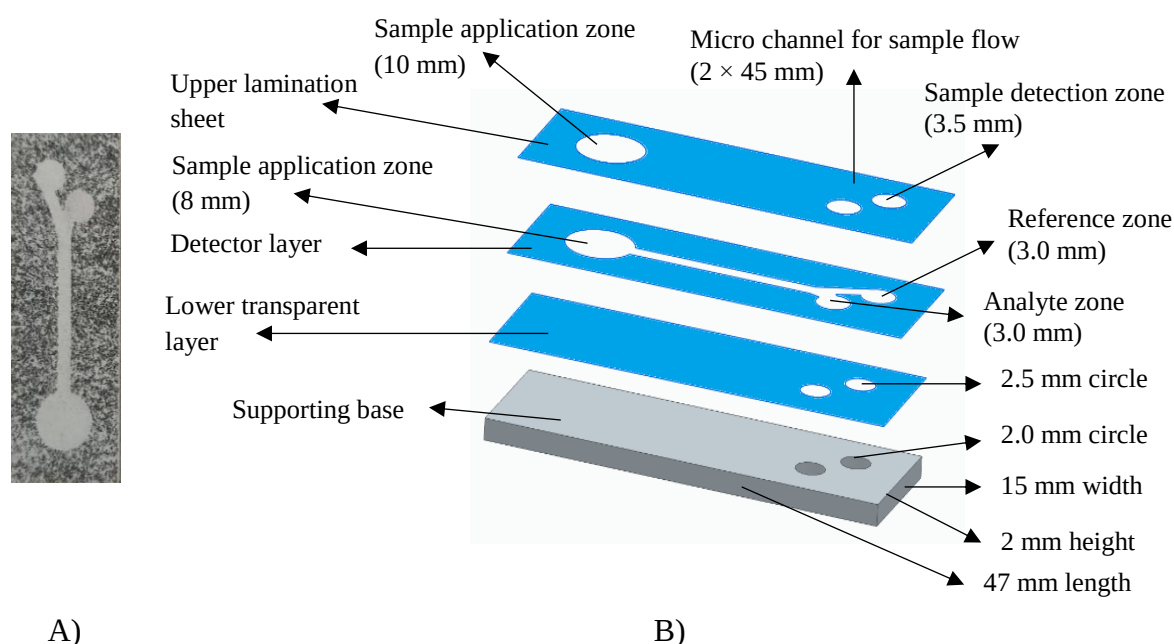


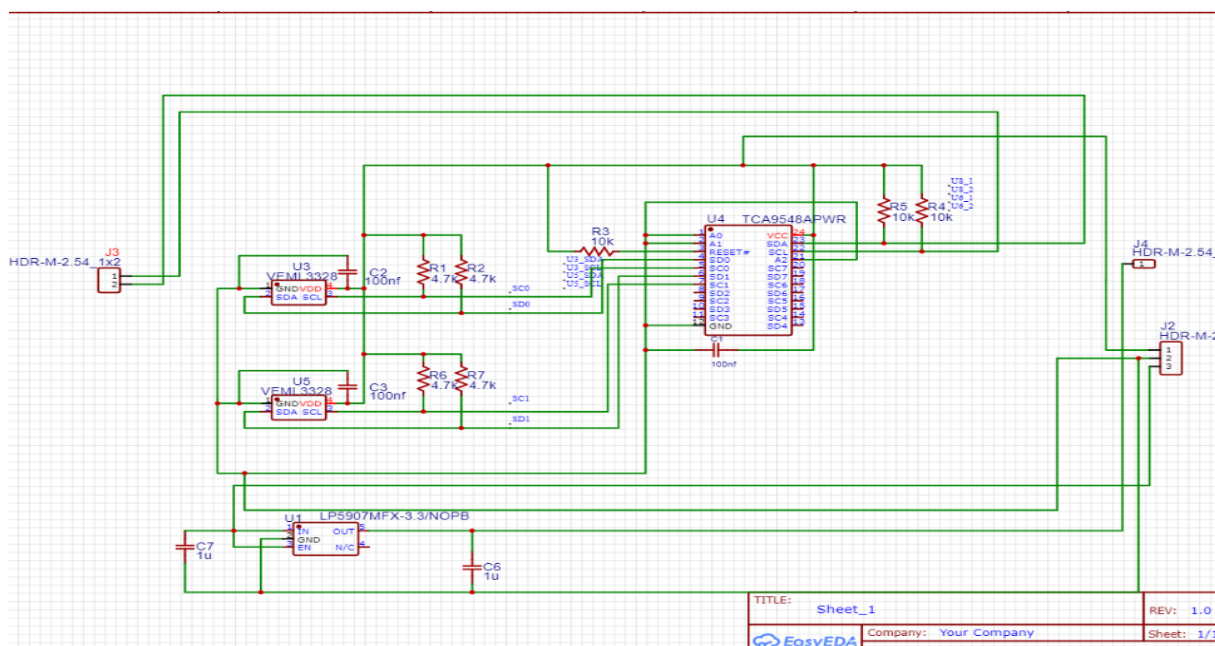
Fig. 2 Before the immobilisation of the analyte (A), and different layers of the strip design (B)

Circuit design of handheld instrument

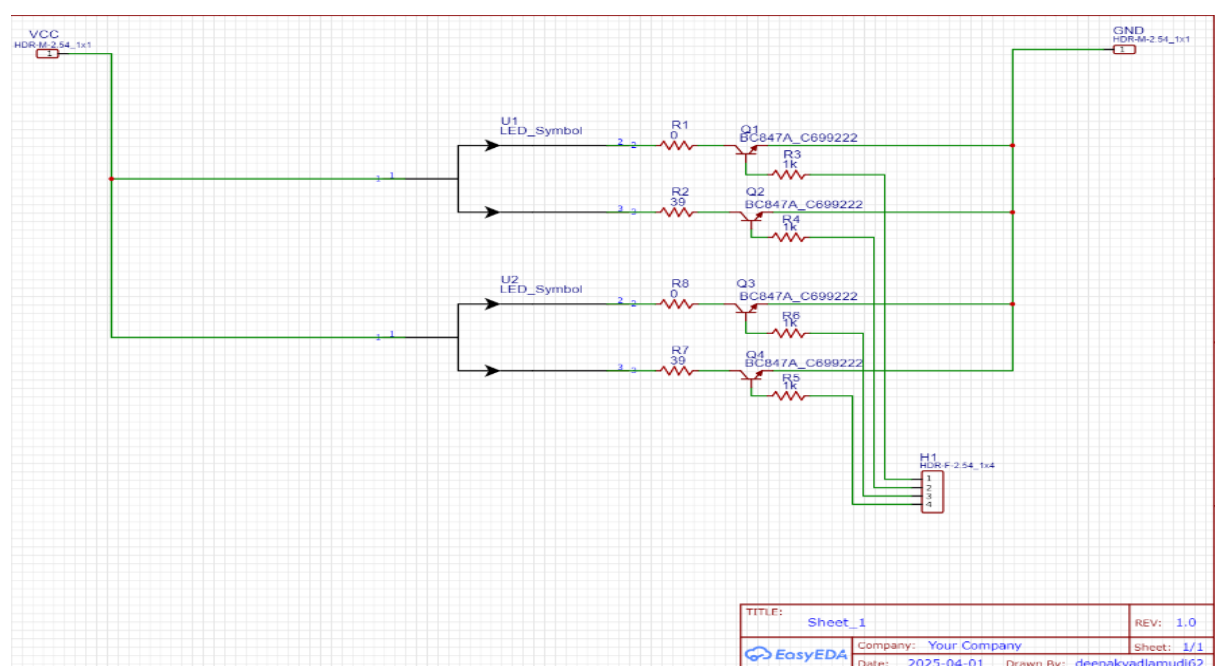
The electronic architecture of the instrument was created to be compact, stable, and benign for data processing. The circuit design used for the development of the device is shown in Fig. 3. All the necessary components (light source, signal multiplexer, colour sensor, voltage regulator, and communication circuitry) were placed on a double-sided printed circuit board to create a lightweight, inexpensive configuration that could be used in portable diagnostics. The detection principle is based on colourimetric sensing, where changes in saliva pH result in an observable colour change in the test strip's analyte region. This colour change corresponds directly to biochemical reactions in the pH-sensitive layer of the paper substrate.

Flowchart of the software developed for the handheld instrument

To quantitatively measure the optical changes, the system employs two light-emitting diodes (LEDs), set to 430 nm and 570 nm, which were purposely selected based on the chromophore absorption of the pH indicator, phenol red. Reflected light from the test zone is detected by a small red-green-blue (RGB) colour sensor (VEML 3328), which has a high spectral sensitivity with low noise. It is also affordable, and the size is smaller, which fits into the design. The sensor converts its output to a digital signal. It sends the information using the I²C communication interface to an ESP32-C3 microcontroller by Seeed Studio XIAO for data acquisition, pre-processing, and data transmission [23].



A)



B)

Fig. 3 Circuit diagram of the colour sensor design (A), and circuit diagram of LED printed circuit board design (B)

This controller is specifically used due to its high speed, ability to draw power from mobile devices, and support for protocols such as I²C, Bluetooth, Wi-Fi, as well as its compact size (21 × 17.5 mm). The voltage regulator is included to stabilise voltage fluctuations because voltage level variation can affect optical readings. The ideal working voltage is 3.7 V, but the system draws 5 V from the mobile, so we used a voltage regulator to optimise the voltage. The multiplexer is used because we are using two colour sensors with the same address. The sensor output is scanned over 500 ms with a moving-average filter. A calibration curve guides pH calculation, which is transmitted to a mobile device via USB for display and storage,

with options for integration with Google Sheets. The flowchart of the microcontroller program in Fig. 4A illustrates how the program calculates the pH from saliva samples and transfers the data via USB 3.0 or Type-C.

The flowchart in Fig. 4B details the sequence of the pH data acquisition and saving process. This data collection takes place over 60 seconds, during which the system continuously reads pH data from the controller via a USB connection. Once the reading has been taken, the system displays the recorded pH value to the user. Once the user has viewed the collected data, they can press the “SAVE” button to save the pH data into a Google Sheets document. This action ends pH data acquisition.

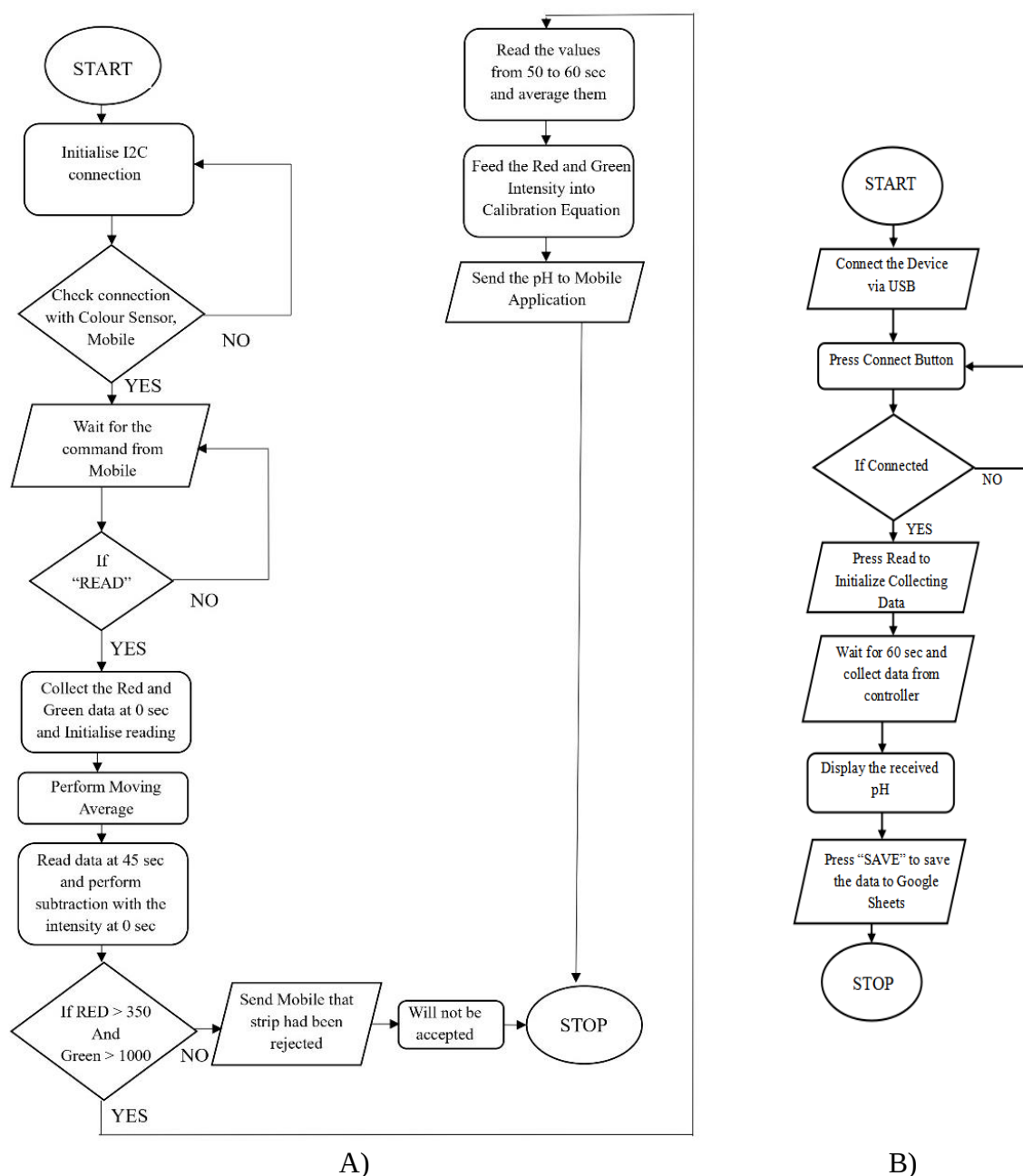


Fig. 4 Flowchart illustrating the working process of the developed system

The ESP32-C3 initialises the I²C connection and checks for established communication with the colour sensor and the mobile device to begin this process. After this connection, the microcontroller waits for the “READ” from the mobile app. When the mobile app sends “READ”, the microcontroller immediately reads the first red and green intensity data in 0 seconds. The microcontroller then initiates a continuous moving average calculation that remains active until the 60 second mark. The microcontroller will read in data for 45 seconds, then read the intensities AGAIN at 45 seconds, then subtract the intensity data from the original 0 second reading to adjust for background light. Then, the microcontroller performs a quality check: if the processed red intensity is less than 350 or the green intensity is less than 1000, the strip is rejected (a rejection message is sent to the mobile, and the process stops). If the strip is acceptable, the microcontroller continues to read intensity information and averages the data between 45 - 60 seconds. Finally, the microcontroller will input the red and green intensity information into the previously established calibration equation, determining the final pH value. Then, it will send the information to the mobile application for display and terminate the process.

Mechanical chassis design of handheld instrument

The chassis of the handheld device was designed using SolidWorks software (Dassault Systèmes Ltd), detailed design views are provided in Fig. 5. The final dimensions are $38 \times 30 \times 31.5$ mm. The chassis will help protect the internal electronics from damage and interference from the surrounding environment. Black acrylonitrile-butadiene-styrene (ABS) is the chassis material selected for this study. The black ABS was chosen because it minimises the influence of external ambient light on the optical pH sensing mechanisms and measurements. The decision to manufacture the chassis using 3D printing via mould manufacturing was made for cost savings and time efficiency, allowing the custom enclosure to be prototyped and produced efficiently. The final 3D printed enclosure helped maintain the precise alignment of the sensing strip with the internal optical components.

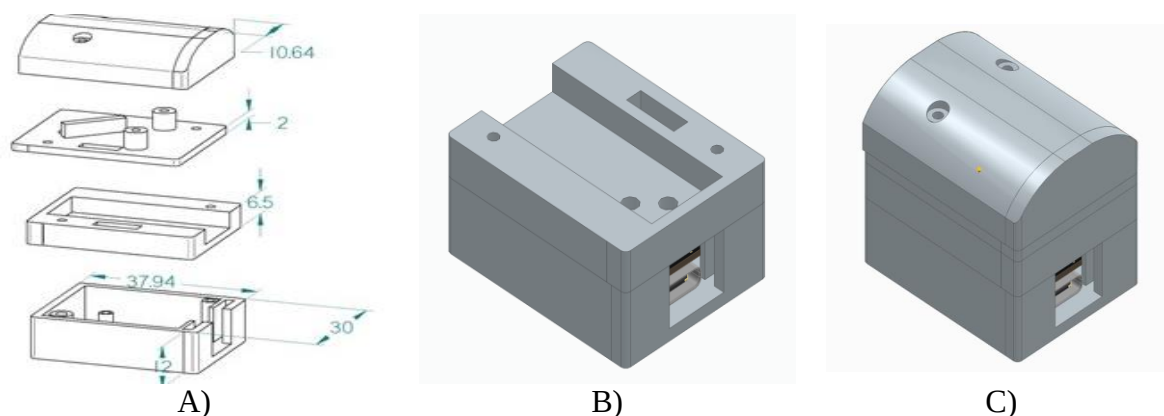


Fig. 5 3D CAD model of the developed detection system (all dimensions are in mm) (A); second layer of the mechanical chassis, showing internal components (B); LED printed circuit board along with the top part of the mechanical chassis enclosing optical and control units (C).

Mobile application development

The app interface is shown in Fig. 6. The MIT app inventor code implements the application logic for handling USB serial communication [25], reading data and integrating an external webpage.

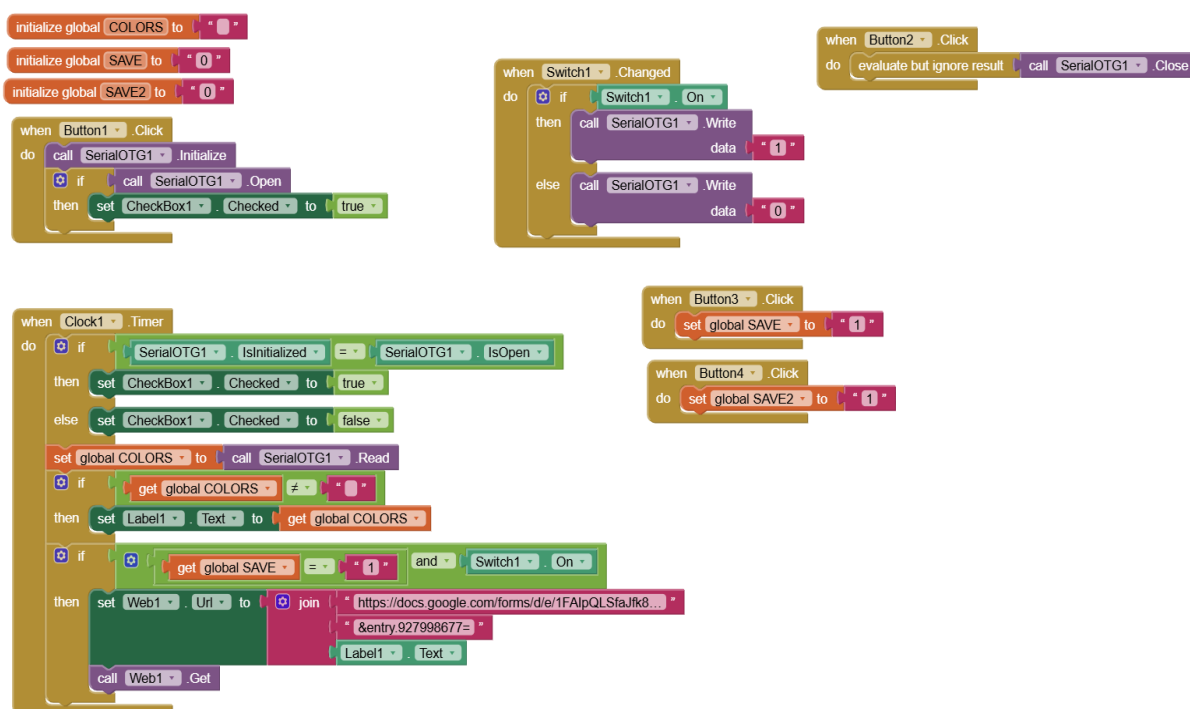


Fig. 6 Block codes of mobile application (MIT app inventor)

The app shown in Fig. 6 utilises the SerialOTG extension, available on GitHub, to manage USB communication. This was initialised in the Button 1 click method, which attempts to open the serial connection and sets Checkbox 1 to checked if successful. The Clock 1 timer block runs continuously, checking if SerialOTG 1 is open properly, which updates Checkbox 1 accordingly. Once connected, the timer initiates a Read operation from the serial port and saves all received data in the global variable COLORS (presumed pH value). The data is also displayed on Label 1. Most importantly, if the global SAVE variable is set to 1 (as in the Button 3 click event), the data is concatenated with the predetermined Google Forms URL and sent via Web 1. Get submitting the value to Google Sheets. Button 2 handles closing the serial connection. Switch 1 handles sending a digital “1” (On) or “0” (Off) command via serial write to the controller.

Results and discussion

Strip development of pH sensor

The chemicals used for the fabrication of the chemical sensor are polyvinyl alcohol, phenol red, sodium phosphate buffers, hydrochloric acid (HCl), and sodium hydroxide (NaOH). A change in the availability of H^+ ions causes the equilibrium to shift towards hydroxypinacolone retinoate (HPR) of PR^- , which changes the colour of phenol red to yellow and red, respectively. The colour changes caused by the phenol red are caused by an acid-base indicator reaction, as described in Eq. (1):



A design was comparatively better than smartphone-based detection, improving the limit of detection and sensitivity while reducing interference levels. The entire process is illustrated in Fig. 7. It is straightforward, so no prior expertise is required.

The first step in making the analyte is to make the polyvinyl alcohol solution, reason using polyvinyl alcohol is to entrap the analyte so that it stays in place and also attaches to the Whatman™ standard grade 1 paper, make polyvinyl alcohol of 0.1 mM of 100 ml, mix 0.1 g of phenol red in 14.2 ml of 0.02 mM NaOH, mix both solutions, and dilute to 250 ml with deionised water. To adjust the pH to 7, which we used as the base pH, mix the appropriate amounts of HCl or NaOH. Finally, we added sodium dihydrogen phosphate monohydrate and dibasic monohydrogen phosphate, which are used to create a buffer solution (0.05 mM).

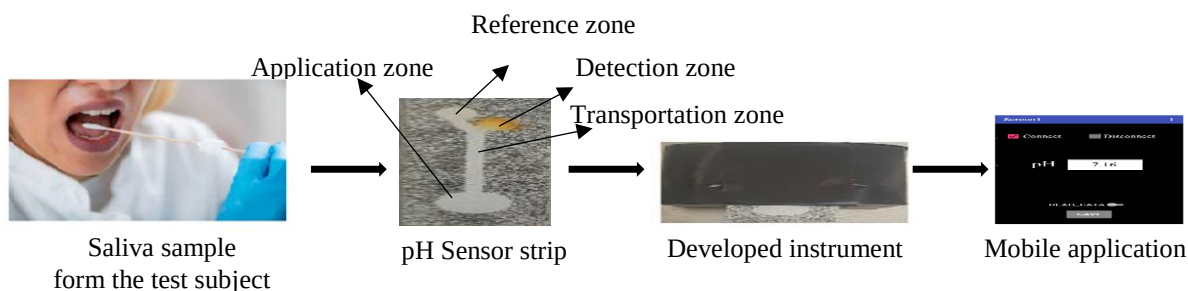


Fig. 7 Procedure to measure the pH of the user using the developed detection system

After creating the blank test strip, the simple strip was heated for ten minutes at 45°C to eliminate all the moisture. Then 2 µl of the stock solution was added to the detection zone and heated for 5 minutes. More 2 µl were added, totalling 4 µl of analyte solution. This is done because the area of the detection zone is small, preventing the analyte from overflowing. The quantity was also fixed after trying multiple samples. The mixture was heated for 5 minutes and then cooled to room temperature. As seen in Fig. 8, the strips' colour changed when they interacted with salivary pH.

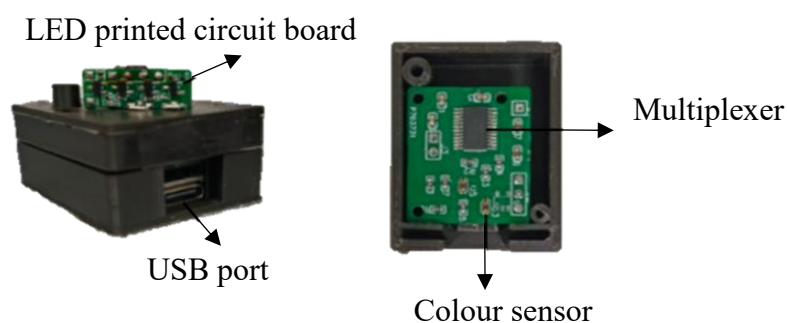


Fig. 8 Internal view of important electronic components

Development of handheld instrument

Fig. 8 displays the internal view of the instrument. This device senses the intensity of colour on paper using the principles of transmission spectroscopy. The detection zone is positioned between an RGB colour sensor and a dual-wavelength LED light source at 430 nm and 570 nm. To keep the device less expensive than commercial alternatives, such as pH meters, the sensor was chosen for its availability and cost-effectiveness. Although its sensitivity is lower than that of sensors in mobile cameras, it still meets the necessary specifications, though this may affect calibration sensitivity. Data from the colour sensor is collected by an ESP32-C3 microcontroller, which calculates pH based on a stored calibration equation. The results are displayed via an app and USB communication on a mobile device, as shown in Fig. 7. The dimensions are 38 × 30 × 31.5 mm, potentially reducing the size by up to 20% through the

use of a moulded chassis. The device operates independently, unaffected by external lighting, specific light sources, or operating systems, and can be used as a standalone unit. Since no commercial equivalent exists, a direct comparison with similar techniques was impossible. However, our device is calibrated against standard commercial pH meters.

pH sensor characterisation

Saliva with a different pH (adjusted by adding acid or base to change the saliva pH) was applied at the application zone to test the device and pH sensor shown in Fig. 9. The saliva then travelled through the paper using capillary action. In this way, the detection zone remained unchanged, unaffected by foam and other remnants of saliva. Upon contact with the analyte, the transmitted light intensity changed proportionally to the pH of the saliva and the time elapsed.

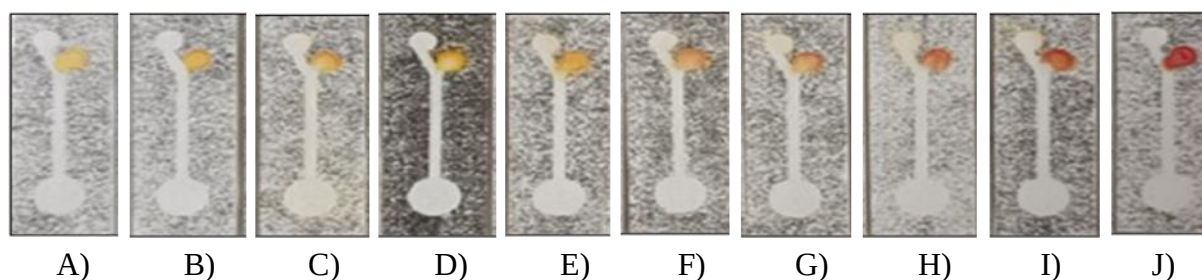


Fig. 9 Frontal view of the strips showing colour changes concerning the pH of saliva samples: 6.0 (A), 6.2 (B), 6.4 (C), 6.6(D), 6.8 (E), 7.0 (F), 7.2 (G), 7.4 (H), 7.6 (I), 7.8 (J).

Fig. 10 shows the response curve of the pH sensor strip across various pH ranges (from 6.0 to 7.8) for two colours (green and red) with intensity values from the colour sensor. These values are obtained by subtracting the reference values of the plain saliva from the reference zone and the original colour sensor values of the respective colours. The response curve of the pH sensor is the 60 second test results are shown in Fig. 10. The graph displays three main regions: the first corresponds to the sample insertion period (ranging from 0 seconds to point x), the second denotes the sample absorption duration (spanning from point x to point y), and the third reflects the pH sensor's response (from point y to conclusion). The observed drift in sensor readings following the sample's dispersion throughout the detection region [6], another issue is that the time it takes for saliva to travel to the detection zone differs when the viscosity of the saliva varies among donors [10], and could be due to the filter paper used whose transparency can vary due to fiber packing density, surface finish. Additionally, an error in the measurement of the micropipette can occur [26], which could be attributed to the filter paper used, whose transparency can vary due to fiber packing density and surface finish [27]. This could be because of the pressure differential between the transport and detection layers.

To address these potential issues, the gadget was configured to wait 45 seconds before calculating the average readings of the previous 15 seconds, which is when the pH sensor typically saturates. This was done by observing the change in intensity of transmitted light when reading is started, i.e., 0 seconds, and the values at 45 seconds. These configurations also helped determine whether the user had refused the test displayed on the mobile device due to entering a reduced sample volume. The configurations reduced errors because the sensors relied on pH variations to function. Furthermore, a 10 point average of consecutive data given at 0.5 second intervals was used for sensor measurements. The calibration curve for the pH sensor demonstrates its sensitivity to changes in pH. The limit of detection was calculated using the sensor noise obtained from this curve. Given these circumstances, a sensor response time of 60 seconds was chosen. Depending on the saliva flow rate at the time of application, there was

a 40-50 seconds lag between sample administration and detection. Eq. (2) for green and Eq. (3) for red follows the four parameter sigmoidal curve

$$Y_1 = 1934 + \frac{8712 - 1934}{1 + \left(\frac{X_1}{7.11389}\right)^{9.94931}} \quad (2)$$

$$Y_2 = 990 + \frac{2248 - 990}{1 + \left(\frac{X_1}{6.43047}\right)^{9.49906}} \quad (3)$$

where X_1 is pH; Y_1 is green colour intensity change; Y_2 is red colour intensity change.

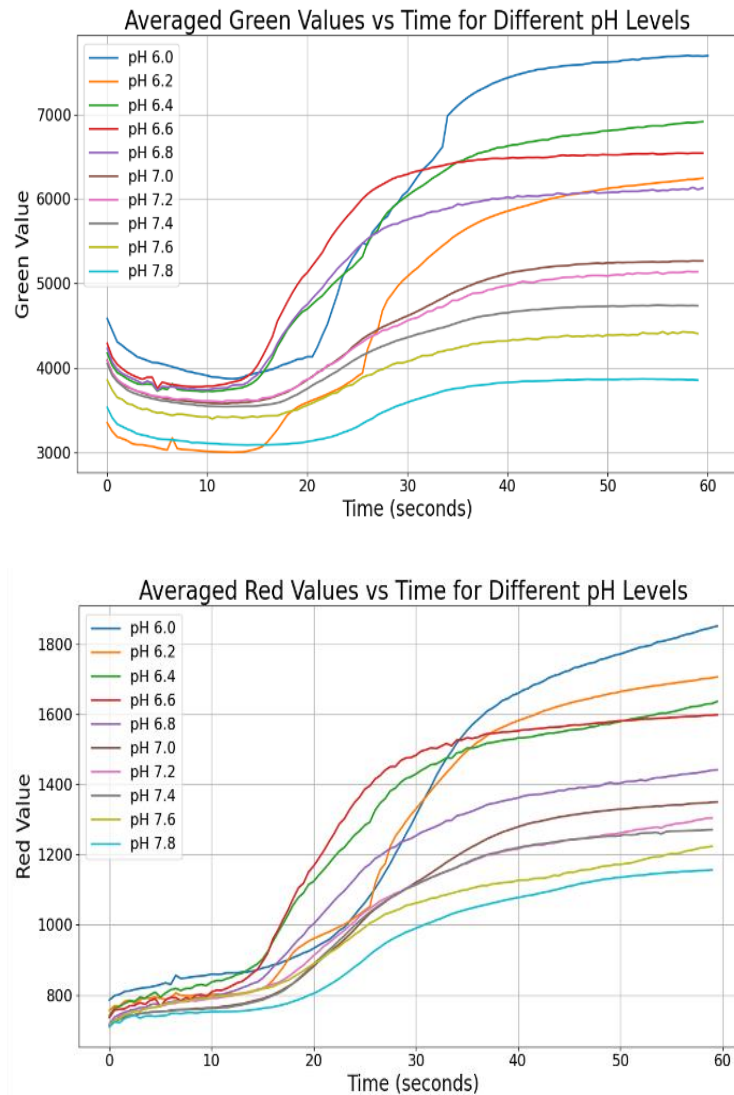


Fig. 10 Time response curve of the pH sensor strip plotted for different pH vs sensor output for 60 second time intervals: x represents the time in seconds, y represents the light intensity of green and red, respectively (pH 6.0 to 7.8).

These observations indicate that the system provides temporal stability and adaptive accuracy in saliva samples, despite biological and physical variability. The identified lag phase, which allows for the transport and stabilisation of the fluid in the detection zone, means that acquiring a consistent sample volume and viscosity is crucial for obtaining reproducible readings. The automated averaging condition and the 45 seconds delay condition effectively

reduced the effect of section variations, improving signal uniformity and repeatability of the sensor. The calibration curves (Fig. 11) confirm that the optical readout mechanism is reliable for quantitative colourimetric analysis. These data demonstrate that the system's rapid response and calibration behaviour are suitable for real-time pH sensing, with the potential for incorporation into portable diagnostics and continuous monitoring applications.

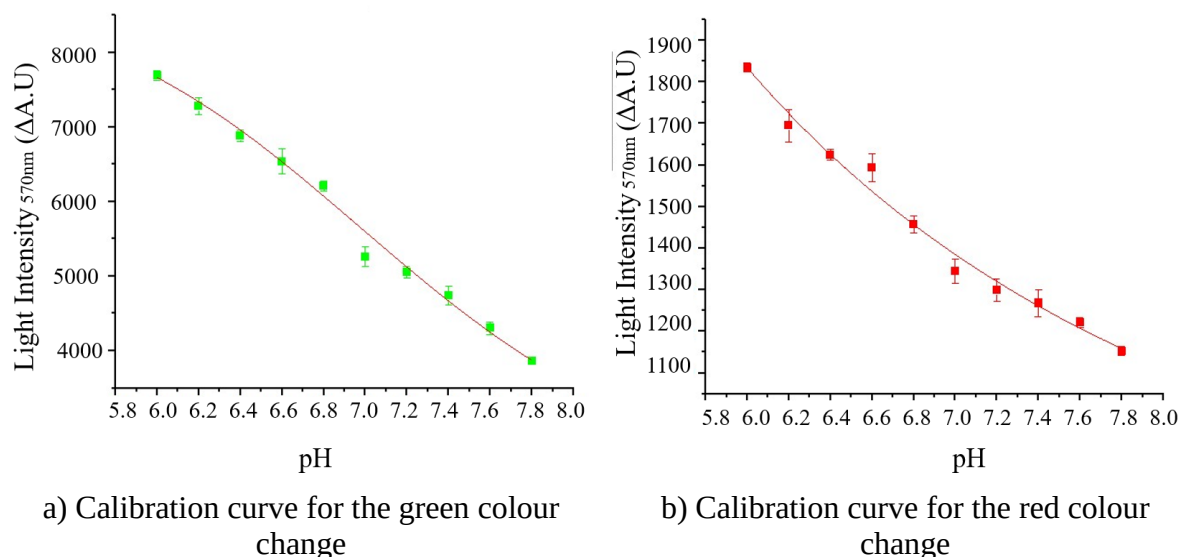


Fig. 11 The calibration curve shows the readings taken over 10 seconds from 5 samples with the same pH, averaging and plotting the values

The limit of detection of the pH sensor was derived from the equations of the calibration curves at pH 0.166 for the response from the green intensity and at pH 0.244 for the red intensity response, each at a signal to noise ratio of 56.7 for the green and 36.7 for the red channels. The calibration curves demonstrated a high level of linearity throughout the pH operational range, confirming that the optical detection system is accurate. Sensor sensitivity was determined to be 2371.4 intensity counts per pH unit for the green channel and 464.2 intensity counts per pH unit for the red channel over a measurable pH detection range of 6.0 to 7.8, which is physiologically relevant for human saliva samples.

The developed system demonstrated remarkable resistance to external lighting interference (ambient light), wind exposure, and temperature variations, thanks to its enclosed optical path and internal reference algorithm, which provides an advantage over imaging-based or open field optical devices. Additionally, the device eliminates variability caused by operational differences in camera resolution, light source intensity, or mobile operating systems by utilising a fixed illumination source and a fixed optical distance. This reliability demonstrates that the proposed platform is robust and repeatable for field and portable biochemical sensing applications.

Validation of pH sensors

The relative standard deviation in sensor readings, which indicates the system repeatability and reliability, remained consistent across multiple trials conducted in identical and distinct test strips over various time intervals. The relative standard deviation stability reflects the robustness of the developed sensing platform. Nonetheless, minor variations occur due to the manual fabrication process, such as inconsistencies in test strip preparation, uneven solution loading or fluctuations in the quantity of saliva applied during testing. These factors introduce

a degree of variability expected in manually fabricated biosensing systems. However, these sources of error can be substantially reduced through the adoption of automation. This would further enhance the reproducibility and precision of the system (Fig. 12).

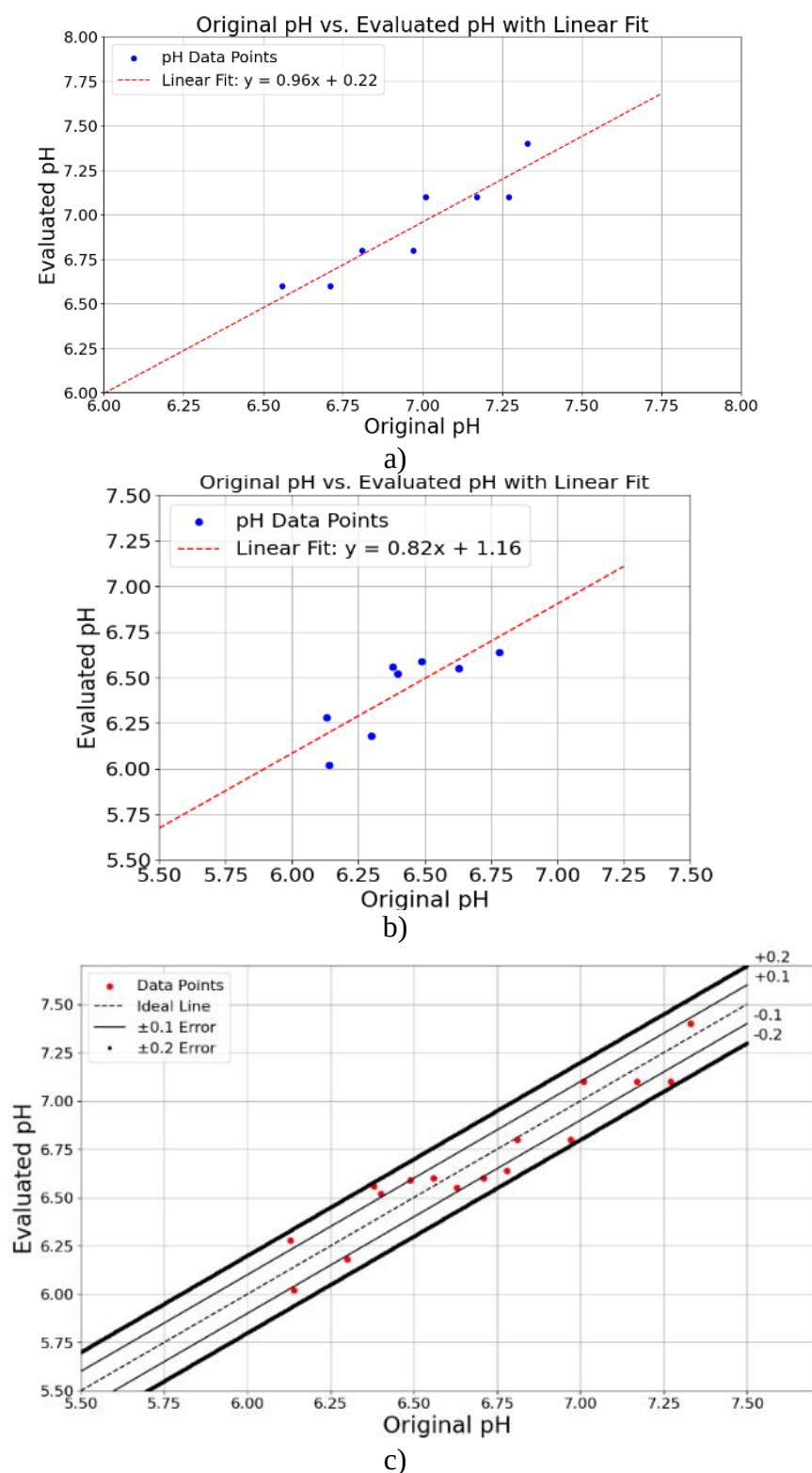


Fig. 12 Statistical correlation between original pH vs the evaluated pH for healthy patients: before eating lunch (A), after lunch (B), and pH accuracy evaluation with ± 0.1 and ± 0.2 tolerance bands (C)

Fig. 12A shows that the mean absolute error was pH 0.09, indicating the system's accuracy and a low discrepancy between predicted and reference values. The R squared value of 0.864 indicates a strong linear fit, demonstrating that the sensor is calibrated and predictable. pH readings taken from subjects displayed physiological pH values for saliva (6.5-7.5), indicating normal oral health with no increased risk of oral diseases or an imbalance in acidity.

As shown in Fig. 12B, after eating, the mean absolute error increased slightly to pH 0.126 with an R squared value of 0.760, indicating a moderate decrease in correlation due to the influence of natural biological variation. This decrease is expected, as salivary pH typically shifts to an acidic state after eating due to the metabolic actions of oral bacteria fermenting dietary sugars and starches to produce organic acids, such as lactic acid. The resulting organic acids temporarily reduce the saliva's pH while establishing conditions that promote the growth of bacteria. Salivary components, especially bicarbonate ions, act as buffers to neutralise this acidity. Still, buffering occurs over a timed process that happens mostly within 30-60 minutes after the meal, consistent with findings by other researchers [15]. In addition, salivary output increases postprandially, releasing digestive enzymes (e.g., amylase) that alter the biochemical composition and pH stability of saliva, as well as other bioactive components.

As displayed in Fig. 12C, the analysed changes in pH occur within the limit of ± 0.2 , supporting the conclusion that all measured values comply with the realistic tolerances of deviation. The small error margin demonstrates the device's accuracy, repeatability, reliability, and viability under differing physiological and environmental conditions. The fact that the measured pH and the reference pH agree so closely supports the reliability of the optical detection system and implies its potential for long-term use and monitoring. Therefore, overall, the results confirm that the pH sensor and system built perform comparably to typical commercial glucometers and can thus be recommended for clinical diagnostics and point of care usage.

In addition, recent studies focusing on optical biosensors have highlighted the influence of ambient light as a major source of interference, which compromises signal accuracy in open or uncontrolled environments. To limit this, the developed device was tested under diverse lighting conditions. Interestingly, the results demonstrated that the system's performance remained largely unaffected by ambient light fluctuations, as the 3D printed chassis completely encloses it. This remarkable stability can be attributed to two main design elements: differential data processing algorithms, which compensate for signal drifts caused by environmental variations, and optomechanical chassis design, which isolates the system from external light. These features ensure the sensor maintains high accuracy and stability, even in noncontrolled real world conditions.

The test strip's microfluidic architecture introduces challenges, such as potential sample evaporation in windy or air-exposed environments. In conventional open-strip systems, this leads to an unstable signal or premature drying of the reaction zone. However, in the current design, the detection zone is completely enclosed within the instrument, where the air or wind would have a significantly reduced effect on the strip. The structural protection not only prevents evaporation but also ensures consistent fluid dynamics and reliable optical detection.

To further reduce the measurement variability, sample collection protocols were carefully standardised. Saliva donors were instructed to follow a standard operating procedure that included drinking water 30 minutes prior to sample collection and refraining from consuming food or beverages for at least 30 minutes. This controlled conditioning of the oral environment

minimises fluctuations in saliva pH caused by dietary or hydration factors. The standard operating procedure ensured the scientific validity and ethical compliance of the testing process. Combining thoughtful engineering, algorithms, and standardised collection practices, the developed system demonstrates high repeatability, resilience to ambient interference and suitability for reliable, point of care biochemical sensing applications.

Conclusion

Through the use of a transmission-based optical detection approach, a portable, smart, and mobile-based pH sensor for measuring salivary pH was developed. With a response time of roughly 60 seconds, it showed limit of detection of pH 0.166 for green and pH 0.244 for red, which were within the clinically significant range.

Unlike traditional pH measurement methods, this device minimises interference from environmental factors such as ambient light and wind. It uses an affordable RGB colour sensor instead of costly mobile phone cameras, ensuring compliance.

Moreover, older smartphones with certain operation system versions may become obsolete, requiring regular updates to detection apps, as well as recalibration and revalidation of the pH sensor. This is important because the sensor's performance and accuracy are directly linked to the specifications of the smartphone camera. Our current effort to develop an independent meter aims to eliminate the disadvantages associated with smartphone-based, traditional strips and pH meters, enabling the public to regularly monitor their pH levels. Additionally, we intended to design a handheld system suitable for use in resource-limited environments. We maintained simplicity in the system's operation by integrating a mobile application, which ensures that it is user friendly and has little human intervention. The entire procedure: from sample collection to results presentation on the app, takes less than 2 minutes, enabling online data transmission and storage. A saliva sample is gathered from beneath the tongue using a cotton swab and applied to the test strip. The pH sensor is designed to minimise interference from ambient conditions, such as wind and light, and is optimised for use at room temperature. According to the approved standard operating procedure, users should rinse their mouths before sampling and refrain from eating or drinking for 30 minutes before testing saliva pH.

The results demonstrate a strong correlation between the estimated and actual pH values, confirming the device's effective functionality. Overall, the developed instrument has all the required features and is suitable for point of care devices, with potential for commercial application.

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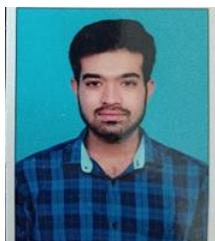
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