Measurement of dermal water content using a multi-wavelength optical sensor

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Abstract—. **Skin hydration is crucial for overall skin health. Maintaining skin hydration levels preserves skin integrity and prevents tissue damage which can lead to several debilitating conditions. Moreover, continuous monitoring of skin hydration can contribute to the diagnosis or management of serious diseases. For instance, sugar imbalance in diabetes mellitus and kidney disease can lead to the loss of bodily fluids and cause dry skin. Therefore, continuous, accurate and non-intrusive monitoring of skin hydration would present a remarkable opportunity for maintaining overall health and wellbeing. There are various techniques to assess skin hydration. Electrical based Corneometers are currently the gold standard in clinical and non-clinical practice. However, these techniques have a number of limitations. In particular, they are costly, sizeable, intrusive, and operator dependent. Recent research has demonstrated that near infrared spectroscopy could be used as a non-intrusive alternative for the measurement of skin water content. The present paper reports the development and in-vitro validation of a noninvasive, portable, skin hydration sensor. The results indicate that the developed sensor can deliver reliable measurements of skin water content.**

I. INTRODUCTION

The outermost layer of the skin, known as the Stratum Corneum (SC) acts as an essential permeable, heterogeneous, composite outer layer that controls the level of transcutaneous water loss and provides protection against desiccation and environmental challenges [1, 2]. The water content or degree of hydration of the SC layer can have profound effects on the transport and mechanical properties of human skin [3]. It maintains skin flexibility and facilitates the occurrence of certain enzymatic reactions which are responsible in driving the SC's exfoliating process, and regulates the epidermal barrier function [4].

The level of SC hydration or water-retaining capacity is highly dependent upon the structure and arrangement of the intercellular lipid matrix, and by the Natural Moisturization Factors (NMFs) which are comprised of water-soluble substances, and by the permeation path length through the SC. In turn, barrier function is influenced by the water content of the skin which regulates hydrolytic enzyme activities involved in SC maturation and desquamation of corneocytes.

Inadequate skin hydration can be an early sign of major complications. For instance, in type 1 and type 2 diabetes mellitus, skin disorders, such as dry skin, have been associated with the increased risk of skin lesions, ulcerations, and diabetic foot [5].

The current gold standard for measurement of skin moisture is based on electrical capacitance probes, but these techniques suffer certain limitations. The measurement can be affected by factors other than skin water content. The instruments are also costly, sizeable, intrusive and operator dependent [6, 7]. These limitations have sparked researchers to pursue alternative techniques. In particular, Near Infrared Spectroscopy (NIRS) has shown several advantages over traditional techniques in the analysis of skin hydration. Water, proteins and lipids absorb light in the NIR region of the optical spectrum. Water, in particular, has several prominent absorption peaks in the NIR spectrum. These properties can be used to deduce information pertaining to various aspects of skin health [8, 9, 10, 11]. Interestingly, it is possible to differentiate between bound and free water inside the skin [12]. These factors have established NIRS an important and active area of research for the assessment skin health [13, 14, 15].

The present study focuses on skin hydration and the delivery of a miniaturized, accurate, continuous, an inexpensive sensor for measuring skin water content. To this end, a four-wavelength wearable sensor was designed and prototyped to obtain rapid reflectance measurements at three water-absorbing wavelengths, namely 970 nm, 1200 nm, and 1450 nm. The fourth wavelength, 940 nm, was used as baseline correction control. An in-vitro experiment was carried out whereby the water content can be sufficiently varied and accurately measured in a controlled environment. Subsequently, part of the recorded data was used to train a model and part of the data was used to validate the model and assess its predictive power.

II. METHODS

A. Design & development of a hydration sensor

1) General system structure

The multi-wavelength Near-Infrared sensor was developed to detect, record and continuously display raw signals at selected wavelengths.

The system consists of two separate modules:

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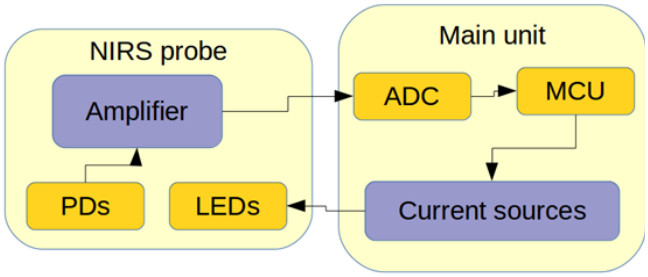

Fig. 1 Hydration sensor system structure

- *Probe:* Containing 4 LEDs, a Photodiode and a dual channel transimpedance amplifier
- Main module with Analog to digital converter (ADC), current sources and USB connection for data transfer.

2) Hydration sensor probe

A block diagram of the system is shown in Fig 1. The NIR probe is connected to the main unit via a flexible cable to facilitate contact with the skin. It is known that the NIR spectrum has a predominant water absorption band near 1450 nm. Two weaker-absorbing bands exist around 1195 nm and 970 nm [16]. Conversely, water does not absorb light around 940 nm, and this makes this wavelength a good choice for correcting for shifts in the absorbance values that are unrelated to water content. The developed probe consists of four surface mounted LEDs operating at wavelengths 940 nm (reference wavelength), 970 nm, 1200 nm and 1450 nm. A single photodiode is incorporated with the detection range of 900-1700 nm. The photodiode is surrounded by the four LEDs, each placed at a distance of 5 mm from the center of the photodiode. The dual-channel transimpedance (TI) amplifier is mounted at the other side of the probe so the detected signal is amplified immediately after the photodiode outputs, therefore, minimizing electromagnetic noise. The LED control signal, the amplified photodiode response and the power supply of the TI amplifier were connected via the cable to the main unit.

The printed circuit board (shown in Fig. 2) was designed and manufactured at the Research Centre for Biomedical Engineering, City, University of London. The probe was covered in clear epoxy to eliminate direct contact between the electronic components and the skin.

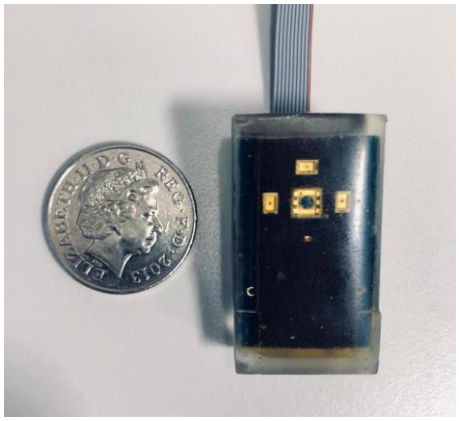

Fig 2. Hydration sensor probe

3) Hydration sensor main unit

The main unit is responsible for acquiring raw DC signals from the skin. A TL431 shunt voltage regulator ensures the stability of the voltages. The signals are digitized using a MCP3912 4-channel sigma-delta ADC chip. All devices' logics, ADC conversions, and power supply are controlled with an Arduino microprocessor in the main unit.

In order to obtain high quality data, it is necessary to isolate the photodiode's steady state response from its transient response. To achieve this, the sampling frequency of the ADC was selected such that each LED's on/off phase was sampled four times and the first three were ignored. This is depicted in Fig. 3.

The acquired data were transmitted to the processing module, where it underwent an exponential moving average smoothing to reduce noise. A graphical user interface was also developed in C++ to visualize the data in real time.

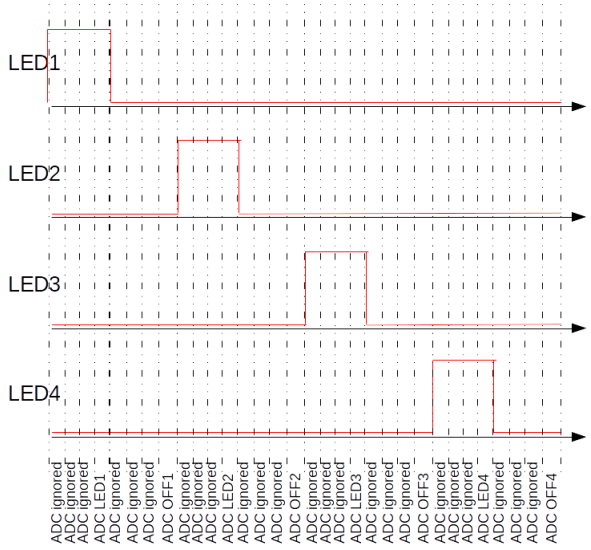

Fig. 3 Measurements timing. Each LED on/ off phase corresponds to 4 ADC samples, the first 3 are ignored and the 4th is used as signal value.

B. *Experimental setup of desorption tests*

This study was approved by the Senate Research Ethics committee at City, University of London prior to the experiments. The experimental setup was designed to simultaneously record the optical absorbance using the developed hydration sensor and reference gravimetric measurements using a precision analytical balance (*TR-24 electronic precision balance, Denver instruments GmbH, Germany*).

The study was carried out on porcine skin as it has the closest properties to the human skin. Fresh porcine skin was obtained immediately after slaughtering. The skin sample was cut using a scalpel into a 4x6 cm rectangle of roughly 1-1.5mm thickness. The specimen was then placed into an environmental chamber KMF 115 (*Binder GmbH, Tuttlingen, Germany*) at 96% Relative Humidity (RH) and 25ºC for 48 hours ensuring maximum hydration level had been reached.

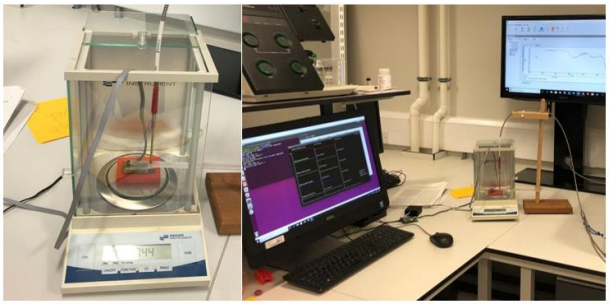

Fig. 4 The in-vitro experimental setup.

In order to mimic the natural dehydration process through the outermost later of the stratum corneum, customized sample containers were 3D printed with a single exposure side that isolated the internal sample and ensured that the desorption process occurred only through the skin surface.

Once fully hydrated, the skin sample was removed from the chamber and fitted inside the customized container, before being placed on the analytical balance. The weight of the container was recorded prior to the insertion of the skin sample, and this value was later deducted from the weight measurements to calculate the water content. The probe of the hydration sensor was placed on the exposed surface of the skin sample and clamped in place for the remainder of the experiment (Fig. 4).

Next, initial weight and optical measurements were recorded. Samples were then left in place to dehydrate over a period of 3 hours, and during that time, data collection was performed in two consecutive steps; (a) intermittent weight measurements were recorded every 15 minutes, enabling the calculation of the water content. (b) The developed sensor recorded DC reflectance values from the sample every 30 seconds throughout the 3-hour desorption process. (c) almost 24 hours after the start of the experiment, and when the specimen was found to be fully dry, a final weight measurement was carried out, reading 1.1183 g. The water content was then approximated by deducting this dry weight from all weight measurements.

III. RESULTS AND DISCUSSION

The Beer-Lambert law establishes a linear relationship between the absorbance and the concentration of analytes,

$$A = \epsilon l c, \qquad (1)$$

where A is absorbance, ϵ is the molar attenuation coefficient, l is the optical path length, c is the concentration of the attenuating species. Therefore, the demultiplexed reflectance values for each wavelength were converted to absorbance using (2),

$$A \propto \log(1/R), \qquad (2)$$

where A denotes absorbance, and R denotes reflectance measurements [17].

The absorbance measurements and the weight readings are depicted in Fig 5. Here, the absorbance of the 940 mm wavelength is subtracted from the other three to account for potential baseline shifts and for visualization purposes.

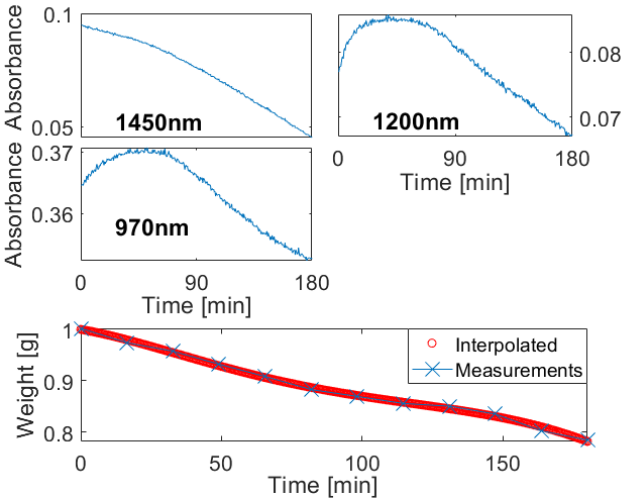

Fig. 5 The three figures on top demonstrate the abosorbance measurements pertaining to the three wavelengths (LEDs) that correspond to water absorption peaks in the NIR spectrum and over the 180-minute duration of the experiment. The absorbance given by the baseline control wavelength (940 nm) has been subtracted from all three for visulisation purposes. The figure at the bottom shows the weight measurements of the specimen (pocine skin). These measurement were carried out every 15 minutes (blue line with cross markers), the red markers depict the interpolated values.

As expected, after the subtraction of the baseline control wavelength (940 nm) from the three absorbance readings, all absorbance values show a downward trend, signifying the loss of water content. In particular, the absorbance values pertaining to 1450 nm, where the second major water absorption peak in the NIR region is, show a major drop and are highly correlated with the water content of the porcine skin. These relationships are more rigorously investigated using linear regression. Table 1. shows the linear regression results when each wavelength is separately used as the regressor, the absorbance value at 940 nm is used as control and the interpolated weight values are used as the regressand.

TABLE I. LINEAR REGRESSION BETWEEN THE ABOSRBANCE VALUES AND THE INTERPOLATED WEIGHT VALUES.

Regression results	**wavelengths**		
	LED 2 - 970 nm	***LED 3 – 1200 m***	***LED 4 – 1450 nm***
Coefficient-β_λ	-6.74 (1.70e-155)[a]	-6.76 (3.22e-136)	2.12 (3.59e-5)
Coefficient control-$\beta_{940\ nm}$	-1.85 (1.15e-79)	-1.80 (2.12e-62)	-4.43 (3.22e-74)
p-value model vs. constant model	≅0[b]	≅0	2.34e-226
Coefficient of determination R^2	0.992	0.99	0.945

a. The values in brackets correspond to the p-value of the coefficient

b. Values less than 9.99e-324 are approximated by zero

In order to combine the descriptive power of all four wavelengths, first their multicollinearity must be addressed. Principal Component Analysis (PCA) is a dimensionality reduction technique that effectively eliminates multicollinearity by projecting the data onto the axes of maximal variance in the input space. Principal Component

Regression (PCR) uses these projected values (PCA scores) as regressors.

In the dataset, 99.95% of the variance is explained by the first three principal components. Prior to training a model, 20% of the samples (72 samples) are randomly selected and held out as a test set, the remaining 80% of samples (289 samples) form the training set. Subsequently a PCR model is developed by regressing the weight measurements on the PCA scores corresponding to the first three PCs. This yields a model with $R^2_{calibration} = 0.997$, Root Mean Squared Error of Calibration (RMSEC) of 0.0033 g, and a p-value that is indistinguishable from 0 (less than 9.99e-324).

Fig 6 depicts the predictions pertaining to the test set against their reference values.

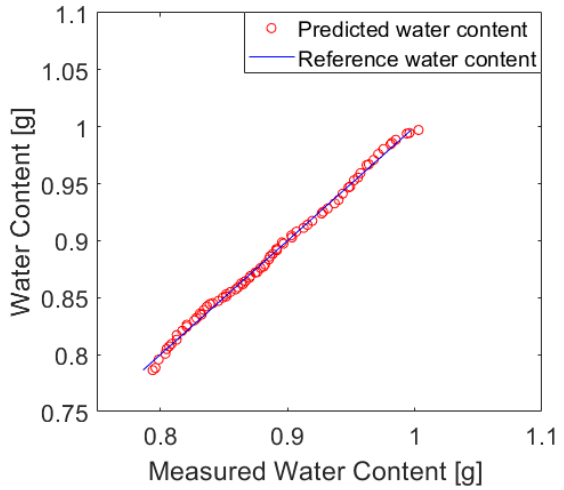

Fig 6. The performance of the model developed based on the measurements of the proposed optical skin hydration sensor. The red circles denote the predictions of the PCR model for a test set with 72 samples. The blue solid line shows the reference values.

On the test set, the model obtains a coefficient of determination, R^2_{Test}, of 0.997 and a RMSE of Prediction (RMSEP) of 0.0032 g.

IV. FUTURE WORK AND CONCLUSION

This paper reports the design and implementation of a prototype, coin-sized, wearable, skin hydration sensor.

The developed sensor uses four LEDs, three of which pertain to the three water absorption peaks in the NIR region of the optical spectrum. The fourth LED, with the wavelength of 940 nm, was used as baseline control. To test the performance of the sensor, an in-vitro experiment was conducted to create controlled variation of water content in porcine skin and record reliable measurements of the specimen's water content.

Following the completion of the experiment, a PCR model was developed to combine the information obtained from the four LEDs. 80% of the observations were used for training and the remaining 20% were used as a test set to assess the predictive performance of the model. The high accuracy of the predictions suggest that optical measurement of skin water may be a viable alternative to electric-capacitance-based approaches, enabling continuous monitoring and incorporation into wearable devices.

Our future work will be dedicated to conducting in-vivo studies on human skin, investigating the effects of skin types on the accuracy of measurements, and finally evaluating the application of nonlinear models such as support vector machines, artificial neural networks, and random forests to cope with inter-subject differences and nonlinearities.

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