Multi-Wavelength Optical Sensing and Monitoring of Dermal Water Content

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Declaration

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Iman Maimunat Gidado

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Abstract

Hydration status is a critical parameter in maintaining skin health and overall physiological balance, influencing a range of biological processes. This research presents the design, development, and validation of a novel multi-wavelength near-infrared (NIR) optical sensing device, specifically engineered for the accurate and non-invasive measurement of skin and body hydration levels. The research conducted spans comprehensive in silico modelling, controlled in vitro and ex vivo testing, and rigorous in vivo experimentation, ensuring a robust solution for real-time hydration monitoring.

The developed device operates by emitting NIR light at carefully selected wavelengths that correspond to absorption peaks of water molecules within biological tissues. By analysing the reflectance signals from these tissues, the device can accurately quantify water content across different skin layers. This addresses the limitations of existing techniques, which often suffer from inaccuracies due to interference or require invasive procedures. In addition, the multi-wavelength approach enhances specificity and depth resolution, providing a more comprehensive assessment of hydration status.

A key component of this research was the development of a calibration model, trained using datasets obtained from the experimental phases and cross-referenced with established hydration metrics. This model incorporates variations in gender, skin types and ethnicities, ensuring accuracy and reliability across a diverse population. The model takes the form of a protocol and device, combining structured workflows and real-time sensor data with user inputs to generate hydration predictions.

In vivo studies demonstrated the device's efficacy in tracking hydration changes induced by various factors, including exercise and topical applications of moisturizers. The results showed a high correlation between the device's readings and traditional hydration assessment methods (R^2 skin = 0.76, R^2 body = 0.58, validating its potential for clinical and personal use. Moreover, the non-invasive nature of the device makes it ideal for continuous hydration monitoring in dynamic environments, such as during physical activity.

This research represents an advancement in the field of hydration monitoring, offering a new tool that combines the precision of optical sensing with the practicality of wearable technology. The device not only enhances our ability to monitor hydration with greater accuracy but also opens up new possibilities for its application in medical diagnostics, sports science, and dermatological research. The findings and methodologies pave the way for future innovations in non-invasive health monitoring technologies, with the potential to contribute to a greater understanding of hydration-related health issues.

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Nomenclature

NIR	Near Infrared
O-H	Hydroxyl Bond
ATRIS	Attenuated Total Reflectance Infrared Spectroscopy
NMR	Nuclear Magnetic Resonance
NIRS	Near Infrared Spectroscopy
CM825	Corneometer (Courage and Khazaka)
MeOH	Methanol
p	p-value
SC	Stratum Corneum
SEM	Scanning Electron Microscopy
MMD	MoistureMeter (Delfin)
ME	Moisture Evaluator
ATI	Force/Torque Sensor
MRI	Magnetic Resonance Imagine
OA	Opto-acoustic
InGaAs	Indium Gallium Arsenide
LED	Light Emitting Diode
ZenPPG	Multi-Wavelength Photoplethysmography Research Platform
DAQ	Data Acquisition Card
PCB	Printed Circuit Board
CaF_2	Calcium Fluoride
TEWL	Transepidermal Water Loss
MLR	Multiple Linear Regression
R	Coefficient of Correlation
R^2	Coefficient of Determination
GUI	Graphical User Interface
MC	Monte Carlo
PLS	Partial Least Squares
UV	Ultraviolet
LCFA	Long Chain Fatty Acid
NMF	Non-Negative Matrix Factorization

PCA	Principle Components Analysis
DEJ	Dermal-Epidermal Junction
OCT	Optical Coherence Tomography
BIA	Bioimpedance Analysis
DEXA	Dual Energy X-Ray Absorptiometry
DNA	Deoxyribonucleic Acid
AD	Atopic Dermatitis
NICE	National Institute for Health and Care Excellence
ADH	Antidiuretic Hormone
H_2O	Water
C-H	Carbon-Hydrogen (CH) Bond
N-H	Amino (NH) Bond
DOT	Diffuse Optical Tomography
NFC	Near-Field Communication
G	Conductance
V	Voltage
R	Resistance
SCR	Skin Conductance Response
EDA	Electro-Dermal Activity
Z	Impedance
I	Current
CAD	Computer-Aided Design
LCR	Inductance-Capacitance-Resistance Meter
ECG	Electrocardiography
IR	Infrared
FTIR	Fourier Transform Infrared
NMR-S	Nuclear Magnetic Resonance Spectroscopy
TTT	Transient Thermal Transfer
CRM	Confocal Raman Microscopy
D_2O	Deuterium Oxide (Heavy Water)
CRS	Confocal Raman Spectroscopy
AC	Alternating Current
EM	Electromagnetic

RH	Relative Humidity
%RH	Percentage Relative Humidity
ADC	Analog to digital converter
USB	Universal Serial Bus
pH	Potential of Hydrogen
HR	Heart Rate
HRV	Heart Rate Variability
BP	Blood Pressure
SBP	Systolic Blood Pressure
DBP	Diastolic Blood Pressure
BUN	Blood Urea Nitrogen
CVP	Central Venous Pressure
ECV	Extracellular Volume
ICV	Intracellular Volume
vs	Versus
OLED	Organic Light Emitting Diode
NCAA	National Collegiate Athletic Association
TIA	Transimpedance Amplifier
NI	National Instruments
DC	Direct Current
AC	Alternating Current
Delta	Difference
w	Weight
t	Time
H(t)	Hydration Index
g	Anisotropy
μa	Absorption Coefficient
μs	Scattering Coefficient
Cal	Calibration
CV	Cross Validation
RMSE	Root Mean Square Error
RMSEC	Root Mean Square Error for Calibration
RMSECV	Root Mean Square Error for Cross Validation

A	Absorbance
R	Reflectance
TEIM	Tetrapolar Electrical Impedance Measurement
R-C	Resistor-Capacitor
MAE	Mean Absolute Error
CSV	Comma Separated Value
VI	Virtual Instruments
IoT	Internet of Things
X	Independent Variables
Y	Dependent Variable

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JOURNAL PAPERS

- Gidado IM, Qassem M, Triantis IF, Kyriacou PA. Review of Advances in the Measurement of Skin Hydration Based on Sensing of Optical and Electrical Tissue Properties. Sensors (Basel). 2022 Sep 21;22(19):7151. doi: 10.3390/s22197151. PMID: 36236250; PMCID: PMC9571519.
- Gidado, I.M.; Nwokoye, I.I.; Triantis, I.F.; Qassem, M.; Kyriacou, P.A. Multi-Modal Spectroscopic Assessment of Skin Hydration. *Sensors* **2024**, *24*, 1419. https://doi.org/10.3390/s24051419
- Gidado, I., Al-Halawani, R., Qassem, M. *et al.* Development and Analysis of a Multi-Wavelength Near-Infrared Sensor for Monitoring Skin Hydration and Validation Using Monte Carlo Simulation. *Photonic Sens* **14**, 240306 (2024). https://doi.org/10.1007/s13320-024-0719-z

CONFERENCE PROCEEDINGS

- Gidado, I.M.; Qassem, M.; Kyriacou, P.A. In Vivo Assessment of Skin Hydration and Moisturisation Using a Multi-Wavelength Optical Sensing Wearable. *In 2024 46th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC). IEEE.*

PUBLICATIONS IN PROGRESS

- *Gidado, I.M.; Qassem, M.; Kyriacou, P.A. Assessing Skin Hydration Dynamics with a Wearable Optical Sensor: Longitudinal Analysis and Regression Model Approach.*
- *Gidado, I.M.; Castro Montano, M.; Qassem, M.; Kyriacou, P.A. Relevance of Dominant O-H bands using NIRS in Healthcare and Wellbeing.*
- *Gidado, I.M.; Qassem, M.; Kyriacou, P.A. Effect of Dehydration on Optical Skin Hydration Status – A Case Study.*

CHAPTER 1:

INTRODUCTION

1.1 Aims & Objectives

Water content is a critical biophysical parameter that relates to various skin conditions and processes, such as transdermal drug delivery, aging, and overall body hydration. Currently, no direct technique for measuring skin water content exists, and existing devices typically rely on indirect methods, such as measuring capacitance, resistance, or transepidermal water loss. These indirect methods limit the accuracy and usefulness of hydration measurements in both clinical practice and research settings.

Previous research at the Research Centre for Biomedical Engineering at City, University of London has demonstrated the effectiveness of Near Infrared (NIR) Spectroscopy for providing a more accurate and comprehensive analysis of skin health [90,94,95,96,107]. This technique allows for non-invasive assessment of dermal hydration and skin barrier function by penetrating the skin and targeting water absorption at various depths. The use of multi-wavelength NIR systems has been shown to deliver precise measurements, offering a detailed understanding of skin hydration levels by analysing specific absorption bands [64,106]. Additionally, the development of early wearable prototypes has shown promise in enabling continuous monitoring of skin conditions, taking into account external factors such as temperature and humidity that can affect skin health [64].

Building on this foundational work, this research aims to design, develop, and validate a wearable multi-wavelength optical sensor for monitoring dermal water content and skin barrier function. By refining the existing technology and improving its capacity for real-time, continuous monitoring, this study seeks to contribute further to the field of non-invasive skin health assessment, offering a reliable tool for tracking hydration and skin integrity in various conditions. The specific objectives of this research are as follows:

a. **In vitro spectroscopic analysis for sensor testing and calibration:** Conduct a range of spectrophotometric analyses of varying hydration levels in tissue phantoms, and other media. This is done to perform decomposition analysis of fundamental absorptions in the skin and to calibrate a developed multi-wavelength sensor against reference methods.
b. **Development and validation of a wearable device:** Enhance the developed sensor into a user-friendly wearable device, followed by ex vivo validation studies

on porcine skin. Cross referencing experiments will also be carried out with electrical methodologies.

c. **Validation study on human participants:** Conduct a series of in vivo validation studies on healthy volunteers to evaluate the developed system. These studies will assess both skin hydration through long-term moisturization and body hydration via induced dehydration, highlighting the sensor's practical applications. Data will be collected for future developments of computational and predictive models and analysed using correlation-based analysis techniques.
d. **Calibration model and baseline referencing:** Develop and validate a calibration model that accounts for individual variability in skin type, age, gender, and ethnicity. This model, along with baseline referencing techniques, will be essential for ensuring the sensor's accuracy and reliability across diverse populations.

1.2 Thesis Outline

Chapter 1: Introduction

This chapter introduces the research objectives, outlining the need for innovation in hydration measurement technologies. It provides a comprehensive overview of previous research conducted within the Research Centre for Biomedical Engineering at City University of London, particularly in the domain of skin hydration measurement. The chapter sets the stage for the subsequent research by identifying the limitations of current techniques and the potential impact of a novel multi-wavelength optical sensor.

Chapter 2: Structural Properties and the Hydration of Skin and Body

This chapter delves into the anatomical and physiological characteristics of the skin, focusing on its structural layers and their respective roles in maintaining hydration and barrier function. Detailed descriptions of the epidermal and dermal layers are provided, including their cellular composition and the function of lipids in controlling transepidermal water loss. The chapter also differentiates between body hydration, skin hydration, and skin moisture, providing a foundation for understanding the various factors influencing skin and body water content.

Chapter 3: Motivations and Applications of the Measurement of Hydration

This chapter explores the various motivations for measuring hydration, both in clinical and consumer contexts. It discusses the implications of dehydration, skin pathologies such as eczema and psoriasis, and the importance of hydration in cosmetic applications. The chapter also addresses the specific needs of vulnerable populations, such as neonates and the elderly, as well as the role of hydration monitoring in athletic performance and general well-being.

Chapter 4: Relevance of Dominant O-H Bands Using Near-Infrared Spectroscopy in Healthcare and Wellbeing

This chapter focuses on the characteristics and significance of O-H bands in near-infrared (NIR) spectroscopy, particularly their strong absorption associated with water molecules. The chapter discusses the interaction of NIR light with biological tissues, emphasizing its potential for non-invasive hydration measurement. It also reviews the

applications of NIR spectroscopy in healthcare, including its use in monitoring skin hydration, assessing tissue oxygenation, and detecting pathological changes in tissues.

Chapter 5: Review of Techniques and Advances in the Measurement of Hydration Based on Sensing of Optical and Electrical Tissue Properties

This chapter provides a comprehensive review of existing techniques for measuring skin hydration, with a focus on both electrical and optical methods. It compares the advantages and limitations of each technique, including skin capacitance, conductance, bioimpedance, and various optical methods such as NIR spectroscopy and Raman spectroscopy. The chapter also discusses the potential of multi-modal approaches that combine electrical and optical sensing for more accurate and comprehensive hydration assessment.

Chapter 6: Current State of the Art for Wearable Measurement of Hydration

This chapter examines the latest developments in wearable hydration monitoring devices, focusing on state-of-the-art technologies that integrate optical and electrical sensing methods. It reviews recent advancements in sensor miniaturization, data processing, and wireless communication, highlighting the challenges and future directions for wearable hydration measurement. The chapter also includes a discussion on the commercial viability of these devices and their potential impact on healthcare and consumer markets.

Chapter 7: Design and Development of Quad-Wavelength Near-Infrared Optical Sensor for Measuring Hydration Levels

This chapter details the design and development process of the novel optical hydration sensor created during this research. It describes the selection of NIR wavelengths, the construction of the sensor prototype, and the integration of hardware and software components. The chapter also compares the performance of the developed sensor with a standard spectrophotometer, providing insights into the advantages of the multi-wavelength approach.

Chapter 8: Validation of Sensor Design via In Silico and In Vitro Experimentation

This chapter presents the results of the initial validation studies conducted on the developed sensor. It includes in silico simulations using Monte Carlo methods and in vitro experiments on silicon phantoms and porcine skin. The chapter describes the experimental protocols, data pre-processing methods, and the interpretation of results, demonstrating the sensor's capability to accurately measure hydration levels in controlled environments.

Chapter 9: Validation of Developed Optical Device via Ex Vivo Experimentation

This chapter extends the validation of the sensor to ex vivo conditions, specifically testing its performance on porcine skin samples. The chapter details the experimental setup, data analysis, and comparison with traditional hydration measurement techniques. It also explores the potential of a multimodal approach, combining optical and electrical methods to enhance measurement accuracy.

Chapter 10: Validation of Developed Optical Device via In Vivo Experimentation on Human Participants

This chapter presents the findings from in vivo studies conducted on human participants to evaluate the sensor's performance in real-world conditions. It covers the methodology, including participant selection, experimental design, and statistical analysis of the data. The chapter also discusses the development of a calibration model based on the in vivo results, highlighting the sensor's potential for clinical and consumer applications.

Chapter 11: Discussion, Conclusions & Future Work

This final chapter synthesizes the findings from the experimental studies and literature review, discussing the implications of the developed sensor for hydration monitoring. It addresses the challenges encountered during the research and outlines potential future work, including the refinement of calibration techniques, further in vivo testing, and the development of a wearable device for continuous hydration monitoring. The chapter concludes by summarising the contributions of this research to the field of hydration measurement and its potential impact on healthcare and consumer technology.

CHAPTER 2:

STRUCTURAL PROPERTIES AND THE HYDRATION OF SKIN AND BODY

This chapter details the complex structure of the skin and its critical functions in maintaining health and hydration. The skin is composed of multiple layers, each contributing to its barrier function and overall integrity. The outermost layer, the epidermis, includes five sub-layers that work together to protect against environmental insults and prevent water loss. The stratum corneum, made up of dead keratinocytes in a lipid matrix, is crucial for this barrier function.

Beneath the epidermis, the dermis supports and nourishes the upper layers. It is divided into the papillary dermis, which supplies nutrients and sensory input, and the reticular dermis, which provides strength and elasticity. Key to maintaining skin hydration are intracellular lipids and natural moisturizing factors (NMFs), which form a barrier that regulates transepidermal water loss (TEWL).

The chapter also explores the skin's unique electrical and optical properties, essential for various diagnostic applications. Techniques such as bioimpedance spectroscopy and near-infrared (NIR) spectroscopy leverage these properties to non-invasively measure skin hydration and integrity.

Furthermore, the chapter differentiates between skin hydration, skin moisture, and body hydration. Skin hydration refers to water content within the cells, while skin moisture pertains to the retention of natural oils. Body hydration involves the total water content in the body, affecting overall physiological functions. Advanced optical sensing techniques, like multi-wavelength NIR spectroscopy, offer precise, non-invasive measurements of dermal water content, enabling continuous monitoring of hydration status.

Overall, understanding the structural properties and hydration mechanisms of the skin and body is crucial for improving skin health, managing hydration, and enhancing performance in various clinical, cosmetic, and athletic contexts. The following comprehensive review underscores the importance of integrating advanced measurement techniques to achieve better health outcomes.

2.1 Skin Structure & Barrier Function

The skin is a complex organ composed of multiple layers, each with distinct structures and functions essential for maintaining overall skin health. The outermost layer, the epidermis, is a stratified epithelium that serves as the primary barrier against environmental insults, pathogens, and water loss. The epidermis is subdivided into five layers: the stratum basale, stratum spinosum, stratum granulosum, stratum lucidum, and stratum corneum (SC). Each layer plays a specific role in skin homeostasis.

The stratum basale, or basal layer, consists of proliferative keratinocytes that continually divide and push older cells towards the surface. This layer also contains melanocytes, which produce the pigment melanin that protects against UV radiation. The stratum spinosum, or spinous layer, provides structural integrity through desmosomal connections between keratinocytes. The stratum granulosum is characterized by the presence of keratohyalin granules, which are essential for the formation of the skin's protective barrier.

The stratum lucidum, present only in thick skin such as the palms and soles, provides an additional layer of protection. The outermost layer, the stratum corneum, is composed of corneocytes—dead keratinocytes embedded in a lipid matrix. This layer is crucial for the skin's barrier function and is typically 5-20 µm thick. The corneocytes are tightly packed and held together by corneodesmosomes, which are enzymatically degraded during the process of desquamation, allowing for the shedding of dead cells and the maintenance of a healthy skin surface[1], [2] .

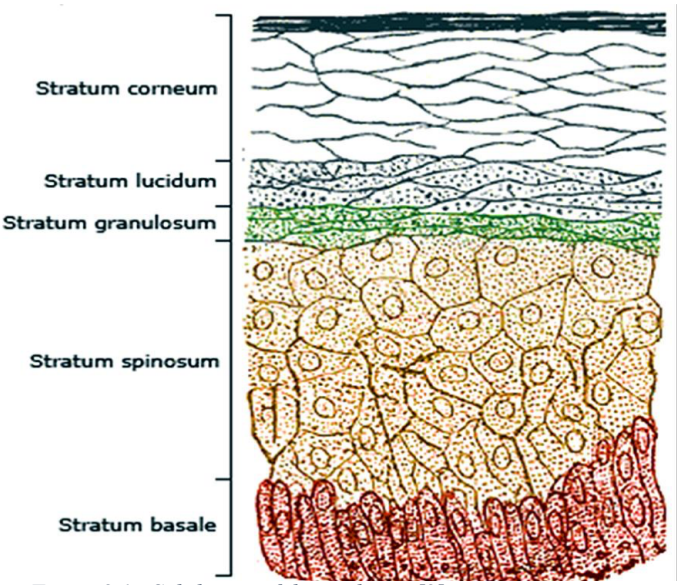


Figure 2.1 : Sub-layers of the epidermis[2]

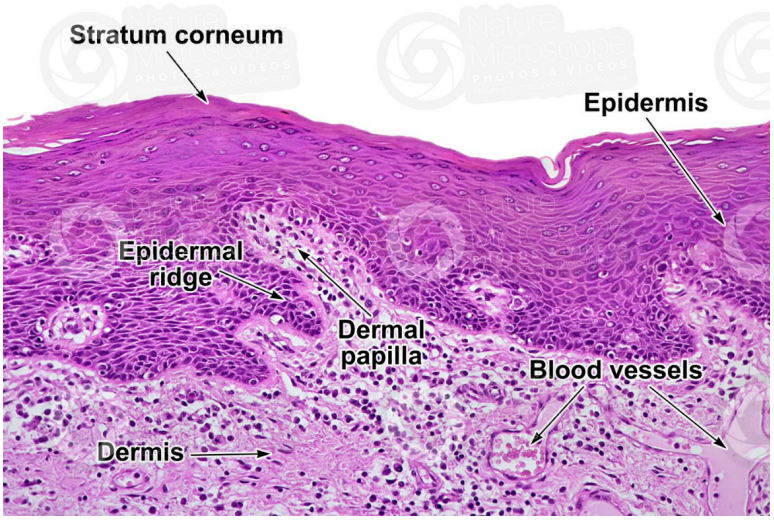


Figure 2.2: Stratum corneum layer of the epidermis[3]

The dermis, located beneath the epidermis, is divided into the papillary and reticular layers. The papillary dermis consists of loose areolar connective tissue with capillaries, lymph vessels, and sensory neurons, providing nutrients and support to the epidermis. Dermal papillae project into the epidermis, enhancing nutrient exchange and anchoring the epidermis to the dermis. The reticular dermis, composed of dense irregular connective tissue, provides tensile strength and elasticity to the skin, enabling it to withstand mechanical stress.

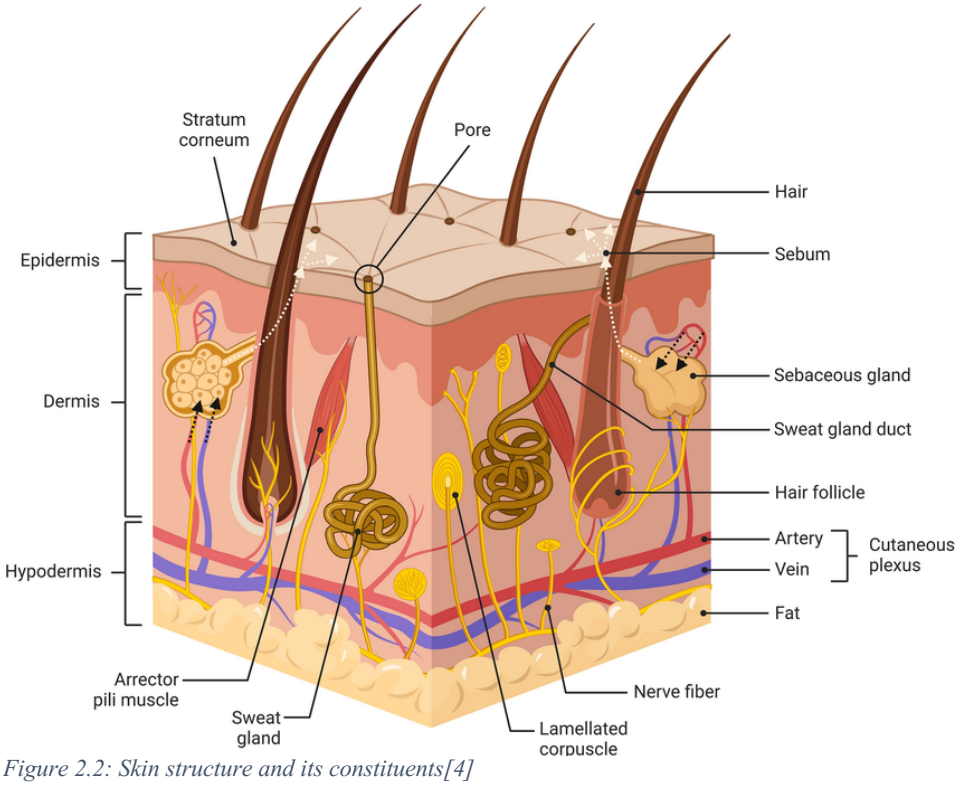


Figure 2.2: Skin structure and its constituents[4]

Intracellular lipids, such as ceramides, free fatty acids, and cholesterol, play a vital role in maintaining the skin's barrier function and regulating transepidermal water loss (TEWL). These lipids form a lamellar structure within the stratum corneum, creating a barrier that prevents excessive water loss while allowing for selective permeability. Long-chain saturated fatty acids (LCFA) are particularly effective in retaining water and minimizing TEWL due to their hydrophobic properties [5], [6].

Hydration levels in the stratum corneum are influenced by several factors, including water delivery from the underlying epidermal layers, evaporation from the skin surface through perspiration, and the water-holding capacity of the stratum corneum. Natural moisturizing factors (NMFs), such as amino acids, urea, lactate, and PCA, are hygroscopic compounds that attract and retain water within the stratum corneum. These NMFs are crucial for maintaining skin hydration and flexibility[7], [8].

The desquamation process, a form of programmed cell death, involves the shedding of corneocytes from the stratum corneum's surface. This process is regulated by proteolytic enzymes that degrade corneodesmosomes, allowing for the removal of aged and damaged cells. Desquamation ensures the renewal of the skin surface, maintaining its protective function and overall health[9].

In addition to the vital functions of the epidermis and dermis, the skin’s barrier function is also critically influenced by the dynamic interplay between the stratum corneum and its underlying structures. The stratum corneum is not merely a static layer but a highly active zone where continuous processes of keratinocyte differentiation, lipid synthesis, and desquamation occur. The effectiveness of the skin’s barrier function largely depends on the integrity and organization of the lipid matrix within the stratum corneum, which is predominantly composed of ceramides, cholesterol, and free fatty acids. These lipids are organized into a lamellar structure that forms multiple layers, contributing to the barrier's ability to prevent water loss and protect against environmental insults. Disruption in the lipid composition or organization, often due to genetic factors, environmental stress, or aging, can lead to increased transepidermal water loss (TEWL) and a compromised barrier function, resulting in conditions such as xerosis and dermatitis[3], [5].

Furthermore, skin thickness and pigmentation vary significantly across different body sites and among individuals of different ethnic backgrounds, affecting both the barrier function and the efficacy of hydration strategies. For instance, individuals with

darker skin types, which contain higher levels of melanin, often have a more compact and thicker stratum corneum. This structural difference can influence the rate of TEWL and the penetration of topical agents, necessitating tailored approaches to skincare and hydration treatments. The presence of melanin also contributes to the skin's natural defence against UV radiation, but it can complicate the assessment of skin hydration using optical methods due to its light-absorbing properties[10], [11].

Moreover, recent research highlights the importance of the dermal-epidermal junction (DEJ), a critical interface between the epidermis and dermis, in maintaining skin integrity and function. The DEJ is responsible for anchoring the epidermis to the dermis, ensuring mechanical stability, and facilitating communication between these layers. It plays a vital role in the repair and regeneration processes of the skin following injury. Alterations in the structure or function of the DEJ, which can occur due to aging or chronic exposure to environmental stressors, are associated with a decrease in skin elasticity and an increased risk of barrier dysfunction[12].

Lastly, the interplay between external environmental factors and intrinsic skin properties is crucial in understanding skin health. Factors such as humidity, temperature, and UV exposure can significantly impact the skin's barrier function and hydration levels. For example, low humidity environments can accelerate TEWL, leading to skin dryness and irritation. Conversely, excessive moisture or prolonged exposure to water can disrupt the lipid matrix, weakening the barrier function. Understanding these interactions is essential for developing effective skincare regimens that protect and maintain the skin's barrier function across diverse environmental conditions[3], [13].

These considerations emphasize the need for a holistic approach in skincare, one that accounts for the individual variability in skin structure and function, as well as the influence of external environmental factors. Ongoing research into the molecular mechanisms governing skin barrier function and hydration will continue to inform the development of targeted treatments aimed at preserving skin health and preventing barrier-related disorders.

2.2 Electrical and Optical Properties of Skin

The skin exhibits unique electrical and optical properties that are crucial for various diagnostic and monitoring applications. The electrical properties of the skin, including capacitance, conductance, and impedance, vary with hydration levels and skin conditions. These properties are exploited in techniques such as bioimpedance spectroscopy, which measures the opposition of biological tissues to the flow of an electric current, providing valuable information on skin hydration and integrity[7].

Optically, the skin interacts with light through processes such as absorption, scattering, and reflection. The absorption characteristics of the skin are primarily influenced by chromophores like melanin, haemoglobin, and water. Near-infrared (NIR) spectroscopy, for example, leverages these properties to non-invasively measure tissue composition and hydration by analysing specific absorption bands of water and other constituents. The depth of light penetration in NIR spectroscopy allows for the assessment of both superficial and deeper layers of the skin[14].

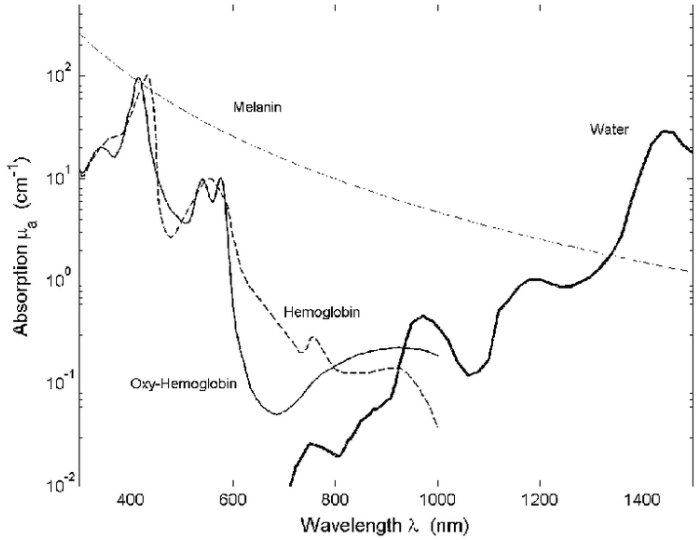


Figure 2.3: Absorption spectra describing the different absorbers in human skin in the near-infrared region[15]

The electrical properties of the skin, such as capacitance, conductance, and impedance, are intricately linked to the skin's hydration state and structural composition. These properties vary not only with hydration levels but also with the thickness of the stratum corneum and the distribution of skin lipids. As water content in the skin increases, both the conductance and capacitance rise, while impedance decreases, reflecting a higher ability of the skin to transmit electric signals. This relationship is

fundamental to the principles behind bioimpedance spectroscopy, a technique widely used in clinical settings to assess body water content, diagnose skin conditions, and monitor therapeutic interventions. Additionally, the dielectric properties of the skin, influenced by both water and lipid content, are critical in the design and optimization of wearable sensors and devices aimed at real-time skin monitoring[7], [15].

In terms of optical properties, the skin's interaction with light is highly complex, governed by factors such as chromophore concentration, skin thickness, and the scattering properties of the various skin layers. Melanin and haemoglobin are primary chromophores in the skin that absorb light at different wavelengths, thereby influencing the skin's optical characteristics. The optical density of melanin, for example, is a significant factor in determining the skin's response to ultraviolet and visible light, while the absorption spectra of haemoglobin are essential for non-invasive diagnostics that rely on blood oxygenation levels. The unique absorption bands of water, particularly in the near-infrared (NIR) region, make NIR spectroscopy a powerful tool for assessing skin hydration by measuring the water content in both the epidermis and dermis layers.

Furthermore, the scattering of light within the skin layers is a critical factor in optical diagnostics. The scattering is predominantly caused by structural components of the skin, such as collagen fibers in the dermis and the microstructure of the stratum corneum. This scattering effect can alter the path of light through the skin, affecting the accuracy of optical measurements. Techniques like optical coherence tomography (OCT) and diffuse reflectance spectroscopy leverage these scattering properties to provide detailed images and information about the skin's microstructure and function. These methods are increasingly being used in dermatology to assess skin conditions, treatment efficacy, and even in cosmetic applications to evaluate skin texture and aging[14], [16].

Moreover, the combination of electrical and optical properties in a single diagnostic platform can enhance the accuracy and reliability of skin assessments. For instance, integrating bioimpedance measurements with optical spectroscopy can provide a more comprehensive understanding of skin hydration and tissue composition. This multimodal approach is particularly valuable in the development of advanced wearable devices for continuous skin monitoring, enabling personalized skincare and more precise medical diagnostics. As research in this field advances, the potential for non-invasive, real-time monitoring of skin health continues to expand, promising significant improvements in both clinical outcomes and consumer healthcare technologies[17], [18].

2.3 Skin Hydration & Transepidermal Water Loss

Intracellular lipids within the stratum corneum play a critical role in maintaining the skin's barrier function and regulating transepidermal water loss (TEWL). TEWL refers to the passive diffusion of water from the body through the epidermis into the external environment, making it a key parameter for evaluating the integrity of the skin barrier. The ability of the stratum corneum to retain water is primarily facilitated by intercellular lamellar lipids, which create a semi-permeable barrier that prevents excessive water loss while allowing for selective permeability. In addition to these lipids, natural moisturizing factors (NMFs), composed of hydrophilic molecules such as amino acids, urea, and lactates, are crucial for retaining moisture within the stratum corneum. These NMFs attract and bind water, which is essential for maintaining skin flexibility, resilience, and overall health[1], [5].

Skin hydration is determined by the water content within the epidermal cells, and its maintenance is dependent on several factors, including water delivery from the dermal layers, evaporation through perspiration, and the stratum corneum's water-holding capacity. When skin cells are well-hydrated, they swell and support the skin's structural integrity, making it more supple and capable of withstanding external stressors. In contrast, a lack of adequate hydration can lead to reduced skin elasticity, increased TEWL, and a higher susceptibility to skin conditions such as eczema and dermatitis.

The concept of skin moisture differs slightly from hydration, as it pertains specifically to the retention of the skin's natural oils by lipid-producing cells, which form a protective barrier that prevents water loss and dryness. Moisturizers work by reinforcing this barrier, reducing water evaporation, and thereby minimizing TEWL. This is particularly important in managing skin conditions where the barrier function is compromised, such as in eczema. Without proper moisturization, the skin is prone to dryness, irritation, and increased TEWL, which can exacerbate these conditions[8].

On a broader scale, body hydration refers to the total water content within the body and encompasses the processes of dehydration and rehydration. While skin hydration is directly influenced by overall body hydration, the two are not synonymous. Total body hydration is crucial not only for maintaining skin health but also for supporting various systemic functions, such as temperature regulation, digestion, and

circulation. Changes in body hydration are often monitored through fluctuations in body mass and the tracking of fluid balance, which are indicative of overall water retention or loss[19].

Water loss from the body can be categorized into sensible and insensible fluid loss. Sensible fluid loss, which includes water loss through urine and faeces, is easily measurable. Insensible fluid loss, however, occurs through processes like skin diffusion and respiration and is not as easily recognized. This type of water loss can increase under certain pathological conditions, such as fever or burns, leading to a significant impact on both skin hydration and overall body water balance[19].

Understanding and managing TEWL is essential for maintaining healthy skin, as it directly influences the skin's moisture levels and barrier function. Advanced skincare regimens and products focus on both hydrating the skin and sealing in moisture to ensure that the skin remains hydrated and resilient, even in the face of environmental and physiological challenges[5].

2.4 Body Hydration

Body hydration refers to the balance of fluids within the body's cells, tissues, and organs, which is crucial for maintaining essential physiological functions. Water is vital for processes such as circulation, digestion, temperature regulation, and waste elimination. It constitutes about 60% of an adult's body weight and supports functions ranging from maintaining blood volume to lubricating joints and aiding in the transport of nutrients and oxygen to cells. Adequate hydration is also critical for cognitive function and physical performance, influencing aspects from energy levels to mental clarity[20].

To assess hydration status, various methods are used, each with unique advantages and limitations. Bioelectrical Impedance Analysis (BIA) is a common technique that estimates body water content by measuring the resistance of bodily tissues to a small electrical current. This method is non-invasive and widely used in both clinical and athletic settings. Dilution techniques, involving the use of tracers like deuterium oxide, offer high accuracy in estimating total body water but are typically reserved for research due to their invasive nature. Imaging techniques, such as Magnetic Resonance Imaging (MRI) and Dual-energy X-ray Absorptiometry (DEXA), also provide

insights into body composition, including fat and water distribution, although they are less accessible for routine use due to cost and complexity[21], [22].

Dehydration occurs when the body loses more water than it takes in, leading to a fluid deficit that can impair physiological functions. Even mild dehydration, defined as a 1-2% loss of body weight due to fluid loss, can cause symptoms such as headaches, fatigue, dizziness, and impaired cognitive function. Severe dehydration, involving a loss of 5% or more of body weight, can lead to life-threatening complications like kidney failure, heatstroke, and hypovolemic shock. Certain populations, particularly the elderly and young children, are more susceptible to dehydration. Older adults often have a reduced sense of thirst and may take medications that increase fluid loss, while infants and young children have a higher body water content and metabolic rate, making them more vulnerable to fluid imbalances[21], [23].

The consequences of chronic dehydration are severe and can lead to long-term health issues such as kidney damage, urinary tract infections, and an increased risk of chronic kidney disease. In athletes, dehydration not only impairs performance but also increases the risk of heat-related illnesses, such as heat exhaustion and heatstroke. Cognitive functions, including attention, memory, and reaction time, are also adversely affected by dehydration, which can significantly impact activities requiring sustained mental focus, such as studying or operating machinery[24].

Maintaining proper hydration is essential for health across all age groups and is particularly critical for vulnerable populations such as the elderly and young children. Advanced methods for assessing hydration status, while often requiring sophisticated equipment, provide valuable insights into body water distribution and are crucial for preventing and managing dehydration. Understanding the signs, risks, and consequences of dehydration is key to ensuring that individuals maintain adequate hydration for optimal health and well-being.

2.5 Body Hydration vs Skin Hydration vs Skin Moisture

Skin hydration refers to the water content within skin cells, which is crucial for maintaining cell structure and function. Proper hydration leads to the plumping of cells, making the skin appear more radiant due to enhanced light reflection. Dehydrated skin, on the other hand, can appear dull and tight, potentially triggering skin to produce excess oil as a compensatory mechanism. While increasing water intake or topical hydrators can help improve skin hydration by infusing skin cells with water, these measures alone may not be sufficient to maintain hydration in the deeper skin layers. This is because the stratum corneum, acts as a barrier that can limit water retention. Therefore, appropriate moisturization is necessary to protect this barrier and lock in hydration[25].

Skin moisture or moisturizing involves applying products that help to trap and seal water within the skin, along with natural oils, to reinforce its protective barrier. This barrier function is essential in preventing transepidermal water loss (TEWL), which is the process by which water evaporates from the skin's surface. Dry, flaky skin often results from an insufficient lipid matrix in the stratum corneum, which impairs the skin's ability to retain moisture. Moisturizers work by replenishing the skin's lipid content and reducing TEWL, making them particularly important for managing dry skin conditions like eczema and psoriasis[8].

Body hydration, on the other hand, refers to the overall water content within the body and encompasses the processes of dehydration and rehydration. While skin hydration is influenced by overall body hydration—since adequate body water levels are necessary to support skin hydration—skin moisture specifically refers to the skin's ability to retain this hydration, which is not directly dependent on overall body hydration. Water loss from the body occurs through various pathways, including skin diffusion, excretion, and respiration. Maintaining adequate body hydration is essential for numerous physiological processes, such as temperature regulation, digestion, circulation, and overall skin health. Dehydration can impair these processes and negatively affect both physical and cognitive performance, underscoring the importance of effective hydration monitoring, particularly in athletes and other high-risk groups[19], [20], [26].

Advanced optical sensing techniques, such as multi-wavelength near-infrared (NIR) spectroscopy, are being developed to provide precise, non-invasive and continuous measurements of dermal water content, to improve the management of skin health.

CHAPTER 3:

MOTIVATIONS & APPLICATIONS OF THE MEASUREMENT OF HYDRATION

3.1 Skin Hydration

3.1.1 Pathologies & Diseases Related to Skin Hydration

Several skin conditions and diseases are closely associated with impaired skin hydration and barrier function. These include xerosis, eczema (atopic dermatitis), psoriasis, and photodamaged skin, among others. Effective management of these conditions often involves the use of moisturizers and topical treatments. Devices capable of assessing the effectiveness of these treatments are crucial, with the Corneometer® CM 825 being the current gold standard for measuring skin hydration[27].

Xerosis, or dry skin, is a common condition characterized by dry, itchy, and rough skin. It is particularly prevalent among the elderly and in dry climates. Individuals with xerosis can experience transepidermal water loss (TEWL) up to 75 times greater than healthy skin due to decreased fatty acids and sebum production. This results in dehydration of the stratum corneum (SC), leading to cell shrinkage, impaired barrier function, and increased sensitivity to irritants and allergens. The natural breakdown of corneodesmosomes is disrupted, causing the skin to appear scaly and rough. Management typically involves the use of emollients and humectants to restore hydration and improve barrier function[3].

Sensitive skin is characterized by heightened intolerance to topical preparations and environmental conditions. Symptoms often include stinging, burning, itching, and tightness. Objective evaluations using TEWL, calorimetry, and skin impedance can provide detailed insights into the skin barrier function of individuals with sensitive skin. Studies have shown that individuals with sensitive skin have increased TEWL and lower hydration levels, indicating a compromised barrier[3].

Photodamaged skin results from prolonged exposure to ultraviolet (UV) radiation. Acute photodamage manifests as erythema, dryness, scaling, sunburn, and pigmentation changes, while chronic exposure can lead to photoaging, melanomas, and skin cancer. UV radiation induces photochemical reactions that oxidize nucleic acids, alter DNA, and modify skin lipids and proteins. These changes increase the thickness of the SC, leading to dehydration, micro-fissures, and increased desquamation. Protection against photodamage involves the use of broad-spectrum sunscreens and antioxidants to neutralize free radicals. Additionally, studies have shown that chronic UV exposure can

degrade collagen and elastin fibers in the dermis, leading to wrinkles and loss of skin elasticity[28].

Eczema, or atopic dermatitis (AD), is a skin inflammation disease that causes dryness and rashes to appear on the skin surface. It is specified by appearance and location and it arises due to an allergic reaction, hereditary cases or aging. Individuals with AD typically suffer from a combination of asthma, eczema and allergies and it is highly associated with sensitive skin. This condition results in skin inflammation, scratching and elevated amounts of IgE, with treatments involving the regular usage of moisturizers. Continued scratching of the skin due to AD damages the mechanical barrier function and alters the composition of the skin content, therefore leading to xerosis and making individuals more vulnerable to allergens. Furthermore, AD decreases the hydration and lipid levels in the SC as well as increasing TEWL[29], [30].

Skin hydration and transepidermal water loss (TEWL) are important skin biophysical parameters for the assessment of childhood eczema and as indicators of barrier function of the skin. In clinical settings, subjective and objective assessments are used in combination to evaluate eczema disease severity. Skin hydration, moisture and TEWL are all objective measurements used to assess the severity and chronicity of eczema, with hydration reporting lower in children with eczema particularly in the first 2 years of life. In a study conducted by Hon et al, skin hydration was measured using Delfin's MoistureMeterSC (serving as a capacitance meter) and TEWL was topically measured using Delphin's Vapometer SWL5 (detecting changes in relative humidity) on children under the age of 18 with and without eczema, alongside subjective assessments on disease status[31].

Individuals with eczema have a damaged skin barrier, making their skin more sensitive to irritants and allergens, as well as reduced ability for the skin to retain water, making keeping the skin's moisture intact important in controlling eczema. A review paper by van Zuuren et al investigated the effect of emollients and moisturizers on eczema. A range of moisturizers were compared to placebo, vehicle, or no moisturizer. Participants considered moisturizers to be more effective for reducing eczema and itch than control. Moisturizers also led to lower investigator-assessed disease severity scores and fewer flares. Overall, , most moisturizers showed some beneficial effects; prolonging tie to flare, reducing number of flares and the number of topical corticosteroids needed to achieve similar reductions in eczema severity[32].

Psoriasis and ichthyoses are also skin diseases relating to an altered skin barrier function. Psoriasis is an autoimmune inflammatory disorder where skin cells are misidentified as pathogens by the immune system, leading to an incorrect signal transmitted that increases the growth cycle. It can be characterized by compromised barrier function, similar to eczema and atopic dermatitis (AD), contributing to its development through alteration of innate and adaptive immunity. The most common type is plaque psoriasis, which presents as red and white hues of scaly patches on the skin surface. This increases the number of activated T cells within the epidermis, causing the hyperproliferation of keratinocytes and an elevated turnover rate. This disease is managed by the application of systemic and topical medications and moisturizers containing salicylic acid. Ichthyoses occurs when there is a dysfunctional relationship between corneocytes and the extracellular matrix of the SC. This generates abnormal keratins that enlarge the SC thickness as it makes cells more adhesive. These defects in SC formation leads to water loss, causing dry and scaly skin. Most tools used to evaluate and assess psoriasis severity are based on subjective components. Objective measures for severity include transepidermal water loss (TEWL) and temperature. TEWL is stated to be higher and stratum corneum hydration lower in individuals with psoriatic plaques[33], [34].

Topical treatments are usually applied to the affected skin as the primary option to combatting the lesions that arise from psoriasis in mild to moderate cases. In cases where individuals have chronic plaque psoriasis, topical corticosteroids are used due to their higher potency. Different topical treatments include emollients, which are moisturizing treatments applied directly to skin to reduce water loss and cover it in a protective film, as well as steroid creams and ointments and vitamin D analogues. If topical treatments are not effective, phototherapy with UV light and systemic medications are then considered[35], [36].

3.1.2 Skin Hydration Measurement for Cosmetic Applications

The measurement of skin hydration is a critical aspect of the cosmetics industry, particularly in the development and evaluation of skincare products designed to enhance skin moisture and barrier function. Companies such as L'Oréal and others in the industry focus on creating hydrators and moisturizers tailored to general use and for individuals with specific skin conditions like eczema, psoriasis, xerosis, and sun damage. These products aim to improve skin hydration, reinforce the skin's barrier, and alleviate the symptoms associated with these conditions.

Clinical trials are essential for validating the efficacy of these products, using a variety of techniques to measure skin hydration. Among these, transepidermal water loss (TEWL) measurements are a widely used method to assess the skin's barrier function by quantifying the rate of water evaporating from the skin surface. Electrical-based methods, such as skin conductance and capacitance measurements, provide insights into the water content of the stratum corneum, the outermost layer of the skin. The Corneometer, a standard device in the cosmetics industry, measures skin capacitance as an indicator of hydration levels. This tool is particularly valued for its precision and reliability, helping companies substantiate product claims related to skin hydration and improve their formulations[31], [34].

The Corneometer's ability to provide accurate measurements of skin moisture allows cosmetic companies to refine their product formulations and ensure that moisturizers effectively enhance the skin's lipid barrier, promoting water retention and improving overall skin health. Advances in bioengineering and material science have further driven the development of more sophisticated moisturizers. These products often incorporate active ingredients such as ceramides, hyaluronic acid, and peptides, which are known for their ability to provide prolonged hydration and support barrier repair. Additionally, innovative technologies like encapsulation and controlled-release systems are increasingly being explored to improve the delivery and efficacy of active ingredients in skincare products. These advancements allow for targeted and sustained release of moisturizers, thereby enhancing their effectiveness in maintaining skin hydration over extended periods. The Corneometer device can be seen in Figure 3.1 below.

Figure 3.1 Corneometer CM825 device (Courage & Khazaka)

Moreover, the assessment of skin hydration in cosmetic applications is evolving with the development of new technologies and methodologies. For instance, imaging techniques like confocal microscopy and Raman spectroscopy are being integrated into hydration studies to provide a more detailed analysis of skin structure and hydration at a microscopic level. These advanced techniques not only offer more comprehensive data but also contribute to the creation of more effective and scientifically-backed skincare products.

As the cosmetics industry continues to evolve, the importance of accurate and reliable skin hydration measurement remains paramount. By leveraging both traditional methods like the Corneometer and emerging technologies, companies can better understand skin hydration dynamics, leading to the development of products that offer enhanced benefits for consumers.

This expanded focus on precise hydration measurement and advanced formulation techniques highlights the ongoing commitment within the cosmetics industry to improving skin health and developing innovative solutions for diverse skincare needs.

3.2 Body Hydration

3.2.1 Dehydration

Dehydration, while often viewed as a temporary inconvenience, is a condition with profound effects on the body, particularly when it becomes chronic or severe. At its core, dehydration represents an imbalance between fluid intake and loss, which can disrupt homeostasis and impair numerous physiological processes. The body relies on a delicate balance of fluids to maintain cellular function, organ integrity, and overall health. When dehydration occurs, this balance is disrupted, leading to both immediate and long-term consequences on body hydration[21].

Dehydration occurs when the body loses more fluids than it takes in, disrupting the balance of water and electrolytes essential for bodily functions. It can be classified into three types: isotonic dehydration, where there is a balanced loss of water and sodium; hypertonic dehydration, characterized by greater water loss leading to increased solute concentration and cellular dehydration; and hypotonic dehydration, where sodium loss exceeds water loss, causing water to move into cells and potentially leading to dangerous cellular swelling. Dehydration impacts body hydration by reducing overall fluid volume, which can lead to impaired cognitive and physical performance, as well as serious health complications such as kidney failure, cardiovascular strain, and in severe cases, hypovolemic shock. Proper management involves understanding the type of dehydration to ensure appropriate rehydration strategies, which are crucial for restoring fluid balance and preventing the potentially severe consequences of dehydration[37].

Body hydration, which refers to the total water content within the body, is integral to maintaining blood volume, regulating temperature, and facilitating the transport of nutrients and waste products. When the body loses more water than it takes in, through processes such as sweating, urination, and respiration, it enters a state of dehydration. The body's cells begin to lose water, causing them to shrink, which impairs cellular metabolism and enzyme activity. This cellular dehydration is particularly detrimental to tissues that require a high level of hydration for optimal function, such as the brain, kidneys, and muscles[20], [21].

The consequences of dehydration are widespread and can range from mild symptoms, such as headaches and fatigue, to more severe complications, including kidney failure and cardiovascular issues. In the UK, dehydration is a significant health

issue, especially among the elderly and those with chronic health conditions. It is estimated that dehydration is a contributing factor in many hospital admissions for older adults, often due to a diminished sense of thirst, which can lead to insufficient fluid intake. This lack of adequate hydration is associated with an increased risk of urinary tract infections, kidney stones, and even increased mortality in severe cases[24].

Dehydration also has a direct impact on body hydration levels, as it reduces the overall volume of fluids in the body. This reduction can lead to hypovolemia, a condition where the blood volume decreases, causing the heart to work harder to pump blood, and potentially leading to hypovolemic shock in extreme cases. Hypovolemia is particularly dangerous because it can impair the delivery of oxygen and nutrients to vital organs, leading to organ failure if not promptly treated[24].

The effects of dehydration extend beyond physical health, affecting cognitive function as well. Research has shown that even mild dehydration, characterized by a loss of just 1-2% of body weight, can impair cognitive abilities, including attention, memory, and critical thinking skills. This is particularly concerning for vulnerable populations, such as the elderly, who are more susceptible to dehydration and its effects. In these individuals, dehydration can lead to confusion, delirium, and an increased risk of falls and accidents[23], [38].

In terms of body hydration, maintaining adequate fluid levels is crucial for all aspects of physiological function. The body uses fluids not only to support cellular function but also to regulate body temperature through sweating and respiration. As dehydration progresses, the body's ability to regulate temperature diminishes, increasing the risk of heat-related illnesses such as heat exhaustion and heatstroke. This is particularly relevant for athletes and individuals who engage in strenuous physical activity, as they are more likely to lose significant amounts of fluid through sweat[39].

Addressing dehydration involves more than simply increasing water intake; it requires a comprehensive approach to managing body hydration. This includes monitoring fluid loss, particularly in populations at risk, and ensuring that hydration strategies are tailored to individual needs. For example, athletes may need to incorporate electrolyte solutions into their hydration regimen to replace not only water but also essential salts lost through sweat. Similarly, older adults may require more frequent reminders to drink fluids, particularly during hot weather or when ill[39].

In conclusion, dehydration is a serious condition that affects body hydration and overall health. It has widespread implications, particularly for vulnerable populations, and can lead to severe complications if not managed properly. The key to preventing dehydration lies in understanding the body's hydration needs and taking proactive steps to maintain fluid balance, whether through increased fluid intake, monitoring fluid loss, or using advanced hydration strategies tailored to individual needs. Continued research and public health initiatives are essential to raise awareness about the dangers of dehydration and to promote effective hydration practices across all age groups. These considerations emphasize the importance of hydration not just for immediate relief from symptoms but as a critical component of overall health management.

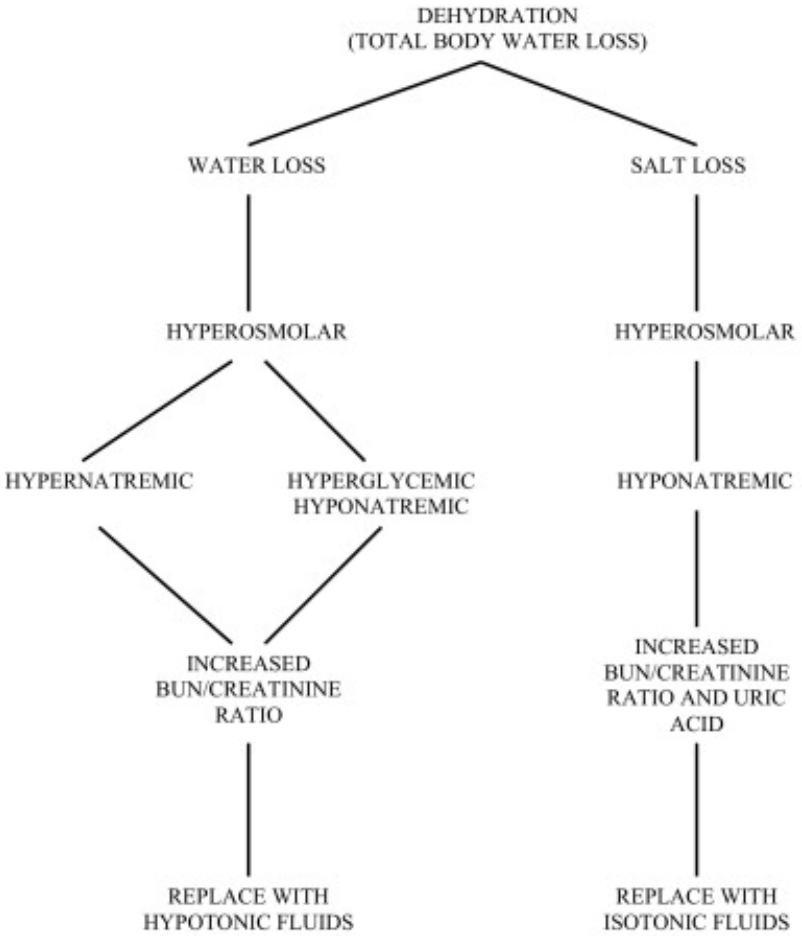


Figure 3.1: Flowchart illustrating types of dehydration[38]

3.2.2 Hydration Monitoring for Neonates & the Elderly

Hydration monitoring in clinical settings is paramount, particularly for vulnerable populations such as neonates and the elderly, who are at a higher risk of dehydration due to their unique physiological characteristics. For neonates, the risk of dehydration is heightened by their higher metabolic rates and greater water requirements per unit of body weight. This makes them particularly susceptible to fluid imbalances, which can rapidly lead to serious health issues. Dehydration in infants is often challenging to diagnose, as signs such as prolonged capillary refill time, abnormal skin turgor, and irregular respiratory patterns are indirect and may vary significantly among individuals. Given these challenges, the gold standard for diagnosing dehydration in neonates remains the comparison of body weight before and after rehydration. This method, although accurate, is not always feasible in real-time clinical situations. Consequently, clinical guidelines, such as those developed by the National Institute for Health and Care Excellence (NICE), recommend a combination of clinical symptoms for assessing dehydration and shock in infants[40].

In the case of the elderly, dehydration poses a significant risk due to age-related physiological changes, including a reduced sense of thirst and decreased renal function. These changes result in a decreased ability to conserve water and respond to fluid loss, making elderly individuals more vulnerable to dehydration even with minor fluctuations in water volume. Additionally, the total body water content in elderly individuals decreases by 10-15% as part of the natural aging process, further compounding the risk. Monitoring fluid intake is essential in this population to prevent dehydration and its associated complications. Current strategies often involve manually tracking fluid intake, but these methods are prone to inaccuracies and are often insufficient for ensuring adequate hydration[41].

Recent advancements in technology have led to the development of continuous hydration monitoring systems, which provide real-time feedback and encouragement to maintain adequate hydration levels. These systems are particularly beneficial for the elderly, as they can help reduce the incidence of dehydration-related complications such as urinary tract infections, kidney stones, and cognitive impairments. Research has shown that personalized hydration plans, when combined with continuous monitoring, can significantly improve hydration status and overall health outcomes in the elderly[41].

The use of wearable hydration monitoring devices is also being explored as a means to provide continuous, non-invasive assessment of hydration status in both neonates and elderly patients. These devices, which often rely on bioimpedance and other non-invasive technologies, offer the potential for more accurate and timely detection of dehydration, allowing for earlier intervention and better management of hydration. Wearable devices can be particularly useful in clinical settings where regular monitoring is required, as they can reduce the burden on healthcare providers and improve patient outcomes by ensuring that hydration levels are maintained within optimal ranges[42].

Moreover, integrating these wearable devices with other health monitoring systems can provide a comprehensive approach to managing hydration in vulnerable populations. For example, in neonates, combining hydration monitoring with other vital signs such as heart rate and respiratory rate can offer a more complete picture of the infant's health status, enabling more precise interventions. In the elderly, these devices can be integrated with fall detection systems and cognitive monitoring tools to provide a holistic approach to care, addressing the multifaceted challenges associated with aging[41].

In summary, while traditional methods of hydration monitoring remain important, the integration of advanced technologies, such as wearable devices and continuous monitoring systems, represents a significant step forward in the management of dehydration in neonates and the elderly. These innovations not only improve the accuracy and timeliness of dehydration detection but also support more personalized and effective hydration strategies, ultimately leading to better health outcomes for these vulnerable populations. Continued research and development in this area are essential to further refine these technologies and expand their accessibility in clinical settings.

3.2.3 Hydration Monitoring in Athletics & Fitness

Hydration monitoring is crucial in athletics and fitness, as maintaining proper hydration is essential for both optimal physical and cognitive performance. Athletes, particularly those engaged in high-intensity or endurance activities, are at a heightened risk of dehydration, which can significantly impair their performance. Dehydration reduces blood volume, increases cardiovascular strain, and limits the body's ability to regulate temperature, all of which can lead to heat-related illnesses such as heat exhaustion and heatstroke. Moreover, even mild dehydration can negatively impact strength, endurance, and cognitive function, emphasizing the importance of monitoring hydration levels closely during training and competition[43], [44].

Wearable hydration monitoring devices have become increasingly popular among athletes, offering real-time feedback that can be used to optimize hydration strategies. These devices often use sweat-based sensors to monitor electrolyte levels, providing insights into fluid and salt loss during exercise. Other devices employ electrical or optical sensors to assess cellular hydration levels, offering a more comprehensive understanding of an athlete's hydration status. By integrating these technologies into their training routines, athletes can receive personalized hydration recommendations that account for their activity levels, environmental conditions, and individual sweat rates. This personalized approach helps ensure that athletes maintain euhydration—optimal body water levels—thereby enhancing endurance, reducing recovery times, and minimizing the risk of dehydration-related complications[45].

Traditional methods for assessing hydration, such as monitoring urine colour, frequency, and volume, as well as tracking changes in body mass to estimate sweat rate, remain widely used. These methods, while simple and effective for general monitoring, may not always provide the precision needed for high-performance athletes. More advanced techniques, such as measuring blood volume and plasma osmolality, offer a more accurate assessment of pre-exercise hydration status and are often used in professional sports settings. These measures can help athletes fine-tune their hydration strategies to match the demands of their specific sport or activity, ensuring they are optimally hydrated before, during, and after exercise[43].

Bioimpedance analysis (BIA) is another non-invasive method for estimating total body water content. While BIA may not detect small changes in hydration status, it

provides a useful foundation for developing more affordable and accessible real-time hydration sensors. Advances in sensor technology, combined with sophisticated data analytics, are leading to the creation of more accurate and user-friendly hydration monitoring devices. These innovations enable athletes to receive real-time data on their hydration status, allowing for immediate adjustments to their hydration strategies during training or competition. The ability to tailor hydration plans based on real-time data is particularly valuable for athletes who compete in varying environmental conditions or engage in prolonged physical activity[22].

Looking ahead, the integration of hydration monitoring into wearable fitness devices is expected to become even more sophisticated. Future advancements could include more sensitive sensors capable of detecting minute changes in hydration status, as well as improved algorithms that can predict an athlete's hydration needs based on factors such as exercise intensity, duration, and environmental conditions. These developments hold significant promise for enhancing athletic performance, reducing the risk of dehydration, and improving overall recovery times. As the technology continues to evolve, athletes of all levels may benefit from more precise and effective hydration strategies that are tailored to their individual needs[45].

In conclusion, hydration monitoring in athletics is a critical aspect of performance optimization. By leveraging both traditional methods and cutting-edge technologies, athletes can maintain proper hydration, which is essential for achieving peak performance and preventing dehydration-related complications. As research and technology advance, the ability to monitor and manage hydration in real-time will become increasingly accessible, offering significant benefits for athletes across all levels of competition[39].

3.3 Summary

This chapter comprehensively explored the motivations and various applications for measuring hydration, emphasizing its critical importance across different populations and scenarios. Dehydration, defined as the condition where the body loses more fluids than it takes in, was discussed in terms of its primary forms: hyperosmolar dehydration, resulting from water loss, and hyponatremia, involving the loss of both salt and water. Maintaining proper hydration is crucial as it affects numerous bodily functions, including cognitive performance, thermoregulation, and cardiovascular health. The regulation of fluid levels is primarily controlled by osmoreceptors in the hypothalamus, which stimulate thirst and the release of antidiuretic hormone (ADH) to maintain homeostasis.

The chapter highlighted that as individuals age, the sensation of thirst becomes less reliable due to declining sensitivity of osmoreceptors and other physiological changes. This decline is particularly significant for athletes, where insufficient hydration can impair performance, thermoregulation, and sweat production. Dehydration equivalent to a 2% loss in body weight can significantly affect physical and cognitive performance, underscoring the need for effective hydration strategies that consider individual requirements and environmental conditions.

The discussion expanded to the relationship between skin hydration and various skin conditions. Several skin conditions and diseases, such as xerosis, eczema (atopic dermatitis), psoriasis, and photodamaged skin, are closely linked to impaired skin hydration and barrier function. These conditions often require the use of moisturizers and topical treatments for effective management. The Corneometer was identified as the gold standard for measuring skin hydration, providing essential data for assessing the efficacy of these treatments. This device, along with others, plays a crucial role in the cosmetics industry, where accurate measurement of skin hydration is vital for the development and evaluation of skincare products. Techniques such as visual analysis, transepidermal water loss (TEWL) measurements, and electrical-based methods like conductance are used to assess product efficacy. Advances in bioengineering and material science have further led to the development of sophisticated moisturizers that provide prolonged hydration and enhanced barrier repair.

Hydration monitoring is particularly critical in clinical settings, especially for neonates and the elderly, who are more susceptible to dehydration. Infants are at higher

risk due to their higher metabolic rates and greater water requirements per unit of body weight, making accurate and timely monitoring essential to prevent severe complications. In contrast, elderly individuals face dehydration risks due to physiological changes associated with aging, such as reduced thirst sensation and decreased renal function. Monitoring fluid intake, along with the use of real-time systems that provide feedback, can significantly reduce the incidence of dehydration-related complications in these populations. The development of wearable hydration monitoring devices offers a promising approach to continuous, non-invasive assessment, enhancing the management of hydration in these vulnerable groups.

Athletes also benefit significantly from hydration monitoring, as proper hydration is critical for optimal performance. Wearable devices that monitor hydration in real-time help athletes optimize their hydration strategies, thereby enhancing performance and recovery. Traditional methods for assessing hydration status, such as urine analysis and changes in body mass, have been complemented by advanced techniques like bioimpedance analysis (BIA) and sweat-based sensors. These technologies provide personalized hydration recommendations, improving athletes' endurance, reducing the risk of heat-related illnesses, and aiding in faster recovery.

In summary, this chapter detailed the importance of accurate hydration measurement across various contexts, including clinical settings, athletic environments, and cosmetic applications. Advances in technology, such as wearable sensors, optical techniques, and personalized hydration strategies, hold significant promise for improving hydration monitoring and management. These advancements can lead to better health outcomes, enhanced performance, and overall improved quality of life. The continued development of these technologies, particularly in terms of accessibility and accuracy, is essential for supporting hydration management across all populations, from vulnerable individuals to high-performance athletes.

CHAPTER 4:

RELEVANCE OF DOMINANT O-H BANDS USING NEAR-INFRARED SPECTROSCOPY IN HEALTHCARE AND WELLBEING

4.1 Characteristics of O-H bands in NIR

The O-H (hydroxyl) bands in the near-infrared (NIR) region are of particular importance due to their strong absorption characteristics, which are highly relevant in a range of biomedical and healthcare applications. These O-H bonds, primarily associated with water molecules, exhibit several overtone and combination bands in the NIR spectrum, typically spanning from 700 nm to 2500 nm. The most prominent absorption peaks occur around wavelengths of 970 nm, 1190 nm, 1450 nm, and 1940 nm, where water molecules strongly absorb NIR light. This absorption is a result of the vibrational transitions of the O-H bonds when photons in the NIR spectrum interact with water molecules. The energy from the photons causes the O-H bonds to stretch and bend, leading to distinct absorption peaks that can be observed on the NIR spectra. These peaks represent the energy levels at which water absorbs the most, making the O-H bands critical markers for assessing tissue hydration and other water-related properties in biological systems[46], [47]. Although Figure 4.1 illustrates the water spectrum of methanol, it also can be used to display the change in the spectral peaks of these O-H and C-H bands in response to changes in water concentration.

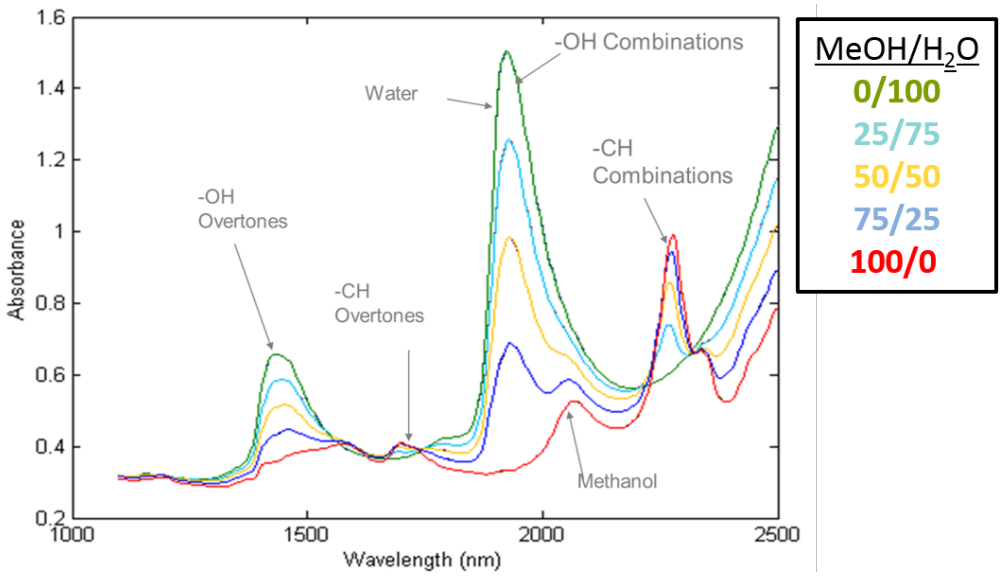


Figure 4.1: Absorption bands of varying water-alcohol (MeOH) solution concentrations in the NIR region

Water, as the most abundant molecule in biological tissues, plays a crucial role in NIR spectroscopy. The absorption by the O-H bands can indicate the presence and distribution of water within tissues, reflecting both physiological and pathological states. For instance, the absorption at 1450 nm is often used to assess the hydration levels of the

stratum corneum in skin studies, providing insights into skin health. Similarly, the absorption band around 1940 nm, which corresponds to the vibrational transitions and combination of O-H stretching and H-O-H bending, is highly sensitive to water content and is commonly used to measure hydration in more water-rich tissues like the cornea or muscles. These are primarily found in water molecules, which absorb NIR light at these wavelengths, enabling NIR spectroscopy to effectively measure hydration levels in biological tissues, making it a valuable tool in both clinical and research settings[46], [48].

These O-H absorption bands are also indicative of the molecular structure and bonding environment of water within tissues. The differentiation between free water, which is abundant in extracellular spaces, and bound water, which is closely associated with proteins and cellular structures, is particularly important. Depending on whether water is free, bound to proteins, or confined within cellular structures, the precise location and intensity of these bands can vary, providing insights into the hydration status and molecular dynamics within the tissue. For example, in tissues where water is tightly bound, the O-H absorption bands might shift slightly, reflecting changes in the hydrogen bonding environment. This makes the O-H bands particularly useful for distinguishing between different hydration states at a molecular level, offering a deeper understanding of water's role in maintaining cellular function and tissue health[49].

Besides O-H bonds, other absorbers in the NIR region include C-H and N-H bonds, which are found in various organic molecules within the body. These bonds also exhibit overtone and combination bands in the NIR spectrum, although their absorption is generally weaker compared to the strong absorption by O-H bonds. The presence of these additional absorbers, alongside water, contributes to the complex spectra observed in biological tissues, allowing NIR spectroscopy to provide comprehensive information about tissue composition and hydration[50].

In summary, the O-H bands in NIR spectroscopy are invaluable for non-invasive diagnostics, offering precise assessments of hydration and tissue health. The ability to detect and analyse these specific water absorption bands makes NIR spectroscopy a powerful tool in biomedical research, clinical diagnostics, and the monitoring of conditions such as dehydration, oedema, and skin disorders. The technology's capacity to differentiate between various hydration states and its sensitivity to molecular changes further underscore its significance in healthcare applications.

4.2 Light-Tissue Interactions

Understanding light-tissue interactions is crucial for interpreting NIR spectroscopy data, especially when analysing the O-H bands associated with water molecules. When near-infrared (NIR) light interacts with biological tissues, several key processes occur, including absorption, scattering, and reflection. These processes are influenced by the specific chromophores present in the tissue, with water being a dominant chromophore due to the significant presence of O-H bonds. The absorption of NIR light by these O-H bonds is central to determining the water content within tissues, as these bonds have strong absorption characteristics at specific wavelengths, particularly around 970 nm, 1190 nm, 1450 nm, and 1940 nm. The interaction of NIR light with water molecules results in vibrational transitions of the O-H bonds, leading to the absorption of photons and the generation of distinct absorption peaks in the NIR spectrum. These peaks are indicative of the presence and concentration of water, which is why they are so valuable in assessing tissue hydration[14], [47].

Scattering in tissues, on the other hand, is influenced by the tissue's microstructure, including the size and density of cells, collagen fibers, and other macromolecules. This scattering affects how deeply NIR light can penetrate and how it is reflected back to the detectors. For instance, the density and arrangement of collagen fibers can significantly alter the scattering properties of the tissue, thereby affecting the overall light-tissue interaction. The balance between absorption and scattering is critical for accurately interpreting NIR spectroscopy data, as it influences the penetration depth of NIR light and the resolution of the resulting measurements[14].

NIR light is capable of penetrating deeper into tissues compared to visible light, typically reaching depths of several millimetres to centimetres, depending on the specific wavelength used and the type of tissue being examined. This deep penetration is particularly useful in medical diagnostics, as it enables the non-invasive monitoring of tissue composition and hydration at various depths. For example, in cerebral imaging, NIR spectroscopy is employed to monitor brain oxygenation and hydration levels by detecting the absorption characteristics of both water and haemoglobin. The ability to reach deeper tissues without invasive procedures makes NIR spectroscopy an invaluable tool in various clinical settings[51].

The scattering and absorption properties of tissues in the NIR region also vary with the physiological state of the tissue. For example, during dehydration, the water content in tissues decreases, leading to a reduction in the absorption at the O-H band wavelengths. This change can be detected and quantified using NIR spectroscopy, providing a non-invasive method to monitor hydration levels in both clinical and wellness settings. The sensitivity of NIR spectroscopy to these changes in water content makes it a powerful tool for tracking hydration, whether in patients recovering from surgery or athletes managing their hydration status during intense physical activity[14].

Moreover, the interaction of NIR light with water in tissues impacts the thermal properties of the tissue. Absorption of NIR light by water molecules can cause localized heating, which is particularly relevant in therapeutic applications such as laser treatments, where controlled heating is used to target specific tissues. This thermal effect must be carefully managed to avoid damage to healthy tissue, but it also presents opportunities for therapeutic interventions that leverage controlled heating[52].

Additionally, the absorption and scattering properties of NIR light are crucial for imaging techniques like optical coherence tomography (OCT) and diffuse optical tomography (DOT). These techniques exploit the differential absorption and scattering of NIR light to produce detailed images of tissue hydration and structure. Such imaging is essential for diagnosing and monitoring conditions like tumours, where abnormal water content and distribution are often early indicators of pathology. By understanding and refining these light-tissue interactions, researchers and clinicians can develop more accurate and non-invasive diagnostic tools that monitor critical physiological parameters in real-time, enhancing the effectiveness of both diagnostic and therapeutic applications[53].

4.3 Applications Using Dominant O-H bands

The dominant O-H bands in near-infrared (NIR) spectroscopy have been extensively leveraged across a wide range of healthcare and wellness applications due to their sensitivity to water content in biological tissues. This capability makes NIR spectroscopy an invaluable tool for monitoring various physiological and pathological conditions. In dermatology, NIR spectroscopy is widely used to evaluate skin hydration

levels by analysing the absorption at the 1450 nm O-H band, which is indicative of water content in the stratum corneum. This application is particularly valuable in the development and testing of skincare products, where precise measurement of skin hydration is essential. By quantifying the water content in the skin, researchers can assess the efficacy of moisturizers and other topical treatments, ensuring that they provide the intended hydrating effects[54].

Beyond dermatology, NIR spectroscopy's application extends to monitoring hydration status in clinical settings, especially for vulnerable populations such as neonates and the elderly. The ability to non-invasively monitor fluid balance through O-H band absorption changes helps healthcare providers prevent complications associated with dehydration or overhydration, both of which can have severe consequences in these groups. Additionally, NIR spectroscopy is used to assess hydration and oxygenation levels in muscle and brain tissues, providing critical information for managing conditions like cerebral oedema and muscle fatigue. These applications are particularly important in intensive care settings, where precise fluid management is crucial for patient outcomes[55], [56].

Recent advances in wearable NIR devices have further expanded the use of O-H band monitoring beyond clinical environments into everyday wellness applications. These wearable sensors enable continuous monitoring of hydration levels, providing real-time feedback that is especially valuable for athletes, individuals in extreme environments, or those managing chronic health conditions. The integration of NIR spectroscopy with other sensing technologies, such as bioimpedance or temperature sensors, enhances its utility by enabling comprehensive monitoring of health and hydration status in diverse settings[57].

The dominant O-H bands in NIR spectroscopy also have significant applications in oncology, particularly in cancer detection and monitoring. Tumours often exhibit altered hydration levels due to abnormal vascularization and cellular activity. NIR spectroscopy can detect these changes by measuring variations in water content, offering a non-invasive method for identifying and monitoring tumours. This technique is especially useful in distinguishing between malignant and benign tissues, as well as guiding surgical resection by providing clear delineation of tumour margins[58].

In wound healing, particularly in cases of burns and surgical flaps, monitoring tissue hydration using NIR spectroscopy is critical. Adequate hydration is essential for

tissue regeneration, and NIR spectroscopy allows clinicians to assess hydration levels in healing tissues, thereby facilitating timely interventions to improve outcomes. This application is particularly relevant in reconstructive surgery, where maintaining the appropriate hydration levels in grafts and flaps can determine the success of the procedure[59], [60].

Moreover, NIR spectroscopy is being explored for its potential in non-traditional applications, such as alcohol detection and glucose monitoring. By measuring the water content in exhaled breath, NIR spectroscopy can be used to estimate blood alcohol levels, providing a non-invasive alternative to traditional breathalysers. Similarly, NIR spectroscopy's sensitivity to hydration levels in tissues is being investigated as a potential tool for non-invasive glucose monitoring, which could offer a significant advancement in diabetes management[61], [62].

Beyond healthcare, the O-H bands in NIR spectroscopy have diverse applications in other fields. In agriculture and food science, they are used to assess moisture content in crops and food products, which is essential for ensuring quality and preventing spoilage. In the pharmaceutical industry, these bands help monitor hydration levels in drug formulations, optimizing product stability and efficacy. Environmental monitoring also benefits from NIR spectroscopy, particularly in measuring soil moisture to support sustainable agricultural practices. Additionally, in material science, the O-H bands are crucial for studying moisture content in polymers and textiles, affecting material properties and durability. Furthermore, in biological research, these bands are utilized to investigate water content in tissues and cells, contributing to a deeper understanding of physiological processes and disease mechanisms[63].

In conclusion, the dominant O-H bands in NIR spectroscopy offer a powerful tool for a wide range of applications in healthcare, wellness, and beyond. Their sensitivity to water content makes them indispensable in monitoring hydration and tissue health, providing a non-invasive, real-time method for assessing and managing health across various populations and scenarios. As technology continues to advance, the applicability and accuracy of these methods are expected to grow, further enhancing their role in medical diagnostics, treatment, and other industries.

4.4 Summary

The chapter provided an in-depth exploration of the importance of O-H bands in near-infrared (NIR) spectroscopy, highlighting their critical role in various healthcare and wellness applications. The O-H bands, linked to the hydroxyl groups in water molecules, exhibit strong absorption in the NIR spectrum, making them essential for assessing water content and tissue hydration. These bands are particularly effective in non-invasive diagnostic techniques, where changes in the molecular structure and hydration levels of tissues are monitored using NIR light.

The chapter also examined the interactions between NIR light and biological tissues, focusing on how absorption and scattering are influenced by the presence of water and other chromophores. These interactions are crucial for the success of NIR-based imaging and diagnostic tools, which are used to monitor hydration and detect pathological changes in tissues. The ability of NIR light to penetrate deep into tissues makes it especially valuable for assessing hydration at various depths, providing critical insights for both clinical diagnostics and therapeutic applications.

Beyond healthcare, the chapter discussed the broader applications of O-H bands in fields such as agriculture, food science, pharmaceuticals, and environmental monitoring. In these areas, NIR spectroscopy is employed to assess moisture content, ensure product quality, and study material properties. The versatility of O-H bands underscores their significance across multiple domains, illustrating the wide applicability of NIR spectroscopy in both research and practical applications.

Overall, this chapter emphasized the relevance of O-H bands in NIR spectroscopy as a powerful tool for monitoring hydration and water content across diverse fields. The ongoing advancements in NIR technology and the understanding of light-tissue interactions are expected to enhance the precision and utility of these methods, contributing to improved health outcomes and progress in various industries.

CHAPTER 5:

REVIEW OF TECHNIQUES AND ADVANCES IN THE MEASUREMENT OF HYDRATION BASED ON SENSING OF OPTICAL AND ELECTRICAL TISSUE PROPERTIES

There are multiple techniques that can be used for the assessment of skin hydration[37]. Methods based on the electrical properties of tissue and water (henceforth termed as “electrical methods” or “electrical sensing” in this paper), such as the measurement of skin capacitance and conductance levels, are based on the concept that alterations in the electrical properties of the stratum corneum layer signify differing skin hydration levels. An alternative sensing approach involves interrogating the optical properties of tissues (henceforth termed as “optical methods” or “optical sensing” in this paper), including the use of near-infrared spectroscopy (NIRS). The latter involves the analysis of spectra within the near-infrared wavelength range, since intensities in the NIR spectrum of the skin consist of absorption bands directly related to its water content.

The aim of this section is to comprehensively review the recent advances in developed sensors and wearable devices that utilize electrical and/or optical techniques to non-invasively measure skin hydration levels. The concept of combining techniques for multi-modal properties will be discussed and comparisons will be drawn on the advantages and disadvantages of both methodologies. In this review, there will primarily be a focus on experimental studies and a coverage of the current state of the art within the area of non-invasively measuring skin hydration. As mentioned, the motivation of this paper is to focus on the differences in measurement techniques in relation to accuracy, both in isolation and when combined. The idea of these devices being wearable will aid the real-time usage and provide a more portable and widely used device for use in various applications relating to skin hydration. Figure 5.1 below illustrates a timeline detailing the chronological evolution of the fundamental techniques used in the measurement of skin hydration[38], [64].

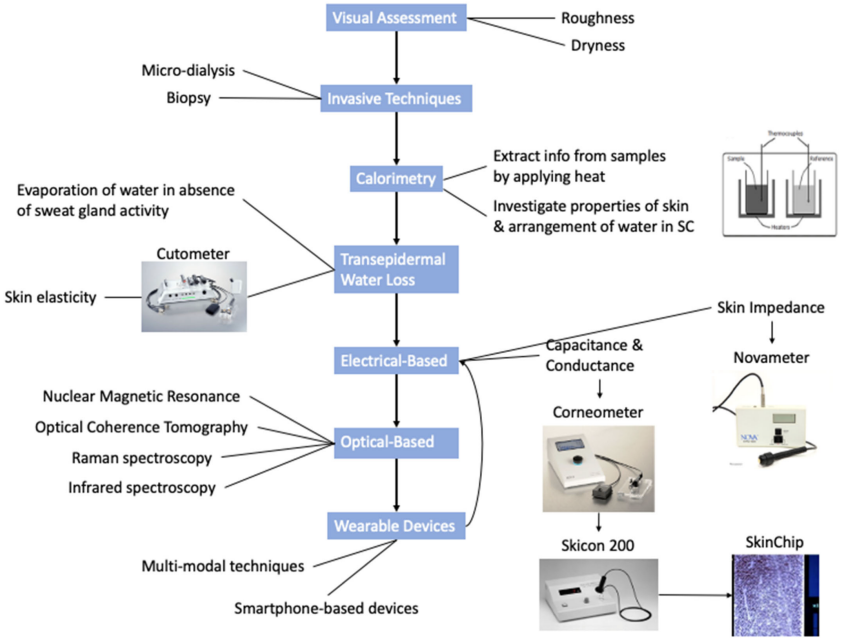


Figure 5.1: Flowchart illustrating the timeline of the chronological evolution of skin hydration measurement techniques

5.1 Electrical Techniques for Measuring Skin Hydration

5.1.1 Use of Skin Capacitance

Both skin capacitance and skin conductance have been shown to provide a good estimation for detecting high or low skin water content levels. Multiple devices developed using similar methodological ideas but different sensor designs, such as the SKICON and Moisture Evaluator, have been found to follow similar trends and exhibit high correlations when compared to the Corneometer device. However, capacitance-based devices like the Corneometer have decreased sensitivity for monitoring high hydration values, whereas conductance-based devices have decreased sensitivity for low values. These limitations arise from the distinct physical principles governing capacitance and conductance measurements. Capacitance measurements, which rely on the dielectric properties of the skin, become less responsive at higher hydration levels due to saturation effects, while conductance measurements, based on ionic movement, struggle with accuracy at low hydration levels because of reduced conductivity in drier skin. As a result, neither method provides comprehensive sensitivity across the full hydration spectrum, indicating that electrical techniques alone may not be sufficient for reliably and accurately measuring hydration levels under all conditions.

Skin properties such as hydration can be acquired by analysing electrical parameters. These techniques consider a basic electrical model where the skin acts as a resistor placed parallel to a capacitor. This model can calculate impedance values, and other electrical parameters such as capacitance and conductance can also be established via electrical-based sensing devices using electrodes in combination with an applied alternating current.

Most commercial devices are based on capacitive measurement techniques, with the gold standard being the Corneometer. These devices record capacitance via two charged electrode plates creating an electric field, with the maximum charge produced being the measured capacitance value. A dielectric medium, such as skin placed between the charged electrodes, allows the capacitance to vary according to its permittivity. Since water has a high dielectric constant of 81, it enhances the capacity of a capacitor, implying that skin water content is directly proportional to skin capacitance. However, while these devices provide useful information about overall skin hydration levels, they may not fully

capture localized variations in hydration across different areas of the skin's surface or account for microstructural features, which can influence hydration distribution.

A study by Logger et al. used capacitance as its measurement technique to investigate anatomical site variation of water in the stratum corneum layer. The Epsilon device, which records skin capacitance and is similar to the Corneometer but with multiple sensors, was used. Its multi-sensor design allows for simultaneous measurements. Significant differences in water content were found with large inter-individual variations, the largest being in the cheek and the smallest in the mid-calf region. The resulting values from the Epsilon device were lower than conventional Corneometers but followed a similar trend with $P < 0.001$[65].

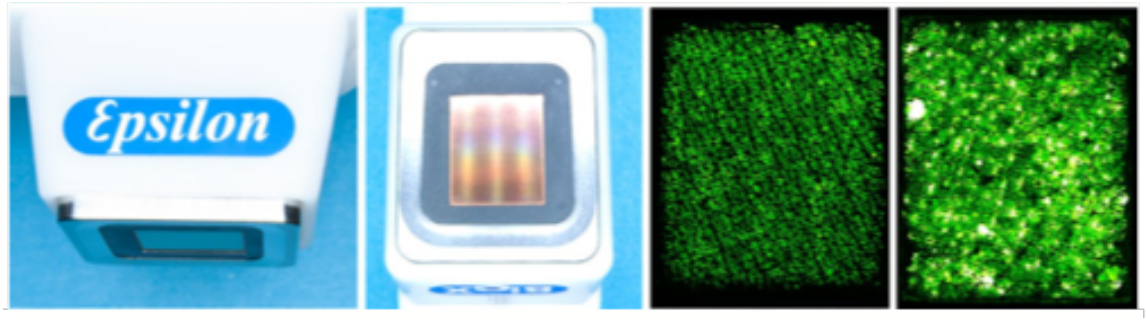


Figure 5.2: A, Close-up of the Epsilon measurement head with the metal bezel. B, The sensor surface embedded in an epoxy frame. C, Typical contact image of the inner forearm skin. D, A contact image of the skin in the face with visible sweat gland activity[65]

In a novel use of capacitive measurement principles, the Skin Hydration Sensor Patch (SHSP) developed by Flament et al. uses a user's smartphone to measure skin moisture wirelessly. It combines capacitive techniques and NFC technology to allow self-recording of skin hydration via a probe attached to the back of a phone. A study comparing this device to the Corneometer showed a high correlation of $r = 0.55$ and $p < 0.0001$. A further in vivo experiment consisted of participants with dermatologically assessed moderate dry skin, following the application of a hydrating Xanthane-based gel. The results showed that the values obtained from the SHSP had a strong positive correlation when compared to the Corneometer device output. In terms of the differences in hydration between differing skin sites, the face was found to have a lower hydration level than the forearm, with recordings from the face exhibiting a higher variability. Both skin sites presented similar trends subsequent to the application of the gel product, with hydration levels displaying a significant increase followed by a progressive decline in

hydration across time. The hydration of the face expressed a higher amplitude, suggesting a differing requirement for hydration dependent on the skin region[66].

Another device that obtains capacitive measurements via an imaging technique called in vivo mapping is known as the SkinChip® (L'Oréal, Paris, France). Developed by D Batisse et al., the device works by using the capacitance method to obtain components from the grey-level histogram of skin images to provide a non-optical representation of skin hydration [40]. Results from experiments displayed a linear correlation that is shown to be highly significant between the Corneometer recordings and the grey levels measured by the SkinChip device. Such devices have the ability to express the texture diversity of the skin surface, enabling inhomogeneity of skin hydration to be studied[67].

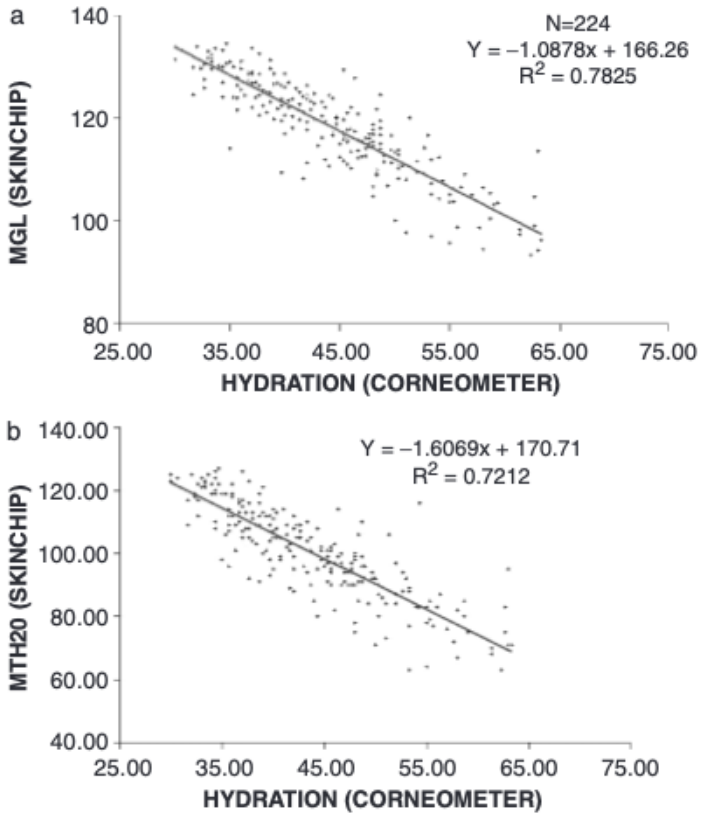


Figure 5.3: Linear correlation between the average grey level of the histograms of each image (a), the grey level obtained on histograms with a 20% threshold on each image (b) and the measurements obtained by the Corneometer. Forearm measurements, N=224.[67]

5.1.2 Use of Skin Conductance

Conductance, the inverse of resistance, is defined by the ability required for an electric current to pass through a substance, and thus conduct electricity. By applying Ohm's law, the conductance (G) can be calculated as:

$$G = \frac{i}{v} = \frac{1}{R}$$

The skin conductance response (SCR), also referred to as electrodermal activity (EDA), measures increased activity detected by the body's sympathetic response system in response to a stimulus. This response measures skin conductance, which varies with moisture levels. Skin momentarily alters its conductivity depending on internal or external stimuli, and this measure increases with higher water content in the stratum corneum. Electrical resistance, being the reciprocal of conductance, is concentrated 99% at the skin and varies with location and hydration levels. Measuring electrodermal activity via skin conductance variations is suggested to be a good indicator of differing hydration levels. However, there is high sensitivity to stimuli for the nervous system, such as fear and temperature, which can affect recordings of hydration levels. An example of a developed device using the conductance response method is the Skicon, displayed in Figure 5.4 below[68].

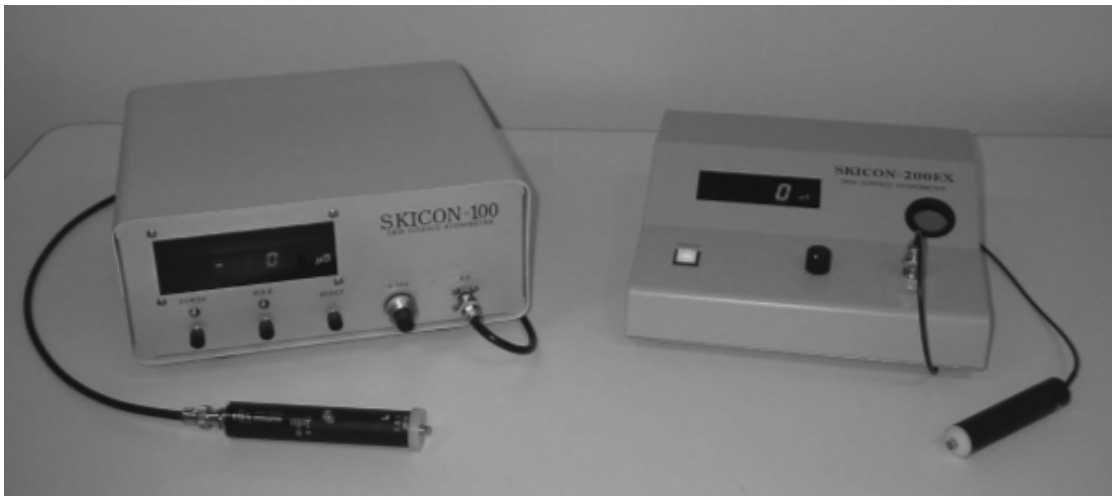


Figure 5.4: Original Skicon-100 (left) and the latest model, Skicon-200EX (right)[68]

Skin conductance, in addition to transepidermal water loss (TEWL) and skin elasticity, can be examined as an index of the barrier function of skin. A study by Nishimura et al. investigated the effect of fine water particles on the moisture and viscoelasticity of facial skin. Skin conductance was recorded using the SKICON TEWL by the Vapometer and skin distention by the Cutometer at the cheek every 60 minutes. Participants were tested under three conditions with water particles of different sizes. It was found that skin conductance of the stratum corneum was higher with smaller water particles. At a 120-minute interval after spraying the water particles, conductance was significantly increased compared to the baseline under all conditions. These water particles, being small in diameter and non-charged, can permeate intercellular spaces in the epidermal layer to the dermal layer, maintaining water retention[69].

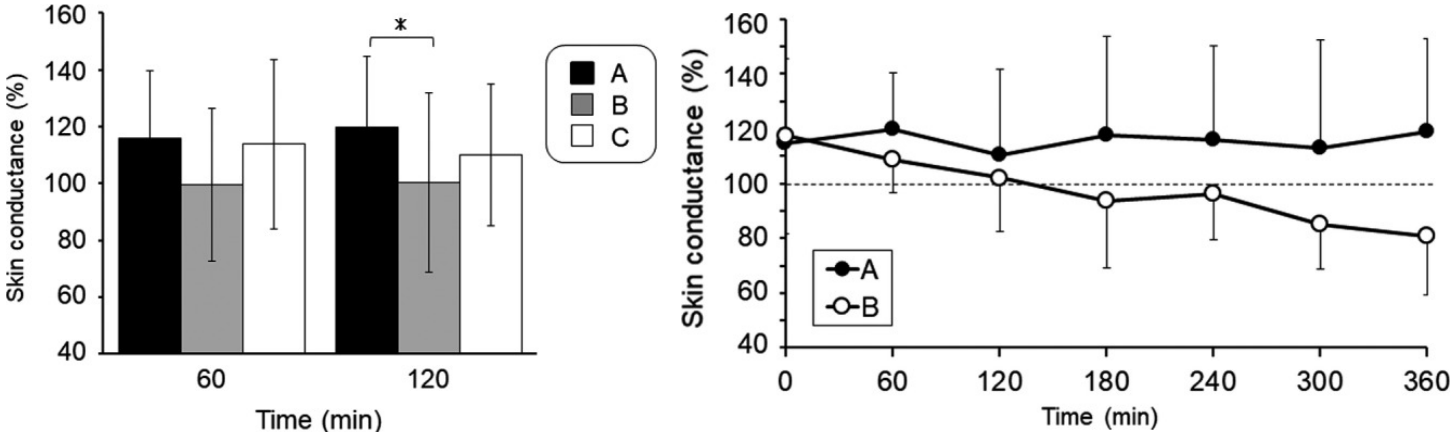


*Figure 5.5: left) Effect of spraying fine water particles on the skin conductance of the facial skin. *(P<0.05). right) Time course of skin conductance of the facial skin after spraying fine water particles[69]*

Additionally, T Andre et al. developed a conductance-based device known as the Moisture Evaluator to directly measure skin hydration during object manipulation. The probe consists of gold-covered electrodes connected to a resistor-capacitor circuit to measure hydration levels based on conductance. Significant correlations to the Corneometer device were found, indicating its efficacy[70].

5.1.3 Use of Bio-Impedance

The electrical impedance of the skin depends primarily on temperature, voltage, and frequency. Impedance varies as a function of frequency, and when the stratum

corneum has low water content, it acts as a dielectric medium. With increased water content, it becomes more responsive to an electric field while maintaining its barrier function. The formula to calculate impedance using Ohm's law is: [71]

$$Z = \frac{\mathrm{V}}{\mathrm{I}} \qquad \textit{Equation (5.1)}$$

When electrical current flows through biological material, the opposing resistance is referred to as bioimpedance. The frequency of biological material depends on reactive and resistive characteristics occurring in the spectra. The recorded impedance of the electrode is high at low frequencies, declining with an increase in detected frequency.

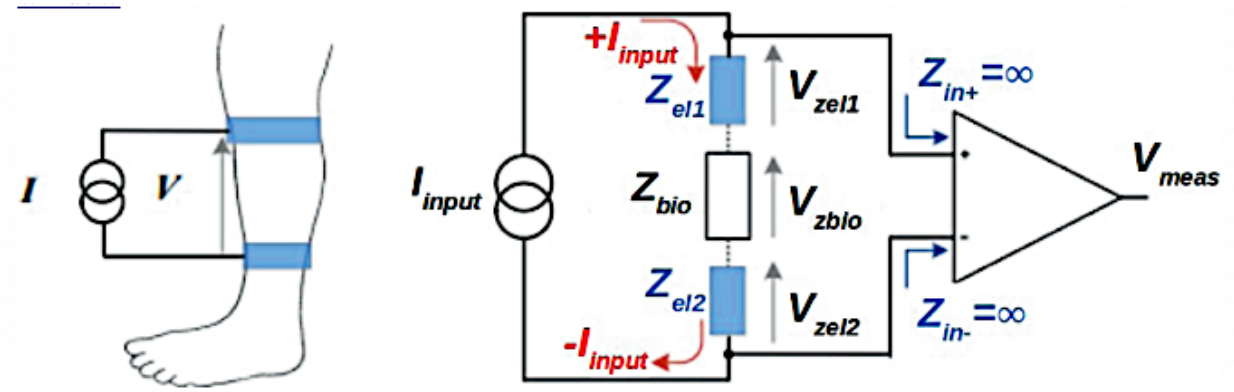


Figure 5.6: Circuit of a bipolar bioimpedance technique, where Iinput is injecting current, Zel is electrode impedance, Zbio is biological sample, Zin is input impedance of the measuring amplifier, Vzel and Vzbio are the voltage across the electrode and sample[71]

An example of a skin hydration measurement device using impedance to obtain skin electroconductivity measures is the Nova Dermal Phase Meter or Novameter (DPM 9003). Skin water content is provided as a function of the skin's dielectric constant value, and impedance is obtained once equilibrium is achieved[72].

The bio-impedance method can be used in skin hydration and barrier function experiments through its measurement principle of the electroconductivity of the skin. An example study by L Davies et al. produced a three-dimensional cell model using CAD software formed of 10 thin layers of hexagonal cells acting as the stratum corneum and 4 thick layers of epidermal cells below. Electrical properties were acquired by applying an electric field to cells suspended in a conducting medium, recording the resulting impedance. An input sinusoidal voltage varying between 0 and 10 volts was tested at

different frequencies between 100 Hz and 10 MHz Conductivity response to skin hydration was simulated at varying conductivities. Results showed that at low frequencies, conductivity had no effect on overall impedance, but at frequencies exceeding 100 kHz, impedance decreased with lower conductivity. This model supports the idea that increasing skin hydration improves clinical results by reducing impedance[73].

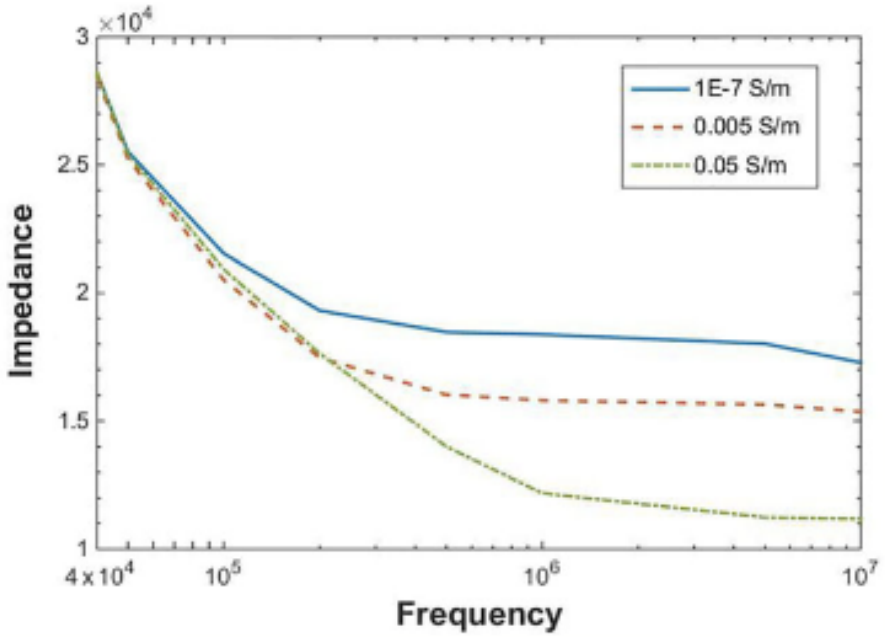


Figure 5.7: Change Impedance due to varying SC cytoplasm conductivities between 0.005 and 0.05 Sm-1 at 5 signal frequencies[73]

Another study assessed body hydration on cyclists using bioimpedance. A baseline bioimpedance measure was taken, followed by hypo-hydration and rehydration measures after cycling with low water intake. Results showed a correlation between hydration and measured resistance, with sensitivity up to around a 700ml change in hydration across multiple tests[73].

Matsukawa et al conducted an in vivo experiment where skin impedance was measured on participants via nanomesh electrodes connected to an LCR meter to monitor skin hydration levels. The use of nanomesh electrodes has many advantages over standard devices that use rigid electrode probes, such as in the Corneometer, which require being in direct pressure with the skin. This can inadvertently alter hydration levels due to deformation of the skin surface. Additionally, the use of biocompatible electrodes that do not need pressure applied to the skin is considerate in terms of usage by individuals with skin conditions and diseases. The design for the electrode prototype

in this experiment and technique via impedance measurements is shown in Figure 5.8. It was found that the recorded hydration levels showed a decrease as the skin impedance increased, with a negative correlation coefficient of R = −0.86, which supported the recognised hypothesis[74].

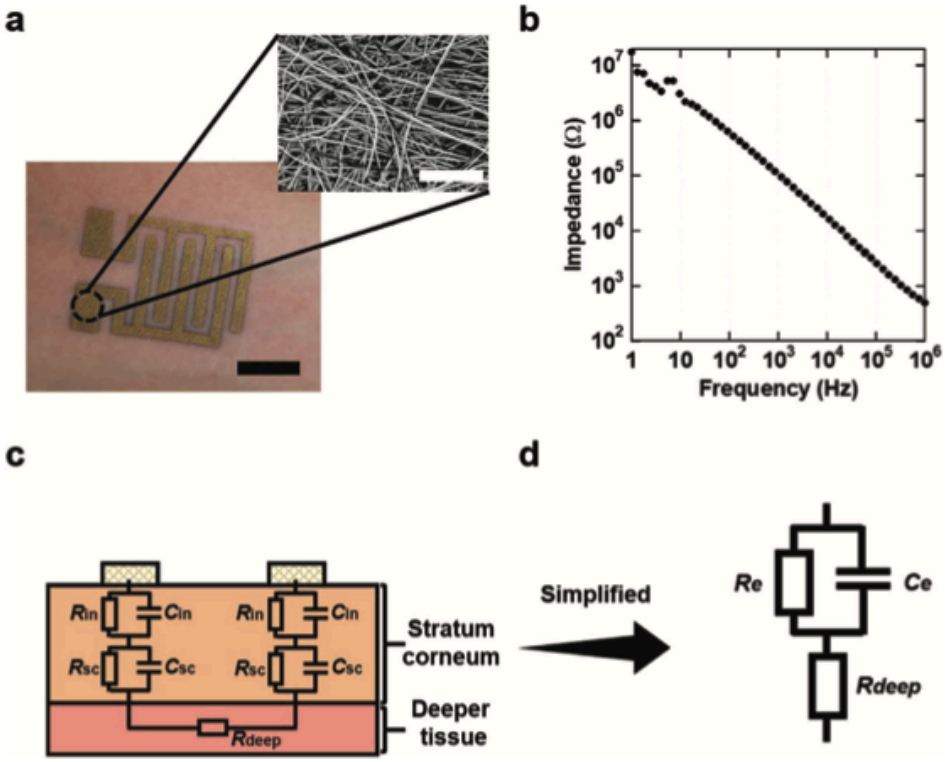


Figure 5.8: Measurement of skin impedance using nanomesh electrodes. a) Optical photograph and SEM image of nanomesh electrode pair attached to the skin, b) measured skin impedance, c) equivalent circuit of human skin, and d) simplified equivalent circuit[74]

A study with a different aim of developing a glucose monitoring system used a low-cost bioimpedance sensor to measure skin hydration. This device was tested on salt-water mixtures, a gelatine-based phantom, and human participants. The sensor detected minute changes and correlated impedance with skin hydration. The circuit built for skin impedance measurements and the experimental setup are illustrated below. Sensitivity of the sensor was tested by measuring impedance of differing concentrations of salt-water solutions. The average impedance changes over a frequency range of 30-50 kHz followed an exponential dependence with salt concentration. Gelatine phantom measurements showed an increase in impedance as water content decreased over time,

with less than 3% difference compared to human skin, suggesting suitability for testing correlations between impedance and water content[75].

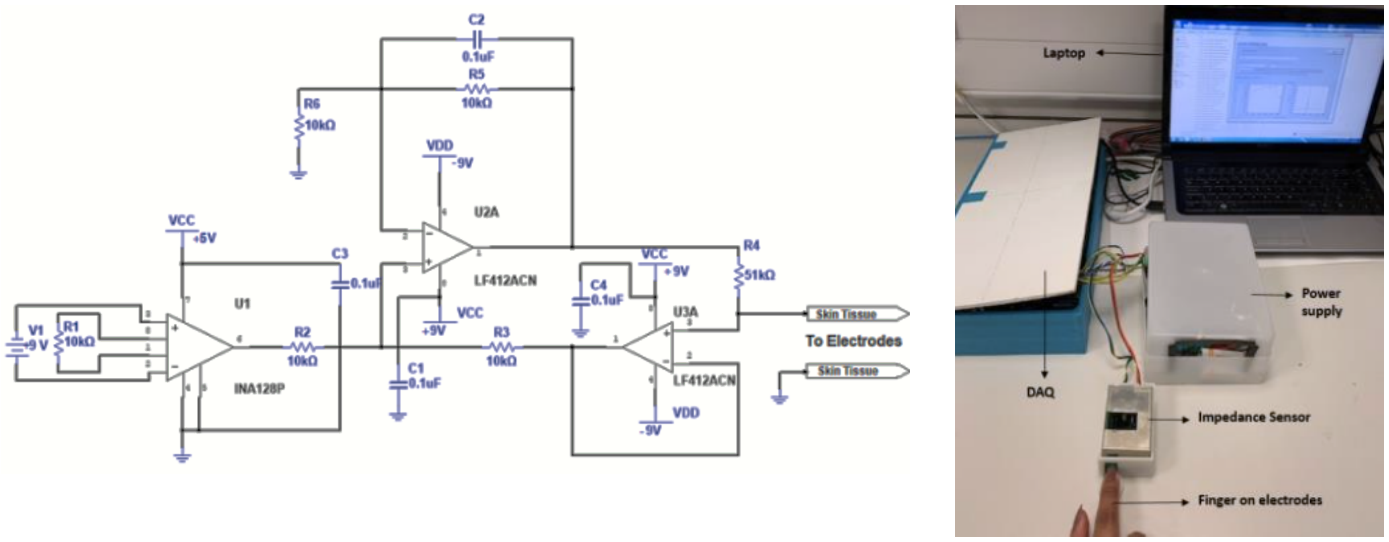


Figure 5.9: left) Schematic of the impedance sensor system. right) Laboratory setup for the sensing system[75]

Ameri et al. designed a sensor similar to a tattoo with high stretchability and consistent conductivity. It consisted of graphene electrodes tattooed directly to the skin surface, remaining functional for multiple days. The electrical impedance method measured humidity between two electrodes, showing highly consistent results with silver-silver chloride gel electrodes[76].

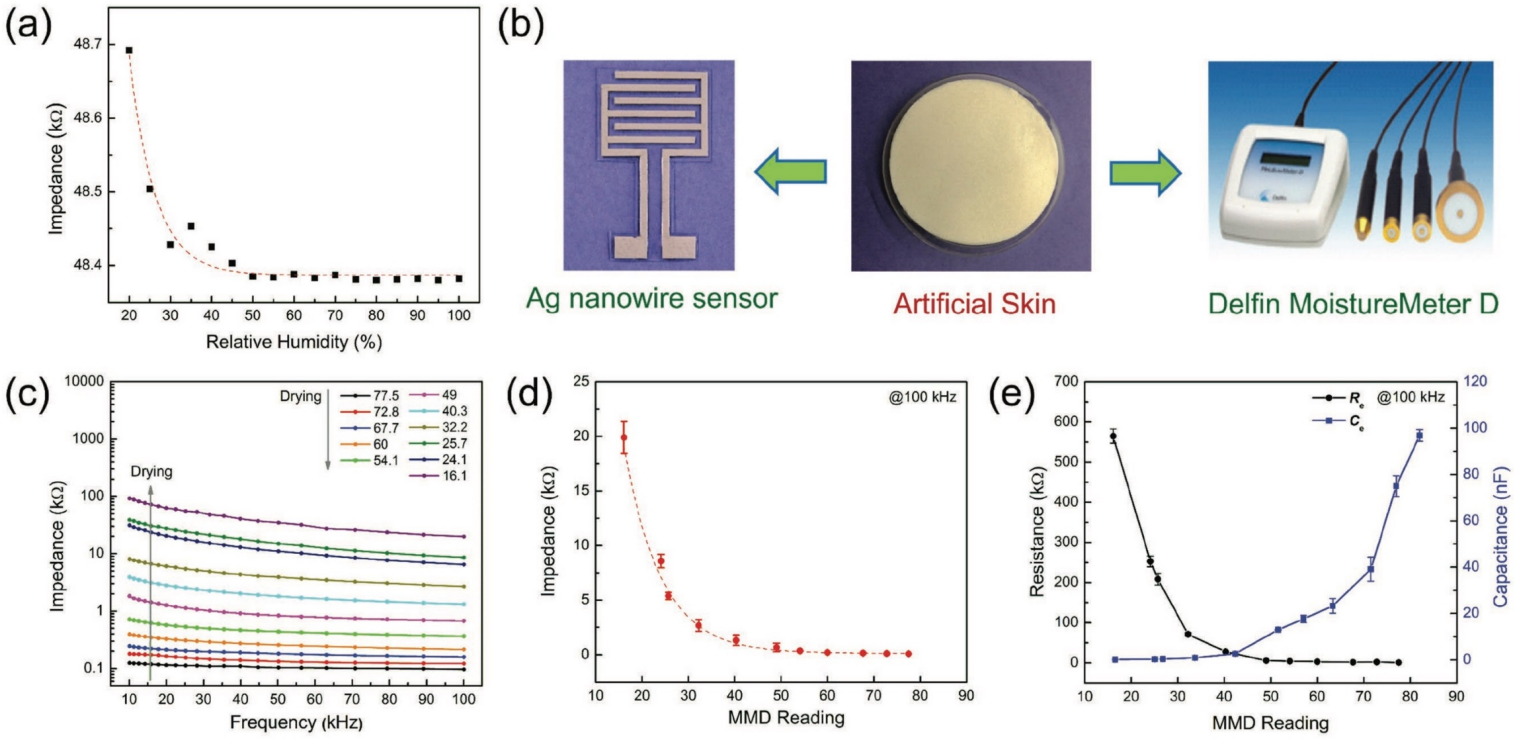


Figure 5.10: a) Impedance at 10 kHz with increasing humidity. b) Calibration with sensor and MMD. c) Impedance from sensor at 10-100 kHz against MMD as skin dries. d) Comparison of skin impedance from sensor and MMD. e) Equivalent circuit model parameters (Re, Ce).[76]

Similarly, Shanshan Yao et al. developed a skin hydration sensor that consisted of conformal silver nanowire electrodes. The sensor used the impedance method, with the capacitor comprising parallel electrodes inside a matrix. The sensor was packaged into a flexible casing that could be placed on the wrist and controlled by a microprocessor with a Bluetooth connection. This design enabled a wireless, low-cost, wearable device that could continuously measure skin hydration levels. The impedance was measured on artificial skin to establish the effect of external humidity on the skin impedance. The measurements confirmed that the sensor gave stable readings despite changes in the external surroundings. In further tests, the dielectric constant of the artificial skin when fully hydrated was measured. The impedance measurements exhibited a 0.62% increase as the water content decreased, with an exponential relationship when compared to the MoistureMeterD® (MMD) device. Furthermore, in vivo experiments involved measurements before and after the application of lotion. As expected, a decrease in skin impedance was noted following the application of the lotion, thus indicating an increase in skin hydration[77].

Rodin et al. (2022) evaluated the practical use of a wearable hydration sensor in a real-world setting. The sensor used bioimpedance to estimate total body hydration and was tested across multiple participants during daily activities. Their results focused on usability, reproducibility, and correlation with reference hydration assessments. While their findings showed promise for continuous, non-invasive hydration tracking using impedance-based methods, the study did not include a spectral or depth-resolved optical approach, nor did it explore skin-layer specificity or calibration to individual physiological differences. While this work illustrates the feasibility of real-world wearable hydration monitoring, it reinforces the need for more physiologically targeted, optically driven methods—such as those developed in this thesis—for improved skin-specific hydration assessment.

Bioimpedance has been shown to be appropriate for the measurement of skin hydration due to its inverse relationship with the electroconductivity of the skin; thus, it has a negative correlation coefficient with skin impedance. However, techniques using bioimpedance have been shown to be unable to offer direct correlations with skin water content as the current within the skin can be influenced by changes in ion movements. Therefore, the idea of multi-modal techniques in combination with standard electrical methods should be explored as well.

5.1.4 Comparison of Electrical Techniques for Measuring Skin Hydration

The Corneometer® is a well-known device based on the principles of capacitance used to measure skin hydration levels. Its probe uses digital technology, allowing for increased stability and a lesser likelihood of interferences. On the other hand, the Skicon® is a device based on the conductance method, where the conductance of a high frequency current is measured at 3.5 MHz. A visual comparison of the devices displays that the probe of the Corneometer® has no galvanic contact of the electrodes to the skin, unlike the Skicon®[78].

Clarys et al. conducted experiments to draw comparisons between an analogue and digital version of the Corneometer®. Experiments were performed on cellulose filters and calibration filter pads immersed in various water-containing solutions. A significantly high correlation was found between the amount of water in the filter and the capacitance and conductance readings, with r being 0.89 and 0.99, respectively. A high correlation was also established between the dielectric constant and the electrical readings. An in vivo experiment was also carried out on different skin sites varying in hydration levels. Highly significant correlations were established between the devices, with an r of 0.98 to the analogue Corneometer® and 0.97 to the digital version. Furthermore, an inverse relationship was shown between the hydration values and the capacitance and conductance recordings. This suggests a lower sensitivity of the Corneometer® at high hydration levels[78].

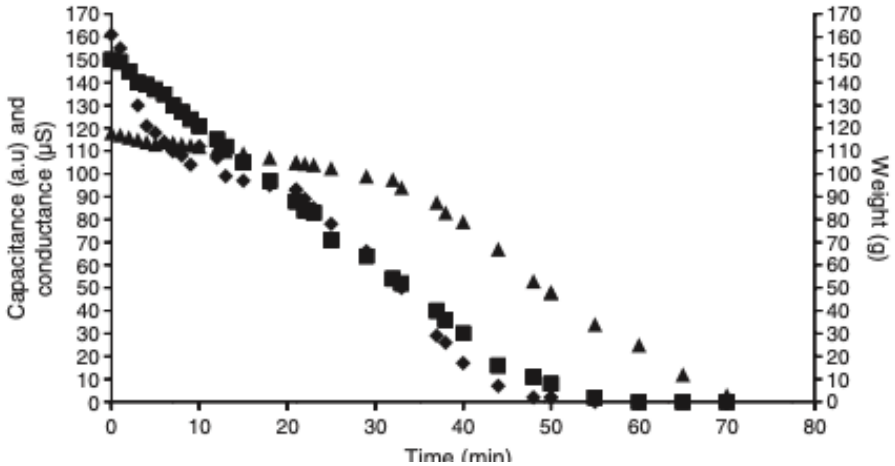


Figure 5.11: Water desorption curves on cellulose filter disk after 150 uL water. (Δ) capacitance (a.u.); (♦) conductance (uS); ( ) weight of absorbed water (mg)[78]

Moreover, Logger et al. investigated a conductance-measuring device known as the Epsilon™, similar to the Corneometer® but consisting of multiple sensors. This device was utilised in a comparative study against the Corneometer®. As the Epsilon™ device has 76,800 sensors at a single probe, the advantage of multiple readings occurring simultaneously is exhibited. The hydration at the skin surface of five different body sites was recorded to explore the variation in the anatomical site on water content. The Corneometer® displayed increased water content readings in comparison to the Epsilon™. However, both devices expressed similar results in that hydration levels increased at deeper layers of the skin. There was also a variation in surface water content at different anatomical regions, with the cheek and forearm displaying the highest water content levels and the calf showing the lowest. This validated that the skin water content measured in more hydrated skin layers will result in higher output values, as they comprise a thinner stratum corneum[65].

As previously stated, Andre et al. developed a skin hydration device known as the Moisture Evaluator. This device was compared to a Corneometer® to determine their correlations, which was found to be highly significant (0.2 N: $R^2 = 0.78$, $p < 0.01$; 2 N: $R^2 = 0.83$, $p < 0.01$) along with the determination coefficients; however, they displayed opposing sensitivity measures. There was found to be a decreased sensitivity with the developed Moisture Evaluator device for readings signifying dry skin. This is highly comparable to the suggestion that devices that utilise the conductance method have poor sensitivity for low levels of hydration. Additionally, the Moisture Evaluator demonstrated a higher sensitivity for higher levels of hydration than that of the Corneometer®, with responses still being exhibited at the point of saturation in the Corneometer® output[70].

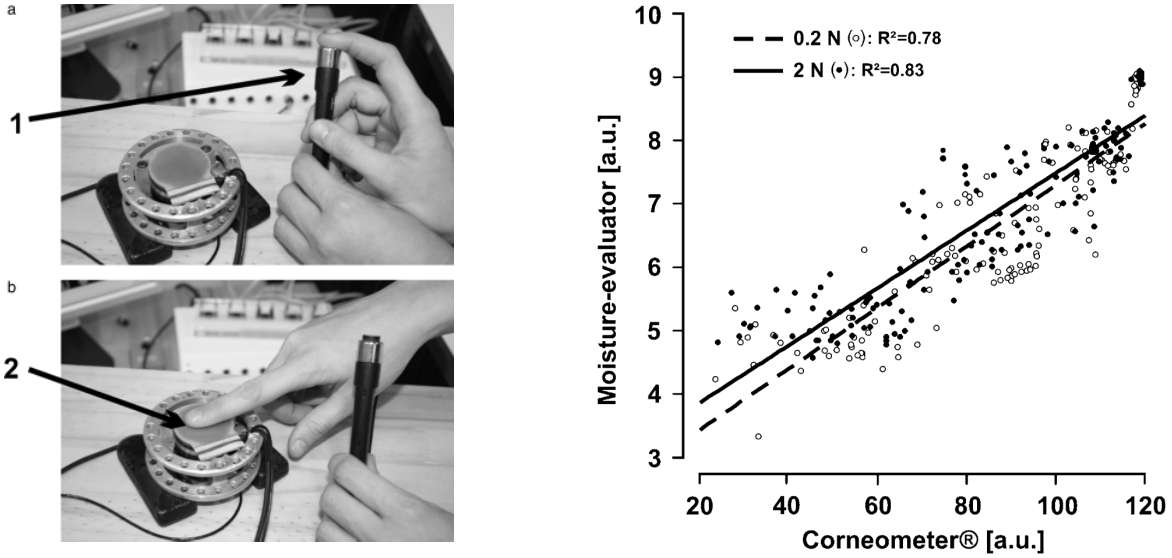


Figure 5.12: left) (a) Measurements of fingertip moisture with the Corneometer (1). (b) Measurement of fingertip with the Moisture Evaluator (2) attached to the ATI force transducer. right) Relationship between Corneometer and Moisture Evaluator (ME) measurements. Two levels of normal force exerted on the ME are presented, 0.2 and 2 N [70]

SkinUp® is a device developed by Westermann et al., which uses the impedance method to measure skin moisture and oil levels. A study was conducted to compare SkinUp® to the Corneometer CM825® to test the validity of the device. As moisturiser efficacy can be determined based on the skin's electrical properties, this was also tested using the SkinUp®. The process in which this device utilises the impedance method followed the concept that hydrated, fat-free tissue has less electrical resistance or impedance, enabling the path for electric current.

During the in vivo experiments, measurements using both devices were acquired on the forearm, cheek and forehead of the participants and these results were correlated via the calculation of Pearson's correlation coefficient. Measurements were also taken after the application of two different formulations, being Lanette wax and distilled water with alcohol on a cotton pad. The results illustrated that highly significant correlations were found between both devices on all skin locations prior to any treatment application. The relative hydration level after the cotton pad application was 77% for the SkinUp® device and 92.8% for the Corneometer®. After wax application, relative hydration was 54.4% with SkinUp® and 25.3% with the Corneometer®. This expressed that the Corneometer® was less sensitive to moisturizing formulations yet was seen to be higher with the application of water[79]

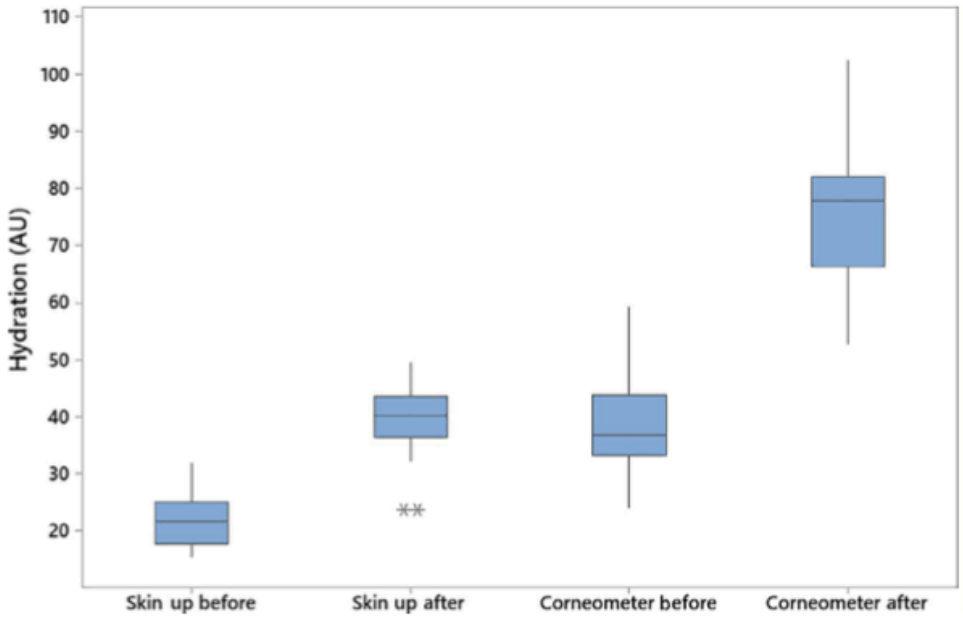


Figure 5.13: Boxplot hydration forearm graphic before and after the application of wet cotton pad measurement using SkinUp and Corneometer[79]

5.2 Optical Techniques for Measuring Skin Hydration

Optical techniques have become an essential tool in medical imaging, particularly for non-invasive assessment and analysis at the cellular and molecular levels. These techniques exploit the interaction of light with biological tissues to provide detailed images and measurements that are critical in both clinical diagnostics and research. Skin hydration measurement is one such application where optical imaging plays a crucial role, especially given the skin's importance as a barrier and its involvement in various physiological and pathological processes.

Optical imaging techniques cover a broad spectrum of methodologies, ranging from those that operate within the ultraviolet (UV) region to those in the near-infrared (NIR) spectrum. Each of these techniques offers unique advantages depending on the depth of penetration required and the specific tissue properties being measured. Optical microscopy is one of the more traditional methods, providing high-resolution images of skin structures at the microscopic level. However, its use in measuring skin hydration is limited due to its shallow penetration depth and the need for biopsied samples.

More advanced techniques, such as Doppler imaging and optical coherence tomography (OCT), allow for deeper tissue analysis. Doppler imaging, which measures the frequency shift of light as it reflects off moving particles (such as blood cells), can provide insights into skin perfusion and microcirculation, indirectly related to hydration. OCT, on the other hand, uses low-coherence interferometry to create cross-sectional images of the skin, providing detailed information on skin structure and thickness. OCT's ability to penetrate a few millimetres into the skin makes it particularly useful for assessing hydration in different skin layers, including the epidermis and upper dermis.

Diffuse optical imaging encompasses techniques that use scattered light to probe deeper tissue layers. This category includes both fluorescence imaging and near-infrared spectroscopy (NIRS), which are particularly relevant for skin hydration measurement. Fluorescence imaging, which relies on the emission and absorption of light photons by fluorescent probes, allows for the visualization of specific molecules within the skin. These probes can be designed to target specific components of the skin's extracellular matrix or water content, providing a detailed map of skin hydration at a molecular level. Advances in fluorescent probes, particularly in the NIR range, have increased the depth

of penetration and the specificity of these measurements, allowing for more accurate assessments in clinical and cosmetic dermatology.

NIR spectroscopy (NIRS) is especially significant in the measurement of skin hydration due to its ability to penetrate several millimetres into the tissue and its sensitivity to water content. By analysing the absorption spectra at specific NIR wavelengths, NIRS can provide quantitative measurements of water content in the skin, making it an invaluable tool for both clinical diagnostics and skincare product development. The deeper penetration of NIR light compared to visible light allows for the assessment of hydration levels in the dermis, which is critical for understanding the overall water content of the skin and its impact on skin health.

One of the key advantages of these optical techniques is their non-invasive nature, which allows for continuous monitoring of skin hydration without the need for tissue samples or invasive procedures. This is particularly important in studies involving vulnerable populations, such as infants or the elderly, where maintaining skin integrity is crucial. Furthermore, the real-time feedback provided by these techniques enables dynamic studies of skin hydration, such as the effects of topical treatments or environmental changes.

In summary, this section will cover various optical techniques used for measuring skin hydration, focusing on attenuated total reflectance infrared spectroscopy (ATRIS), nuclear magnetic resonance (NMR), Raman spectroscopy, and near-infrared spectroscopy (NIRS). Each of these methods has its own unique methodology and application, providing a comprehensive toolkit for assessing skin hydration in both research and clinical settings. The selection of the appropriate technique depends on the specific requirements of the study, including the depth of penetration, resolution, and the type of information needed about the skin's water content. As these technologies continue to evolve, their application in skin hydration measurement is expected to expand, offering even greater accuracy and insight into the hydration dynamics of the skin.

5.2.1 Use of Attenuated Total Reflectance Infrared Spectroscopy (ATRIS)

Attenuated Total Reflectance Infrared Spectroscopy (ATRIS) is a widely used technique in infrared spectroscopy, particularly valued for its ability to analyse the structural and compositional properties of samples with minimal preparation. ATRIS operates on the principle of internal reflection, where infrared light is directed onto a sample at a specific angle, causing the light to reflect multiple times within an internal crystal. This reflection generates an evanescent wave that penetrates a few micrometres into the sample's surface, allowing for precise interaction with molecules, particularly in thin films, liquids, and powders[80].

One of the key advantages of ATRIS is its effectiveness in measuring the water content in the stratum corneum, the outermost layer of the skin. By analysing specific absorbance peaks, such as those at 8.94 µm and 9.65 µm, or the water absorbance peak at 4.76 µm, ATRIS can detect small changes in water content, which are crucial for assessing skin hydration levels. This capability makes ATRIS particularly useful in dermatology and skincare research, where understanding hydration is essential for evaluating the efficacy of topical treatments.

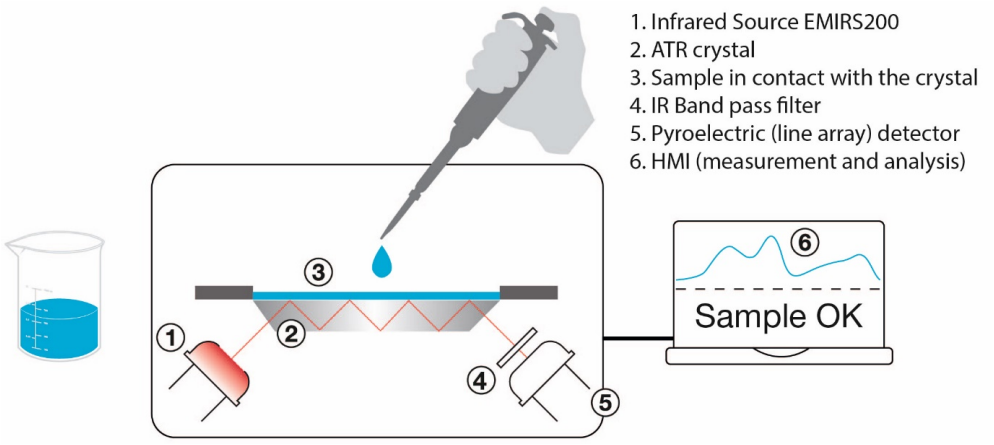


Figure 5.14: Attenuated Total Reflectance Infrared Spectroscopy Measurement Process[80]

ATRIS is often combined with Fourier-transform infrared spectroscopy (FTIR), which enhances the sensitivity and resolution of the spectra obtained. FTIR converts raw data into detailed molecular spectra, making it especially useful for studying complex biological samples like skin. Historical studies by researchers such as Potts and Bommannan have demonstrated the utility of these techniques in understanding the

relationship between skin hydration and barrier function, highlighting the critical role of water in maintaining skin integrity[80].

Beyond skin hydration, ATRIS has broad applications in biomedical research, material science, and other fields where preserving sample integrity is crucial. Its non-destructive nature and ability to provide detailed compositional information make it an essential tool in both research and clinical settings.

5.2.2 Use of Nuclear Magnetic Resonance (NMR)

Nuclear magnetic resonance (NMR) is an imaging method that measures the interaction of nuclear spins when an applied magnetic field is present. The nuclei within the molecules of the sample under investigation begin to precess when in the magnetic field. The sample is subsequently exposed to radio waves about the same frequency as the precession, allowing for the NMR spectra to be acquired.

An example of a widely used NMR scanner is the BioSpecs 70/20 7T MRI imager by Bruker. When combined with a specific probe and image acquisition sequence, the device produces images of forearm skin with an axial resolution of 86 μm. Franconi et al used this scanner in a study that measured the relaxation time (T_2) at the epidermal level, which showed an increase after the use of a hydrating cream. As this function is wedded to the content of free water, its increase signifies betterment in epidermal hydration.

Figure 5.15: Bruker's BioSpecs 70/20 7T MRI imager

Philippe Girard et al conducted a study with an aim to determine the capability and the analytical quality of 3 different in vivo, non-invasive, quantitative methods for measuring skin hydration. NMR-S is one of the only direct hydration measurement methods and it measures skin hydration down to the outer dermis with high precision. NMR-S, TTT and corneometry represent 3 possible ways to assess skin hydration. Because they explore different cutaneous depths, they are more complementary than competitive. The work presented evaluated the capability and precision, as well as cutaneous exploration depths, of the 3 methods via in vivo experiments.

5.2.3 Use of Raman Spectroscopy

Raman spectroscopy is a technique based on the principle of light interaction with chemical bonds of a sample. The technique is described by the scattering of light outputted from a laser. Rayleigh Scatter depicts the scattered light that has the same wavelength as the incident light, whereas the Raman Scatter is the valuable information from light scattered at different wavelengths to the source. A Raman spectrum provides chemical information that allows for a material to be easily distinguished. Figure 5.16 below illustrates a typical Raman spectrum of water and two different concentrations of ethanol. Raman Spectroscopy can be combined with other methodologies, an example being mapping where images can be produced to show the distribution of the chemical components within a sample[81].

The full process of Raman spectroscopy involves the application of incident light causing an inelastic scattering process. Energy is then transferred to and from the medium meaning that the scattered photon will either have more or less energy than the incident light photon. If the energy is less it is referred to as Raman stoke scattering, and if the energy is more it is referred to as Raman anti-stokes scattering. The output of the Raman spectrometer expresses a spectrum of scattered intensity against the frequency shift between the incident and scattered photons. The final spectrum comprises of a series of Raman bands consistent with various vibrational modes of the sample molecule.

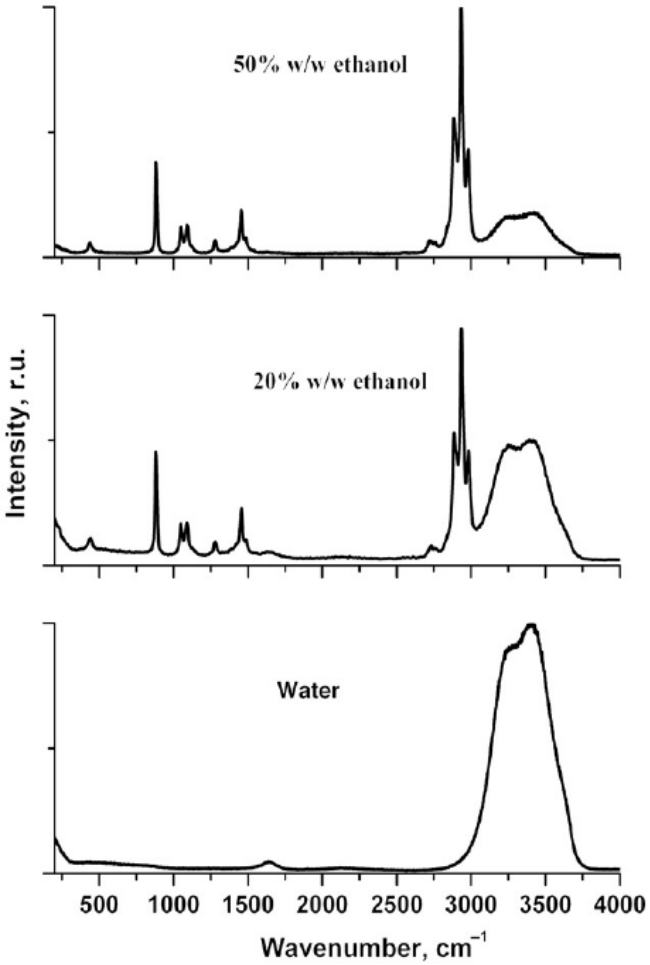


Figure 5.16: Typical Raman spectrum of water at different concentrations of ethanol[82]

Tippavajhala et al conducted in vitro and ex vivo experiments to investigate the efficacy of moisturizers on skin hydration. Confocal Raman spectroscopy was used to analyse the water content and natural moisturizing factor of the stratum corneum layer of skin. This was carried out using the RiverDiagnostics 3510 Skin Analyzer (RiverDiagnostics, Rotterdam, Netherlands) which has a spectral range of 400-4000 cm^{-1}. Values for water content were calculated from the Raman spectra and plotted against the depth of skin to give the water content profiles. Figure 5.17 below illustrates the confocal Raman spectra with its distinct water and keratin peaks. The results of this human study confirmed the effectiveness of using confocal Raman spectroscopy as an analysis technique for differing skin hydration levels. It also suggested that the extent of skin hydration depends largely on the duration of application in terms of moisturization.

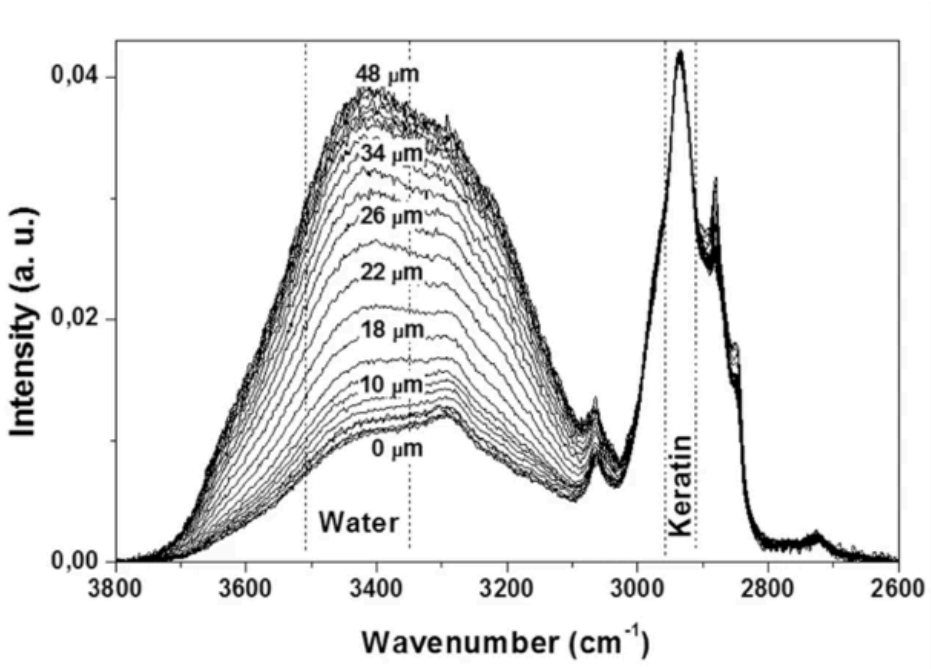


Figure 5.17: Use of confocal Raman spectroscopy of human skin of differing hydration levels showing the water and keratin peaks[83]

Wang et al also used confocal Raman microscopy (CRM) to measure hydration in the stratum corneum. Deuterium oxide (D_2O) was used in this study as a probe via topical application causing the diffusion of D_2O in order to quantify its kinetics. Results of the in vivo experiments established that CRM can be used to acquire distinctive information on the hydration mechanisms of the stratum corneum and the epidermis layers of the skin[82].

Ruini et al conducted an in vivo examination into the effect of moisturizers on human skin measured using both confocal Raman spectroscopy (CRS) and optical coherence tomography (OCT). The OCT device that was used was VivoSight, which functions at a centre wavelength of 1305nm and can achieve a penetration depth of 1.5-2mm, producing two-dimensional cross-sectional images. Subsequent to analysis, the optical attenuation coefficient (AC) was found to be lower on more hydrated skin and would appear darker on the OCT images in comparison to dry skin. CRS was performed on the gen2 Skin Composition Analyzer, which uses vibrational spectroscopy constructed on the principle of the inelastic scattering of light photons. This method gives the capability for the analysis of the molecular compositions, thus can provide quantitative concentration profiles of compounds in the stratum corneum, such as skin water content. The in vivo experiment involved 20 subjects with healthy skin, involving measurements

taken before and after a 2 weeks application of moisturizer on one forearm, with the other forearm acting as a control measurement. Following moisturization, the results presented an increase in epidermal thickness in comparison to a decrease with the moisturized control. A decrease in skin roughness and a positive correlation with the level of water present was also seen with increased epidermal thickness, however a strong statistical significance was not conveyed ($p = 0.41$). These results support the hypothesis that moisturization primarily affects deeper epidermal layers more so than the SC layer. To conclude, it is established that short-term application of moisturizers can be insufficient to achieve significant alterations in skin composition and morphology but achieves an impermanent swelling of the lower epidermis. This also suggests the need for multimodal methodologies to accurately assess hydration levels of the SC[83].

5.2.4 Other Recent Optical Techniques

There are other different optical techniques that are being explored more recently and can be used to measure hydration in the skin. Recent optical techniques include optical tissue probing, optoacoustic methods, and opto-thermal methods, combining optical methods for detailed skin constituent information. These techniques combine optical methods into a multi-modal approach to extract a greater detail of information of the skin constituents.

The use of optical tissue probing with speckle patterns analysis can also be used as a technique to detect human skin hydration. This involves the temporal tracking of back-reflected speckle patterns while applying illumination and episodic vibrations. An optical system using this method was developed by Y T Kelman et al and tested against the Corneometer® (Courage & Khazaka, Cologne, Germany). Speckle patterns were obtained from an area of skin near to the centre of illumination to permit information from a smaller penetration depth, since it is associated with less-scattered light. The developed device is seen in Figure 5.18[84].

The speckle contrast imaging (SCI) technique highlighted here measures skin hydration by detecting changes in the speckle pattern created by coherent light scattering from the tissue surface. However, speckle-based systems are also sensitive to dynamic scatterers, such as red blood cells, making them inherently cross-sensitive to blood flow.

Fluctuations in blood perfusion can influence speckle contrast values, potentially confounding hydration measurements, particularly in areas with variable microvascular activity. This cross-sensitivity poses a limitation when attempting to isolate hydration-related structural changes from haemodynamic effects. Therefore, any speckle-based hydration assessment must consider and, where possible, control for underlying blood flow variability, especially in regions of high vascularisation or during periods of physiological change[84].

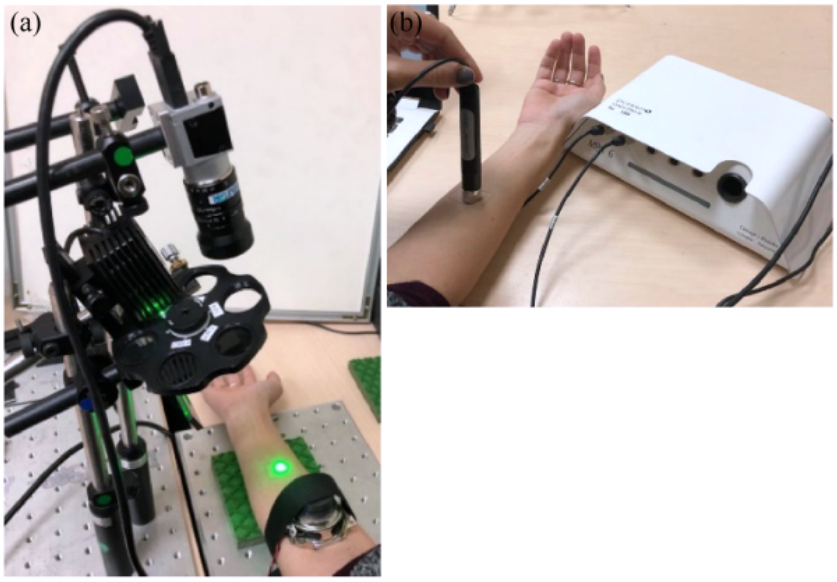


Figure 5.18: The two experimental instruments: (a) Optical setup, extracting the optical parameter. (b) The Corneometer CM825® probe measuring capacitance[84]

The in vivo experiment involved a range of dry to moist skin through the application of moisturizer on the volar arm, with measurements taken every 30 minutes across 3 hours. Results showed that after the application of moisturizer, the optical signature displayed a significant decrease over time from the dry skin signature. Furthermore, it was also found that the higher the level of moisture, the faster the acoustic wave faded. With the calculation of the Pearson correlation coefficient, a negative correlation was identified between the trends of both instruments of R = -0.87. Unlike with the Corneometer, the developed optical system illustrated a large range of hydration readings and a high sensitivity for the identification of high hydration levels[84].

S A Perkov conducted studies on both gelatine tissue phantoms and human skin to investigate the monitoring of water content using optoacoustic (OA) methods. This particular technique is suggested to have high resolution and contrast, due to the combination of ultrasound and optical methodologies, as well as a significant penetration depth for its probes, thus making it extremely appropriate for the monitoring of skin water content. The typical experimental setup for these measurements includes lasers emitted from a fibre bundle and an optical transducer connected to an oscilloscope, as seen in Figure 5.19[85].

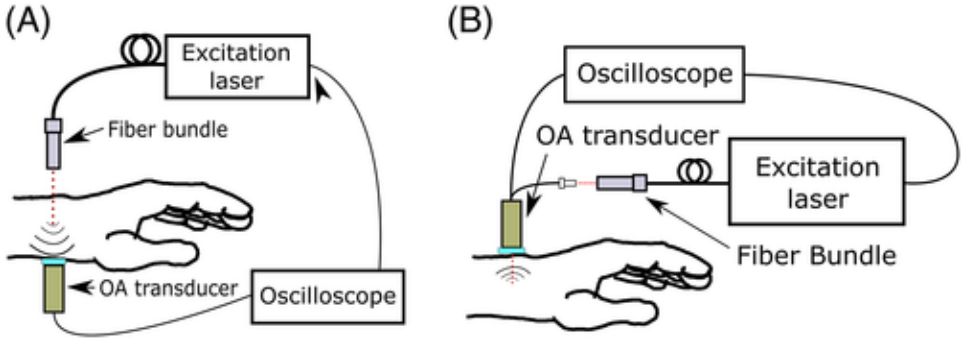


Figure 5.19: Experimental setup for OA measurements in the human wrist in the transmission mode A; and the reflection mode B[85]

Through analysis of the acquired OA signals, an evident second peak was discovered at a depth of 2mm, which signified the signals originate from subcutaneous tissue, since this tissue is found to have a higher water content than more superficial layers of skin. The conducted in vivo experiments established the capability of the optoacoustic technique to effectively detect signals generated by water absorption in different skin layers[86].

In addition, Imhof et al has conducted multiple in vivo experiments using opto-thermal methods to measure the hydration of the SC. A more novel technique they have used is the combination of opto-thermal radiometry and condenser-chamber TEWL. This allows for measurements of both SC surface hydration and evaporimetry, thus providing information on how the water diffusion coefficient depends on SC hydration[87], [88].

5.3 Use of Near Infrared Spectroscopy

5.3.1 Near Infrared Spectroscopy and Properties of the Absorption Spectra of Water in Skin

Near-infrared spectroscopy (NIRS) is a type of vibrational spectroscopy that covers the 780-2500nm wavelength region of the electromagnetic spectrum and provides information on the perfusion of tissues. This wavelength range is high enough to promote the excitation of a molecule, but only to lower excited vibrational states. The information on tissue perfusion is achieved through the measurement of light absorbance to monitor tissue oxygenation. This technique was initially used solely for the assessment of oxygen saturation in the brain but is more widely used for the analysis of other bodily tissues. NIRS is measured using a spectrophotometer, which typically consists of a light source, a monochromator, a sample holder and a detector. These devices have the ability to record the transmittance, absorption and reflectance of samples. Many contain a double beam, meaning that the radiation provided from the source is divided into 2 beams before reaching the sample, one to the reference and the other through the sample.

An advantage of imaging within the NIR region is the high penetration depth into a sample in comparison to imaging in the mid infrared range. This depth can allow for the absorptivity based on the molecular overtone and combination vibrations to be obtained. Although the sensitivity of this technique can be depicted as relatively high, there is the disadvantage of overlapping of broad absorption bands that would consequently require mathematical data analysis. NIRS is based upon combination and overtone bands due to molecular vibrations, and as such, the bands visualized in the NIR region are characteristically broad and yield complex absorption spectra. To overcome this, multivariate calibration methods are required to extract relevant information for subsequent analysis[89].

Water molecules are known to vibrate dependant on its state (being solid, liquid or gaseous state), which directly relates to the absorption of the molecule by electromagnetic (EM) waves. The vibrations seen in gaseous water are made up of symmetric and asymmetric stretches and covalent bond bending. On the other hand, liquid water consists of primarily hydrogen bonds. Vibrational transitions are established to be responsible for absorption within the near-infrared (NIR) region of the EM

spectrum. Figure 5.20 below illustrates the absorption spectra of water in its different states.

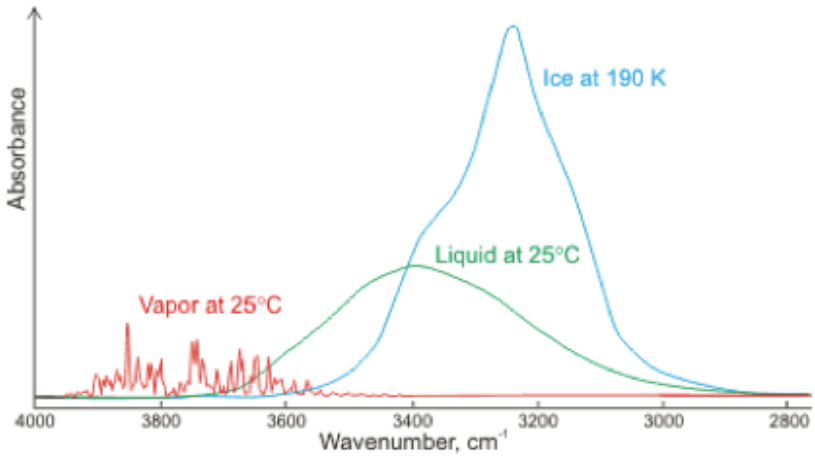


Figure 5.20: Absorption spectra of water in its different states[157]

The absorption bands for liquid water are located around 970nm, 1200nm, 1450nm and 1950nm, which all fall within the NIR region, therefore, in order to measure the absorption spectra of liquid water, near-infrared spectroscopy (NIRS) is required with the use of a spectrophotometer. It has been found that beyond the NIR region towards the infrared (IR) region, liquid water changes in its behaviour. This change is exhibited by large constant peaks presenting itself at wavelengths higher than 2500nm. Conversely, below the NIR region, no clear peaks are noticed due to high transmission through the near-ultraviolet region (up to 400nm).

The typical NIR spectra of human skin consists of absorption bands that relate are sensitive to specific chemical groups, such as OH, which are highly present in water. The intensities within the absorption spectra have been found to be directly proportional to skin water content. Thus, the given NIR spectra has the capability of differentiating various types of water in the stratum corneum layer of the skin. The absorption spectra of water in the near infrared region consists of significant peaks, with two prominent peaks at 1920nm and 1450nm. The peak present at 1920nm is the combination band from the OH stretch and HOH bands, whilst the peak at 1450nm is the overtone band of the OH stretching. There are also additional weaker bands, for example around 1700nm, due to alkyl CH groups present in lipids and proteins within the skin. NIRS has the ability to directly detect water content in the skin through analysis of the intensities of these combination and overtone bands within the NIR region[54].

5.3.2 Use of NIRS in Experimental Studies

In skin hydration studies, in vitro and in vivo hydration investigations are typically carried out on porcine skin and human participants respectively. M Qassem conducted a study which investigated the properties of the stratum corneum in the NIR region of the EM spectrum and visualised the water characteristics in this layer using a spectrophotometer. The results for the in vitro experiments displayed 2 large peaks in both the water and porcine skin spectra at approximately 1450nm and 1920nm, with that of the porcine skin having an upward shift. This is due to porcine skin having a higher absorption coefficient for light than water. The existence of these peaks in the porcine skin spectra confirms the presence of water in the samples. The results for the in vivo experiments showed peaks at 1450nm and 1920nm as expected, with the intensity at a lower magnitude than that of water. Beyond 1900nm, the spectra exhibited high background interference due to the increase in penetration depth. This confirms that deeper tissue is more saturated with water, thus having a higher absorptivity than more superficial skin layers. It was found that even though the probe being in direct contact with the skin allowed for water bands to be more clearly identified, this condition can lead to occlusion, therefore non-contact readings are preferred. Moreover, a single fibre detector with a 6-fibre source allowed for the most reliable and evident readings to be acquired[90].

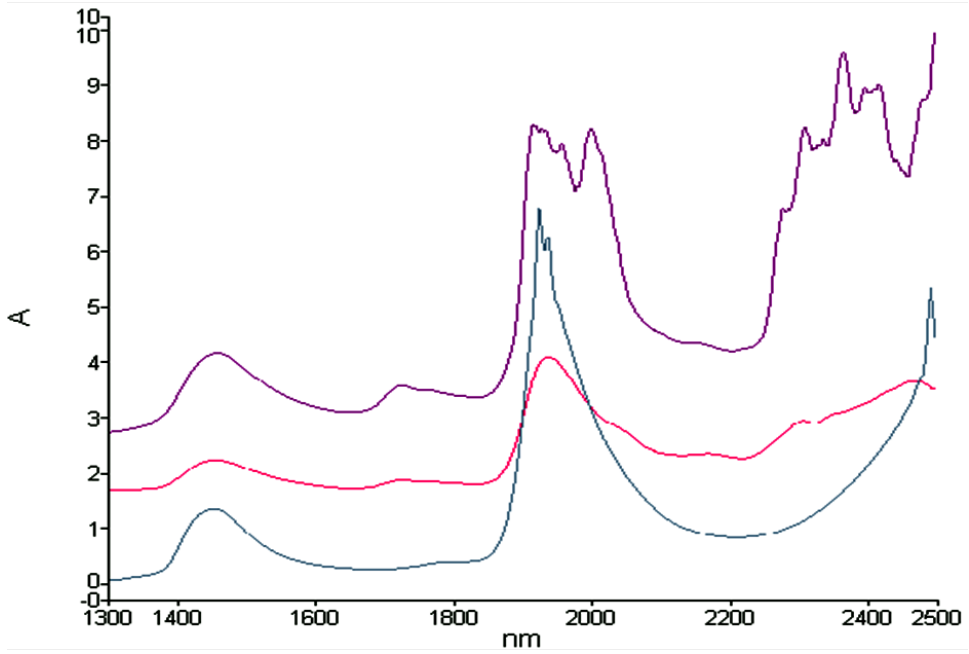


Figure 5.21: Smooth NIR absorption spectra of pure water (grey), 1mm of porcine skin (pink), and 2mm of porcine skin (purple). Smoothing factor = 25[90]

A study with a similar methodology and measurement principle conducted by Kilpatrick-Liverman et al determined differences in skin water content by measuring the absorption spectra also using an NIR spectrophotometer with an optical fibre probe. The influence of relative humidity (RH) on skin water content was assessed via an intro experiment involving porcine skin equilibrated in desiccators containing saturated salt solutions at different concentrations. Gravimetric readings were then taken to measure the weight loss due to dehydration, which is utilized to calculate the percentage of water uptake. It was found that as RH was decreased, lower water content was recording in the skin. This result was established due to the area under the 1936nm band being highest for skin that was equilibrated at 100% RH and lowest at 11% RH. Additionally, the effect of relative humidity (RH) and moisturizers on skin water content was evaluated by product application to participants in various in vivo experiments. In vivo experiments included studying the clinical effect of %RH changes via washing skin with exclusively water, the effect of humectants and the effect of a choline spray on skin water content. Readings were taken using the Skicon and NIR measurements and the obtained data analysed with paired t-tests comparing changes from baseline readings. A direct correlation of R = 0.83 was observed between the %RH and NIR readings. It was concluded from the results of the in vivo investigations that products containing substances such as wax and oils smooth the surface of skin, which increases beam penetration depth. Although these moisturizing products are implied to increase skin water content, they primarily increase the sampling volume due to this increased beam penetration[91].

Similarly, H Arimoto and M Egawa conducted a study investigating measurements of non-contact skin moisture using NIR spectroscopy. In vitro experiments were carried out using porcine skin to explore the relationship between absorbance spectra peaks and water content. In vivo experiments collected diffuse reflected spectra using an optical fibre probe on skin. The measured spectra from these investigations were analysed by multiple linear regression (MLR) and partial-least squares (PLS) regression, with these results then compared to recordings obtained using a capacitance method. Results showed that the recorded measurement depth is highly dependent on the absorption of water, with deeper penetration present in spectral regions with weak water absorption. Moreover, this is found to be greater than the depth measured via the capacitance

method. Correlations between water content and the second derivative were found to be highly significant at 0.99 and 0.98[92].

Furthermore, a subsequent study was conducted investigating the measurement of water content distribution in the skin. This involved using confocal Raman spectroscopy to measure the vertical distribution of water in the skin, followed by an estimation of the sensitivity of water content and a near-infrared imaging experiment. Skin water content was measured immediately after the application of a wet pad and again after 5 minutes. When skin was moist, the water content was shown to decrease from the skin surface to a depth of 5-10μm. Moreover, the water content at the skin surface was lower for the immediate recording. Beyond 5-10μm, both water content recordings were almost identical. The imaging portion of the investigation involved an in vivo experiment that visualized the distribution of skin water content at 3 different wavelengths. At a wavelength of 1300nm, the pixel value was stagnant through 5 minutes after removal of the wet pad. At 1462nm, half the participants presented an increase up to 1 minute, then a subsequent plateau. At 1950nm, all participants displayed an increase up to 1 minute, then again, a subsequent plateau. These results convey that a wavelength of 1950nm has the highest sensitivity to changes in skin surface water content. Although there were absorption peaks at both 1950nm and 1462nm in the NIR spectral range, the pixel value variation at 1462nm appeared to be smaller[93].

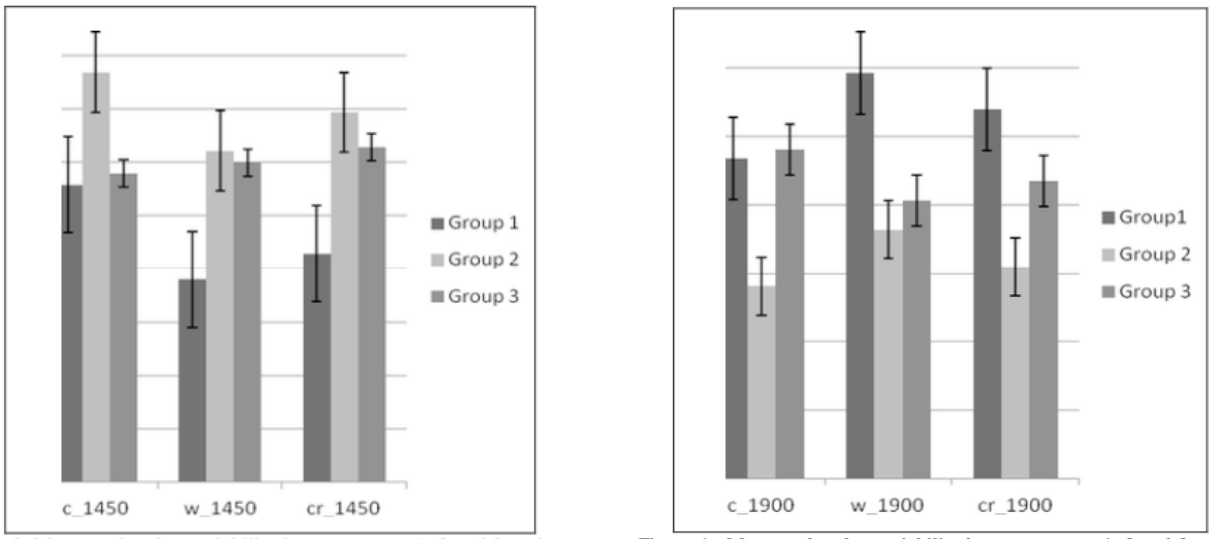


Figure 5.22: left) Mean peak value variability between group 1, 2 and 3 at the OH overtone band around 1450 nm region. right) Mean peak value variability between groups 1, 2 and 3 at the water combination band between the 1900-1920 nm region[93]

Alongside such studies, the effect of moisturization on the skin is also focused upon in the field of skin hydration measurements. A study by M Qassem assessed the

optical properties of skin following water contact and application of moisturizer using a spectrophotometer with a fiber optic probe. The resulting spectra as an average of all participants recorded prior to water or moisturizer application displayed higher peaks of bands near 1450nm and 1780nm. This response was similarly seen when comparing individuals who frequently moisturize and do not moisturize. Results for participants with dry skin were observed to be most contradictory to those with normal skin, regardless of moisturization. This suggests nonconformities in the barrier fucntion charcteristics of skin and an increase in sensitivity with dry skin[94].

The advancement into the development of sensors contained within a built device is becoming more prevalent, as well as investigations into the use of multiple wavelengths when conducting optical related experiments. For example, M Mamouei et al designed and developed a multi-wavelength optical sensor to measure dermal water content. The sensor consisted of 2 separate modules; the probe containing 4 LEDs, a photodiode and a transimpedance amplifier, and the main module. The main module encapsulates analogue to digital converters (ADCs), current sources and also a connection to a USB port for data transfer via an Arduino microprocessor. Samples of porcine skin was prepared in an environmental chamber at 96% relative humidity and 25°C for 48 hours. These conditions ensured for the skin samples to reach their maximum hydration levels. A desorption test was conducted that simultaneously measured the optical absorbance obtained from the developed sensor simultaneously to gravimetric weight measurements. The set up for this experiment including the developed optical sensor is illustrated in Figure 5.23 below and further investigations are discussed in more detail[95].

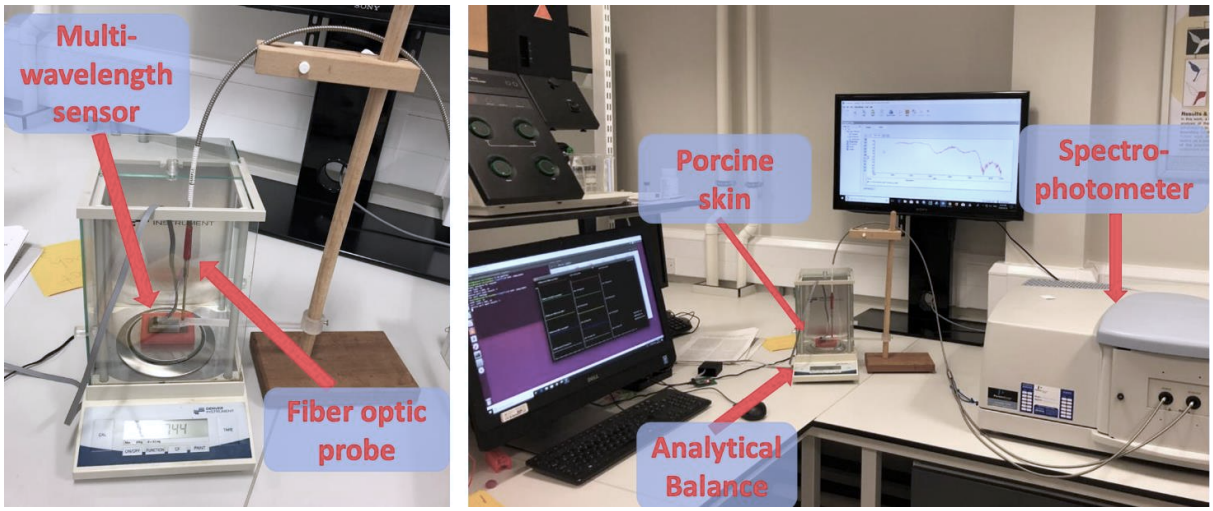


Figure 5.23: In vitro experimental setup[95]

A further study was conducted using the same developed optical hydration sensor. An in vitro experiment was performed on porcine skin to compare the developed sensor with a spectrophotometer to determine its performance and efficacy, with the reference being an electronic precision balance for gravimetric measurements. These weight readings will serve as a ground truth value for the water content in the sample, whilst the spectrophotometers reflectance spectra will be used to benchmark the optical measurements. The porcine skin was segmented and inserted into an environmental chamber at 96% relative humidity and 25°C. For conducting the measurements, both the probe from the developed sensor and the fibre optic probe from the spectrophotometer were placed and secured on to the skin surface. Both optical and weight recordings were obtained throughout a 3-hour period. There showed to be a high agreement between the absorbance results from the developed hydration sensor and the spectrophotometer, with both weight and absorption presenting a decrease as water content diminished. Furthermore, optical measurements at the 1450nm band displayed a higher sensitivity to water content variations. These comparisons indicate that the developed sensor exceeds the prediction accuracy of the spectrophotometer. Future investigations on the developed sensor would involve the addition of in vivo studies to assess its performance on human skin and examining the measurement accuracy when different skin types are considered[96].

Volkova et al. (2023) presented a SmartWatch-integrated multispectral sensor that monitors human skin hydration and sweat loss using sensor fusion techniques. Their system combined data from optical sensors operating at multiple visible and NIR wavelengths to estimate hydration status during physical activity. The study demonstrated that combining signals across a range of spectral bands improved the robustness of hydration estimation, particularly under dynamic, real-world conditions. Their results showed the potential for wearable multispectral sensing in tracking physiological fluid changes, though the emphasis was primarily on activity-driven sweat monitoring rather than stratified dermal hydration. This study supports the value of multispectral approaches in hydration sensing but lacks the depth-specific and model-driven optimisation central to the multi-wavelength NIR strategy proposed in this thesis on non-invasive dermal hydration monitoring.

Studies with developed optical sensors were found to be comparable to standard spectrophotometers with prominent water peaks of their absorption spectra typically

expressing in its correct wavelength regions. An increase in skin hydration led to a rise in the magnitude of these spectral peaks and direct correlations between water content and NIR readings. It has also been proposed that optical sensors with multi-wavelength channels within near infrared region (NIR) will predictably exceed the accuracy of the standard NIR spectrophotometer with an increased sensitivity to minor alterations in the water content within the SC, in particular at higher wavelengths towards 1950nm.

5.4 Multi-Modal Measurement Techniques for Skin Hydration

The use of multi-modal techniques for the measurement of bodily parameters are becoming more valuable for their advantages. Such include an increase in the accuracy of readings, due to the combination of modalities providing a confirmation of the results. This acceleration in the application of multi-modal techniques highlights the increasing need for scientific advancements in this area to enable more reliable measurements. The ability to detect multiple parameters simultaneously with a single device has proved challenging for researchers, therefore driving the need to address this. Although there is not yet any combined analysis of multi-state variates within the experimental side of multi- modal sensing, efforts are being made by researchers to investigate this method of testing. A downside to multi-modal sensors opens the challenge of utilising a larger hardware circuit size and possible trade-offs in performance due to the micro-integration[97].

A study that used a multi-modal sensor in its methodology was conducted by Krishnan et al. They developed a device that introduced multi-modal sensors using both thermal transport and electrical properties to allow for measurements from sensitive areas of the skin. Analysis algorithms were utilised to provide the electrical impedance, temperature, thermal diffusivity and conductivity and heat capacity. In vitro studies were conducted on porcine skin soaked in saline water followed by independent measurements from the separate techniques. Temperature distributions of the porcine skin via infrared thermography presented that the level of hydration in the skin had a direct effect on its thermal behaviours. This is primarily sourced by secondary water in the epidermis, which also has a strong impact on the electrical properties. In vivo studies involve measurements taken from the volar forearm of healthy volunteers, with different

locations of this area tested with or without the application of moisturisers with differing glycerine concentration levels. Each measurement would result in the output of impedance and thermal data, which would then be compared to the hydration readings obtained from a Corneometer device. Additionally, transepidermal water loss measurements were taken using a standard Tewameter device. Overall, the results conveyed a strong agreement between the developed multi-modal device and state-of-the-art techniques, suggesting a use for these device types clinically and for the real-time monitoring of hydration[97].

Another multi-modal approach was adopted by Cho et al. A chronic wound monitoring system was developed that could interface multiple signals, being voltage, resistance and capacitance measurements, using an integrated circuit. This gave the ability to acquire measurements of pH, temperature and humidity levels for diagnosis. In terms of the sensors used, pH was measured using a silicon substrate in combination with an electrode, an analogue voltage-type sensor was used to measure temperature and a capacitive-type sensor for measuring humidity. Conducted experiments included testing with buffer solutions ranging from pH = 4 to 10, using a heater with commercial temperature inputs, and a humidity chamber to acquire measured capacitance values. The results concluded that the developed multi-modal sensor system was able to deliver accurate and reliable diagnoses of chronic wounds at a lower cost and energy consumption, whilst still upholding a higher accuracy[98].

5.5 Summary of Skin Hydration Measurement Techniques

This sub-chapter has provided a comprehensive review and comparison of various techniques used to measure skin hydration, focusing on electrical methods (such as capacitance and conductance), bioimpedance, and optical methods (including ATRIS, NMR, Raman Spectroscopy, and NIRS). Each of these techniques offers unique advantages and limitations, which highlight the importance of multi-modal approaches and ongoing research to enhance measurement accuracy and reliability.

The sub-chapter detailed the principles and applications of each method. Electrical methods, like the Corneometer and Skicon, measure changes in skin capacitance and conductance to estimate hydration levels. These methods have proven

effective but may lack sensitivity at extreme hydration levels. Bioimpedance techniques, such as those used by the Nova Dermal Phase Meter, measure the skin's resistance to electrical currents, offering insights into hydration and barrier function but can be influenced by ion movements within the skin.

Optical techniques, such as NIR spectroscopy, provide non-invasive and high-resolution measurements by analysing the absorption spectra of water in the skin. Studies using NIR have demonstrated the ability to detect changes in skin hydration with high sensitivity, particularly at wavelengths corresponding to water absorption peaks.

The review also explored multi-modal measurement techniques, which combine multiple sensing modalities to improve the accuracy and reliability of skin hydration assessments. These approaches are becoming increasingly valuable, as they provide more comprehensive data by leveraging the strengths of different methods. For instance, integrating optical and electrical sensors can offer real-time, non-invasive monitoring that is more accurate and adaptable to various applications.

In summary, this sub-chapter has analysed both experimental studies and state-of-the-art devices, encompassing optical and electrical techniques for measuring skin hydration. The motivations for measuring skin hydration, as well as the challenges encountered during these measurements, have been outlined. With the ongoing advancements in wearable technology for healthcare and wellness applications, the development of a reliable skin hydration sensor or wearable device is anticipated to occur soon, marking significant progress in the field.

5.6 Measurement of Body Hydration

5.6.1 Body Hydration Assessment and Vital Signs

Body hydration is a critical factor in maintaining physiological homeostasis, particularly in clinical and athletic settings where precise hydration management can significantly impact health outcomes. The traditional approach to assessing body hydration often involves monitoring vital signs such as heart rate, blood pressure, and body temperature, which can indirectly indicate hydration status. For example, dehydration typically leads to tachycardia (elevated heart rate) and hypotension (low blood pressure) due to decreased blood volume, while overhydration can result in hypertension (high blood pressure) and oedema (swelling) as the body retains excess fluid[21].

Cai and Fan explored the relationship between hydration status and health effects in athletes during exercise. It was found that dehydration leads to significant increases in heart rate and decreases in blood pressure, correlating with reduced plasma volume. These findings support the use of vital signs as early indicators of hydration imbalance, particularly in high-intensity environments like competitive sports[99].

To enhance the accuracy of hydration assessment, modern devices are integrating vital sign monitoring with other physiological measurements. The H2-Bio wearable hydration monitor developed by Epicore Biosystems combines sweat analysis with heart rate monitoring. This wearable technology uses microfluidic sensors to analyse sweat composition in real-time, providing feedback on electrolyte balance and hydration status. Studies conducted on endurance athletes using the H2-Bio monitor have demonstrated its effectiveness in preventing dehydration and optimizing performance, particularly in long-duration events where maintaining hydration is crucial.

Porto et al. conducted a systematic review exploring the effects of hydration on vital signs during and after physical activity. The review analysed data from 33 studies, including randomized controlled trials, focusing on how hydration influences heart rate (HR), heart rate variability (HRV), and blood pressure (BP) in adults. The findings revealed that hydration significantly attenuated the increase in HR during exercise and enhanced HRV post-exercise, indicating better autonomic recovery. In addition, hydration led to a modest rise in systolic blood pressure (SBP) post-exercise but did not affect diastolic blood pressure (DBP). This study underscored the importance of adequate

hydration in modulating vital signs and improving cardiovascular responses during exercise, emphasizing its role in maintaining overall body hydration and supporting physiological function during physical exertion[100].

5.6.2 Invasive Measurement Techniques

Invasive measurement techniques are often considered the gold standard for assessing hydration status due to their high accuracy and direct measurement of body fluids. One commonly used invasive method is plasma osmolality, which measures the concentration of solutes in the blood and provides a clear indication of the body's hydration state. Elevated plasma osmolality is a reliable marker of dehydration, as it reflects the increased concentration of electrolytes when water is lost[22].

Forsyth et al. conducted a comprehensive study on the use of plasma osmolality to assess dehydration in elderly patients. The study compared plasma osmolality measurements with clinical assessments of dehydration and found that the former provided a more accurate and earlier detection of dehydration, particularly in patients with compromised renal function. The study underscored the importance of plasma osmolality in clinical settings, especially for vulnerable populations where dehydration can rapidly lead to severe complications. The hydration characteristics with dehydration definitions from this study is seen in Table 5.1 below[101].

Fluid Intake (in ML)	**Mean (Range) (*SD*)**
24 hr w/i admission	983.68 (50–3040) (607.6)
At midpoint of hospitalization	1348.28 (240–3660) (528.8)
24 hr w/i discharge	950.33 (0–4594) (690.6)
Fluid intake < < 1500/24 hr (at admission)	***N* = 46 (25.3%)**
Hematocrit (%)	
Males	40% (4–50) (6)
Females	38% (24–47) (4)
Males with < < 40% HCT (*n* = 89)	***N* = 48 (53.9%)**
Females with < < 37 HCT (*n* = 96)	***N* = 28 (29.1%)**
Sodium	138 (114–151) (9)
Sodium < < 135 (*n* = 195)	***N* = 21 (10.7%)**
BUN (mg/dl)	20 (4–67) (10)
BUN ≤ 18 (*n* = 191)	***N* = 97 (51%)**
Creatinine (mg/dl)	1 (0.6–3.5) (0.4)
Creatinine ≥ 1.5 (*n* = 195)	***N* = 22 (11.2%)**

Table 5.1: Hydration characteristics with dehydration definitions[101]

Another invasive technique involves the measurement of blood urea nitrogen (BUN) and creatinine levels. These markers are commonly used to evaluate kidney function and hydration status. In cases of dehydration, BUN and creatinine levels increase as the kidneys concentrate urine to conserve water. Smith et al. investigated the correlation between BUN/creatinine ratios and hydration status in patients with chronic kidney disease. The study demonstrated that these markers were highly effective in detecting dehydration and guiding fluid therapy in patients with renal impairment, highlighting their utility in both acute and chronic care settings[102].

In addition to plasma osmolality and BUN/creatinine ratios, the use of central venous pressure (CVP) monitoring is another invasive method for assessing hydration. The application of CVP measurements has been explored in critically ill patients, finding that CVP provided valuable insights into the patient's fluid status, particularly during fluid resuscitation. Findings have emphasized the role of CVP monitoring in guiding intravenous fluid administration, reducing the risk of dehydration and fluid overload, however some state that it is not an accurate predictor of fluid responsiveness[103].

5.6.3 Non-Invasive Measurement Techniques

Non-invasive techniques for assessing body hydration have become increasingly popular due to their ease of use and ability to provide continuous monitoring without discomfort. Bioelectrical impedance analysis (BIA) is one of the most widely used non-invasive methods, which estimates body water content by measuring the resistance of body tissues to a small electrical current. BIA is commonly used in clinical settings, fitness centres, and by athletes to monitor hydration levels. Studies have been conducted comparing BIA measurements with plasma osmolality in a cohort of athletes. The results indicated that the accuracy of BIA was affected by factors such as recent food intake and physical activity, necessitating careful control of these variables during measurement[104].

Furthermore, a study by Hahn et al focused on evaluating the effectiveness of BIA as a non-invasive technique for estimating body fluid volumes, specifically extracellular (ECV) and intracellular (ICV) volumes. The methodology involved comparing BIA measurements with fluid volumes calculated using two solute-based equations, under

controlled conditions of fluid shifts induced by hypertonic solutions. BIA, a widely used non-invasive technique, works by measuring the resistance of body tissues to electrical currents, which can then be used to estimate body water content. The study highlighted BIA's appeal as a non-invasive method, particularly for continuous monitoring, but also revealed limitations in its accuracy, particularly in critical care settings where precise fluid management is crucial. The findings suggest that while BIA is a useful tool for assessing hydration and fluid status, it should be complemented with other methods for more accurate results[105].

Near-infrared spectroscopy (NIRS) is another non-invasive technique that has gained attention for its ability to measure tissue hydration by analysing the absorption of near-infrared light by water molecules in the body. Qassem et al. has conducted extensive research highlighting NIRS as a promising non-invasive method for monitoring skin hydration. The technique leverages the absorption characteristics of water in the near-infrared region, allowing for accurate assessment of hydration levels. NIRS has been particularly effective in detecting early signs of dehydration, which can be crucial in preventing performance decline in athletes. The method's ability to provide continuous and accurate data makes it a valuable tool for optimizing hydration strategies during physical activities[106], [107].

Wearable hydration monitoring devices are also playing an increasingly important role in non-invasive hydration assessment. A study by Sandys et al focused on the development and application of a multi-mode optical sensor device called Sixty. Sixty was designed to measure hydration levels accurately by analysing both urine and saliva samples. The device utilizes NIRS, visible light spectroscopy, and refractometry to assess hydration status. The NIR spectroscopy component specifically targets the O-H bonds in water molecules, allowing for the detection of hydration levels by analysing the absorption characteristics of the samples. The methodology also involved collecting urine and saliva samples from participants, which were then analysed by the device to determine their hydration status. The multi-modal approach ensured that the sensor can accurately detect variations in hydration by leveraging the different sensitivities and penetration depths of the various optical methods. The study demonstrates the effectiveness of this technology in providing a non-invasive, real-time assessment of hydration, which would be valuable in clinical and athletic settings[108].

Tsai et al proposed a novel approach to assess skin hydration and sebum content without direct contact with the skin. Traditionally, these measurements have been conducted using contact-based instruments that detect variations in electrical resistance through probes attached to the skin. However, this method carries risks of contamination and infection. To address these issues, a non-invasive technique using optical imaging was developed. The method involved irradiating the skin with green light (528 nm) and near-infrared light (770 nm) and capturing the reflected images. By analysing the differences in the reflected light, the researchers could accurately evaluate skin hydration and sebum content. The study showed that the results obtained through this noncontact method closely matched those from traditional contact methods, with less than a 3% difference in sebum content measurement. Additionally, the method allows for the creation of two-dimensional sebum distribution maps, providing comprehensive insights into skin quality. This technology has potential applications in both the medical and beauty industries, offering a safer and expansive way to monitor skin health[109].

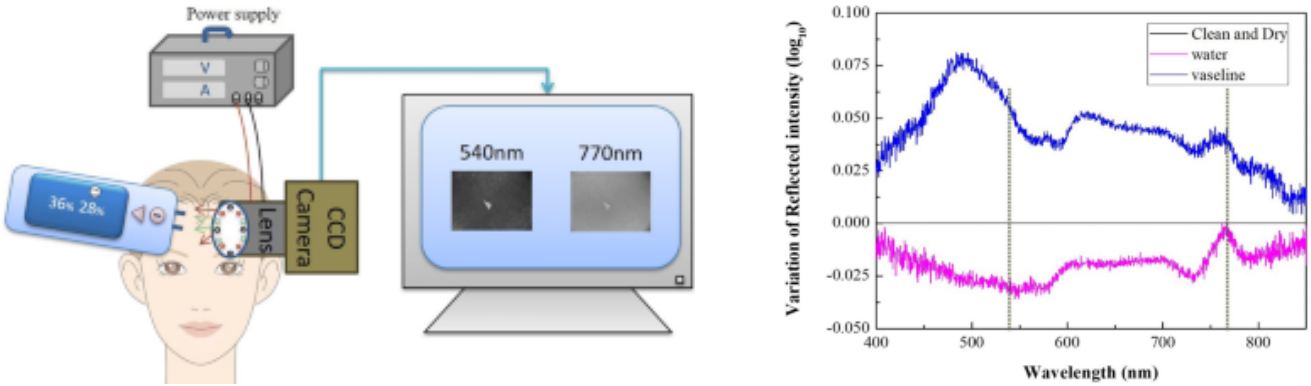


Figure 5.24: left) Experimental setup. right) Variation of reflected intensities between water and Vaseline with respect to clean and dry skin[109]

A further novel technique was explored in a study by Kamran et al, which assessed hydration levels using wearable technology that captures heart rate responses during orthostatic movements. These movements involve changes in posture, such as standing up from a lying position, which affect blood volume and pressure—key indicators of hydration status. The wearable sensors continuously monitor these physiological responses, and the data is processed using logistic regression models to estimate dehydration, specifically a 2% loss in body weight. This non-invasive technique provides a real-time, accurate measurement of hydration levels, making it particularly useful in athletic settings where traditional hydration assessments may be less practical. By leveraging the continuous data from these sensors, the method allows for early detection and intervention in dehydration cases[110].

5.7 Summary of Body Hydration Measurement Techniques

This section has provided a review of various techniques used to assess body hydration, highlighting the importance of accurate and reliable measurement methods in both clinical and everyday settings. The discussion has included both invasive and non-invasive methods, each with its own advantages and limitations.

Invasive methods, such as plasma osmolality and BUN/creatinine ratios, remain the gold standard for hydration assessment due to their high accuracy and direct measurement of body fluids. Studies underscored the importance of these methods in high-risk populations, such as the elderly and patients with chronic kidney disease, where accurate hydration assessment is critical for preventing severe complications. Non-invasive techniques, including BIA and NIRS, offer the advantage of continuous monitoring without the need for invasive procedures. Studies have demonstrated the utility of these methods in both clinical and athletic settings, providing real-time data that can be used to optimize hydration strategies and prevent dehydration. The development of wearable hydration monitoring devices represents a significant advancement in non-invasive hydration assessment, improving health outcomes.

Overall, the ongoing advancements in hydration measurement techniques, both invasive and non-invasive, are paving the way for more accurate, reliable, and user-friendly methods to assess and manage body hydration. These innovations are particularly important in high-risk populations, where maintaining optimal hydration is essential for overall health and performance. As technology continues to evolve, the integration of multiple measurement modalities into a single device holds promise for even greater accuracy and convenience in hydration monitoring.

5.8 Summary of Literature Review

As outlined in the literature review, the evaluation of hydration measurement techniques revealed several important considerations that directly informed the experimental methodology and calibration processes used in this study. A key outcome of the review was recognising the variability in skin hydration across different individuals, influenced by factors such as anatomical site, ethnicity, age, and environmental conditions. These factors often contribute to inconsistent results across

studies and devices. To mitigate this, the calibration approach developed for the optical hydration sensor integrated user-specific parameters, including Fitzpatrick skin type, ethnicity, sex, age, anatomical location, and moisturisation status. By embedding these variables into the calibration model, the system accounts for inter-individual variability, enabling more accurate and personalised hydration assessments. This directly addresses gaps identified in the literature where many existing devices either ignore such factors or apply only generic calibration routines, leading to reduced accuracy across diverse populations. Furthermore, best practices identified from the review highlighted the importance of using controlled environmental conditions during data collection. Fluctuations in ambient temperature and humidity can significantly affect skin surface hydration and the stability of optical signals. Informed by these findings, in vitro, ex vivo, and in vivo experiments were conducted within carefully monitored environmental conditions to minimise external variability. Temperature and humidity levels were kept consistent throughout measurements, particularly during ex vivo testing and human trials. This approach aligns with methodologies described in comparative studies, which demonstrate improved reproducibility when environmental parameters are stabilised.

Another key insight from the review was the value of employing reference measurements to support calibration. Studies using electrical methods such as capacitance or TEWL often include reference phantoms or standardised hydrated materials to benchmark device outputs. This best practice was translated into the experimental design by incorporating baseline sensor readings prior to each hydration measurement. By first recording the optical response of the skin at a known baseline state and comparing this to post-intervention readings (for example, following dehydration protocols), the system could calculate relative changes in hydration more reliably. This method reduced the impact of transient signal drift and enhanced the calibration model's accuracy when applied across multiple sessions and subjects.

In summary, the calibration methodology and experimental design benefited significantly from the review of existing measurement techniques. Best practices, such as compensating for individual variability, controlling environmental conditions, and implementing baseline referencing, were integrated into the sensor system. These measures ensured that the measurements obtained were not only accurate and repeatable but also robust when applied across different user groups, directly addressing limitations found in existing commercial hydration monitoring systems.

CHAPTER 6:

CURRENT STATE OF THE ART FOR NON-INVASIVE MEASUREMENT OF HYDRATION

6.1 State of the Art Devices for Skin Hydration Measurement

Table 6.1: State of the art devices for skin hydration measurement

Device	**Company**	**Technique**	**Description of Measurement Process**
Corneometer (CM825)	Courage & Khazaka	Conductance	Measures the change in the dielectric constant due to skin surface hydration changing the capacitance of a precision capacitor
MoistureMeterSC	Delfin Technologies	Capacitance	Uses a precise (1.25 MHz) electromagnetic field to measure the skin's dielectric constant
Skicon-200EX		Conductance	Measures the conductance of a single high-frequency current of 3.5 MHz [68]
Nova Dermal Phase Meter (Novameter)	NOVA Technology Corp.	Impedance	Measures skin impedance to provide a non-invasive, objective, reproducible method of measurement to quantify biophysical characteristics and relative hydration of the skin
SkinChip	L'Oreal	Capacitance	Based on an active capacitive pixel-sensing technology where the effective feedback capacitance is modulated by contact with the skin close to the surface of the sensor [67]
MoistureMap	Courage & Khazaka	Capacitance	Features a sensor based on capacitive-touch imaging technology where the sensor gives graphical information

			on the near surface hydration distribution and micro-topography of skin
Surface Characterizing Impedance Monitor (SCIM)	U.S. Pat. No. 5353802	Impedance	Integrates readings taken at separate frequencies of the applied alternating current to generate impedance-based values [111]
Tewameter	Courage & Khazaka	Open-Chamber Transepidermal Water Loss (TEWL)	Uses open chamber method of TEWL measurement based on the diffusion principle in an open chamber to measure moisture at two different sites
DermaLab	Cortex Technology ApS	Open-Chamber TEWL	Relative humidity and temperature sensors centred in open chamber for continuous real-time readout of TEWL
Evaporimeter	ServoMed	Open-Chamber TEWL	Measures TEWL by estimating the vapour pressure gradient of water immediately adjacent to the skin surface [112]
Vapometer	Delfin Technologies	Closed-Chamber TEWL	Monitors the increase of relative humidity (RH) inside the chamber and the evaporation rate value (g/m^2h) is automatically calculated from the RH increase
Biox Aquaflux	Biox Systems	Condensed-Chamber TEWL	Overcomes challenges of both open and closed-chamber methods with a condenser that continuously removes water vapour by converting it to ice

Dermal Torque Meter	Dia-Stron	Skin Elasticity	Involves induction of a given amount of stress using a rotating disc adhered to the skin, and then measuring the angular displacement of the resulting skin deformation
Twistometer	Dia-Stron	Skin Elasticity	Involves induction of a given amount of stress using a rotating disc adhered to the skin, and then measuring the angular displacement of the resulting skin deformation [106]
Cutometer	Courage & Khazaka	Skin Elasticity	Uses a suction-based measurement principle using negative pressure and a non-contact optical measuring system at the probe aperture
Dermaflex	Cortex Technology	Skin Elasticity	Uses a suction-based measurement principle but employs a proportional full-thickness strain method rather than the disproportional superficial strain system
Raman Skin Analyzer 3510	RiverD International B.V.	Confocal Raman Microscopy	A sample is illuminated with a low power laser light, where Raman scattering occurs in which the energy of the incoming light is transferred to a molecule, thereby exciting the molecule's vibrational modes and output a direct spectrum

6.2 Recent Advances in the Development of Sensors and Wearable Devices for Body Hydration Assessment

Wearable technology has seen significant advancements in recent years, driven by the growing interest in personal health monitoring. Among the various parameters that can be tracked, hydration is crucial for maintaining overall health and performance. Accurate hydration assessment can help prevent dehydration-related issues, optimize athletic performance, and improve general well-being. This chapter reviews the recent advances in the development of sensors and wearable devices designed to monitor body hydration.

The desire to incorporate biosensors into wearable devices has increased significantly, leading to the monitoring of vitals such as heart rate and sleep cycles. Companies have recently leveraged the understanding of how skin hydration can be accurately measured to integrate this capability into wearable devices. The most commonly used measurement techniques include sweat monitoring via ions and electrolytes and measuring biomarkers from bodily fluids. However, optical and multi-modal techniques are becoming more explored in the wearable market.

A company known as Halo Wearables developed a non-invasive device known as the Halo Edge to monitor a user's hydration levels at a cellular level. Multi-modal techniques are used in this device, combining optical and impedance-based techniques. Using either of these modalities alone can provide a good estimate of hydration but not an accurate measurement. The process involves a learning and training method on the user via baselining. This is achieved by initially wearing the device for a couple of days to allow the algorithm to learn the individual's independent hydrative state rather than relying on a general baseline comparison. The device is worn on the wrist and monitors hydration via the assessment of dehydration measured from the individual's sweat. The sensors in the device monitor fluid levels in the blood and provide a real-time measure of hydration based on a 'Halo index scale' ranging from 1 to 100. This index scale is sectioned into different zones, each indicating the need for hydration. The target audience for the Halo Edge device is those training in sports; however, plans are being made to expand its use to military and medical fields, as well as for personal monitoring. Further developments aim to include factoring in metabolic rates of the body to achieve a more comprehensive outlook on the individual's bodily processes and energy expenditure.

Figure 6.3: Halo Edge wearable device

Another company conducting this is Nix Biosensors, which collaborated with Harvard University to develop a single-use hydration sensor primarily used for alerting low hydration levels in athletes. The feedback information is actionable, allowing users to track their personalized hydration requirements. The developed device uses a sweat-based biometric sensor to monitor changes in biomarkers present in body fluids. This approach focuses on monitoring critical biomarkers such as sodium, potassium, and chloride ions, which are indicative of the body's hydration status.

Figure 6.5: NIX Biosensors wearable hydration sensor

Furthermore, hDrop is a developed wearable device that monitors the body's electrolytes, temperature, and hydration levels. The measurement process involves obtaining a voltage from the sweat response to acquire the electrolyte concentration, as dehydration is directly related to a loss in electrolytes. The sensor conducts a small current through the skin surface to measure the electrical resistance of the detected electrolytes. The device can be linked to an app via Bluetooth, and data can be stored for up to 4 hours. This continuous monitoring capability allows users to receive real-time alerts and historical data analysis.

Figure 6.6: hDrop wearable hydration monitor

Sixty is another wearable device in a watch-like form that measures an individual's heart rate and hydration levels using optical spectrometry. This optical measurement technique consists of 3 LEDs and a photodiode. The LEDs are also used to periodically alert the individual of their dehydration severity through an LED sequence or vibrations. In addition to providing continuous and real-time measurements of hydration levels, a companion app has been developed. When linked to other smart devices, the app provides suggestions and advice on symptoms and serves as a store for later analysis of measured behaviour patterns. The combination of hydration measures and heart rate monitoring via optical techniques allows for increased accuracy in the provided readings.

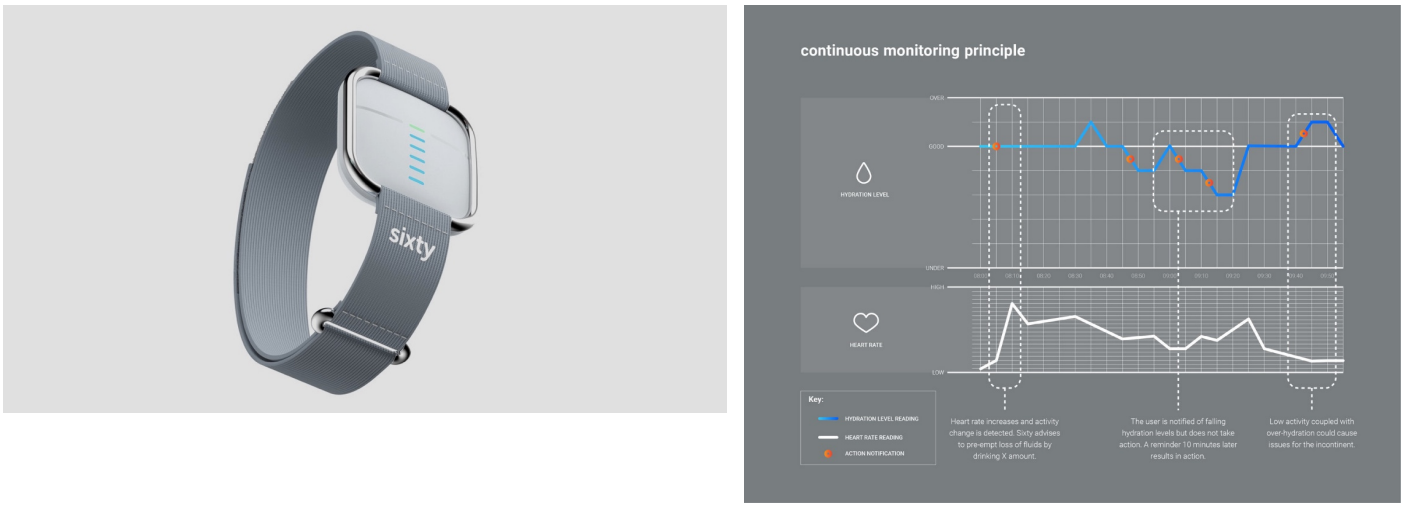


Figure 6.7: left) Sixty wearable hydration monitor. right) Sixty wearable hydration monitor - continuous monitoring principle for hydration levels and heart rate

Additionally, BSX Athletics developed a device known as LVL, which uses LEDs in the near-infrared wavelength range to simultaneously measure heart rate and dehydration on the wrist. This optical method is combined with the measurement of sweat to provide real-time feedback on bodily fluid requirements. The target audience for this device is athletes, and it has been validated on over 250 volunteer athletes with comparisons to the gold standards for hydration measurements, such as blood, urine, and gravimetric techniques. Bluetooth is used to link the device to other smart devices, and it comprises an OLED screen for direct feedback to the user. BSX Athletics previously developed a product known as BSXinsight, used for the measurement of lactate threshold. The LVL device uses the same optical technology as this device, with further improvements built upon the initial methodology.

A team known as Sweatration consists of a group of students from Syracuse University who investigated the issue where 80% of NCAA athletes reported suffering from dehydration. They invented a hydration monitoring wearable that can notify the user when they become dehydrated. The process works on the understanding of a spike in sodium ions in relation to high dehydration levels, forming the methodology basis for the device. The spike in dehydration is monitored through the sodium ion conductivity levels. A prototype of the device has been developed but requires further refinements for market use.

The MASIMO W1 is an advanced health-tracking wearable designed to provide a comprehensive suite of physiological measurements, including the innovative Hydration

Index (Hi). This index is crucial for understanding and managing daily hydration levels, which significantly impact overall health and performance. In addition to the Hydration Index, the MASIMO W1 tracks oxygen saturation, heart rate, respiration rate, pleth variability index, and perfusion index, offering users a broad overview of their health. This device is particularly beneficial for individuals who are keen on monitoring their health data independently, making informed fitness decisions, and understanding how their hydration levels correlate with their overall well-being.

The MASIMO W1's Hydration Index is recognized as a vital metric for daily health management, helping users establish and monitor their hydration baseline. This feature allows users to better understand how their hydration status affects their physical performance and daily health decisions. The MASIMO Health App complements the wearable by providing detailed patterns and trends based on the data collected, enabling users to make informed decisions about their health on a daily basis. This integration of cutting-edge technology and user-friendly data presentation makes the MASIMO W1 a powerful tool for those looking to enhance their health and wellness.

Figure 6.5: MASIMO W1 Wearable Watch

The WHOOP 4.0 is a state-of-the-art wearable designed to track and optimize physical and mental health by measuring metrics scientifically proven to impact overall well-being. This device stands out for its exceptional accuracy, particularly in heart rate and heart rate variability (HRV) tracking, which exceed 99% accuracy, as well as its gold-standard sleep tracking capabilities. The WHOOP 4.0 features an advanced sensor array that includes five LEDs (three green, one red, and one infrared) and four photodiodes,

which work together to capture detailed health data. These sensors use light-based technology to measure physiological parameters, with the accuracy of the device validated through rigorous lab studies and third-party testing.

A unique feature of the WHOOP 4.0 is its ability to track hydration indirectly through its impact on heart rate variability and resting heart rate. Users can monitor their hydration habits using the WHOOP Journal, which allows them to track behaviours and observe their effects on sleep and recovery. Data collected by WHOOP shows that maintaining proper hydration can lead to measurable improvements in HRV and reductions in resting heart rate. On average, users who report sufficient hydration experience a 4.5-millisecond increase in HRV and a 1.7-beats-per-minute decrease in resting heart rate, highlighting the physiological benefits of adequate hydration.

Figure 6.6: WHOOP 4.0 Wearable Band

Sweat analysis has also become a popular method for non-invasive hydration monitoring. Sweat contains various biomarkers, including electrolytes and metabolites, which can provide valuable information about the body's hydration status. Wearable sweat sensors typically involve ion-selective electrodes, electrochemical sensors, and microfluidic systems to collect and analyse sweat samples in real-time[113].

One of the challenges in sweat-based hydration monitoring is ensuring accurate and consistent sweat sample collection. Advances in microfluidic technologies have enabled the development of devices that can continuously and non-invasively collect sweat, minimizing the need for manual sample collection and improving measurement reliability. These systems can also control the flow rate and volume of sweat, ensuring that the sensors receive an adequate sample for analysis[114].

For instance, a recent study by Choi et al. developed a flexible and stretchable sweat sensor integrated into a wearable patch. This device can continuously monitor sweat electrolyte levels, including sodium and potassium, and transmit the data wirelessly to a smartphone application. The sensor employs ion-selective electrodes and electrochemical transducers to provide real-time feedback on hydration status, enabling users to adjust their fluid intake accordingly[115].

Optical methods for hydration monitoring involve measuring the interaction of light with skin tissue. Techniques such as near-infrared spectroscopy (NIRS) and optical coherence tomography (OCT) are being explored for their potential to non-invasively assess skin hydration. These methods offer advantages such as high spatial resolution, the ability to penetrate deeper skin layers, and minimal discomfort for the user[64], [106], [116].

Multi-modal sensors combine different measurement techniques to provide a comprehensive assessment of hydration status. Integrating optical methods with electrical and chemical sensing can offer improved accuracy and reliability in devices. For example, a multi-modal sensor can use NIRS to measure skin hydration and simultaneously impedance spectroscopy to assess fluid distribution in deeper tissues[117].

6.3 Future Directions and Challenges

The development of wearable hydration sensors is advancing rapidly, with numerous promising technologies emerging to address the growing need for real-time, accurate body hydration assessment. However, the path toward fully functional and widely adopted wearable hydration devices is fraught with challenges that require careful consideration and innovative solutions.

One of the primary challenges is calibration and standardization. To ensure that wearable devices provide accurate and consistent measurements across different users and environments, rigorous calibration and standardization protocols are essential. Currently, there is a lack of universally accepted calibration methods, which can lead to variations in device performance. Existing technologies often fail to account for significant differences in skin types, sweat composition, and environmental conditions such as

temperature and humidity, resulting in inconsistent hydration measurements across diverse populations. This research directly addresses this gap by developing a robust calibration model that accounts for variations in gender, skin type, and ethnicity. The model is designed to improve the accuracy and reliability of hydration measurements, ensuring consistent performance across a broad range of users, which is a significant advancement over existing methods[118].

Integration and miniaturization are also critical areas of focus. As wearable devices become more complex, there is a need to integrate multiple sensing modalities into a compact and comfortable form factor. This includes combining sensors that measure not only hydration but also related physiological parameters such as heart rate, body temperature, and electrolyte levels. Advances in flexible electronics, microfabrication, and material science are key to achieving these goals. The development of flexible, skin-compatible materials and the miniaturization of electronic components will enable the creation of devices that are both powerful and unobtrusive, enhancing user comfort and compliance. My research builds on these developments by integrating a multi-wavelength near-infrared (NIR) optical sensor, which provides a more comprehensive and detailed assessment of skin hydration compared to existing single-parameter electrical methods. The multi-wavelength approach allows for a higher level of specificity and depth resolution, which is critical for accurately assessing hydration in different skin layers and overcoming the limitations of surface moisture interference seen in other techniques[117], [118].

Another significant challenge is data management and privacy. Wearable hydration sensors generate large amounts of data that need to be securely stored, processed, and analysed. As these devices become more integrated into medical and personal health applications, ensuring user privacy and data security becomes increasingly important. Robust encryption methods, secure data transmission protocols, and compliance with health data regulations are necessary to protect sensitive information. Moreover, the ability to efficiently manage and analyse large datasets will be crucial for providing meaningful insights and personalized hydration recommendations. This research contributes to addressing these challenges by incorporating data management strategies that align with current health data regulations, ensuring both user privacy and data integrity during real-time monitoring[119].

User compliance and comfort are vital for the effectiveness of wearable hydration sensors. For these devices to be widely adopted, they must be comfortable to wear and easy to use. Design considerations such as device weight, battery life, and skin compatibility are important factors that influence user compliance. Innovations in battery technology, such as the development of low-power sensors and energy-harvesting technologies, can extend device battery life, making wearables more practical for long-term use. Additionally, the use of biocompatible materials can reduce skin irritation and improve user comfort, further enhancing compliance. My research advances this area by focusing on the miniaturization and integration of the NIR sensor into a skin-compatible, lightweight form factor that ensures user comfort during extended use, while also ensuring accuracy and non-invasiveness, making it suitable for clinical and personal environments[118].

Clinical validation remains a significant hurdle for many emerging wearable hydration sensors. While these devices show promise in laboratory settings, extensive clinical validation is necessary to confirm their efficacy in real-world scenarios. Longitudinal studies involving diverse populations, including different age groups, genders, and individuals with varying health conditions, will help establish the clinical utility of these devices. Such studies are crucial for gaining regulatory approval and ensuring that the devices can provide reliable data across a wide range of users. My research addresses this by conducting rigorous in vivo testing in real-world conditions, with data validation against established hydration assessment methods. This ensures that the developed device can reliably track hydration changes across a variety of use cases, such as exercise, clinical treatments, and daily skin care routines[120].

Finally, cost and accessibility are critical factors that will determine the widespread adoption of wearable hydration sensors. Making these devices affordable and accessible to a broad range of users is a significant challenge. Developing cost-effective manufacturing processes and exploring economies of scale will be crucial to ensuring that these technologies can benefit as many people as possible. Additionally, partnerships between technology developers, healthcare providers, and insurance companies may help subsidize the cost of these devices, making them more accessible to individuals who could benefit most from continuous hydration monitoring. This research focuses on the use of cost-effective materials and scalable manufacturing processes that could make the multi-

wavelength NIR hydration sensor accessible to a wider audience, enhancing its potential for broad adoption[118].

In summary, while the field of wearable hydration sensors is progressing rapidly, addressing the challenges of calibration, integration, data management, user comfort, clinical validation, and cost is essential for the successful development and adoption of these technologies. My research advances the state of the art by developing a novel multi-wavelength NIR optical sensor that addresses many of these key challenges, offering improved accuracy, reliability, and user comfort compared to existing technologies, paving the way for broader clinical and personal use.

6.4 Summary

This chapter detailed the current advancements in wearable technologies designed to monitor hydration, highlighting the significance and capabilities of various devices and methods. These devices are crucial in providing real-time, non-invasive hydration assessments, beneficial for both clinical and personal health monitoring applications.

Several state-of-the-art devices for skin hydration measurement were reviewed, including the Corneometer (CM825), MoistureMeterSC, Skicon-200EX, and the Nova Dermal Phase Meter (Novameter). These devices utilize techniques such as conductance, capacitance, and impedance to measure skin hydration. For example, the Corneometer measures changes in the dielectric constant due to skin surface hydration altering the capacitance of a precision capacitor. Devices like the MoistureMeterSC and Skicon-200EX use capacitance and conductance, respectively, to provide detailed insights into skin hydration levels. Other devices like the Tewameter and Vapometer measure transepidermal water loss (TEWL), which is crucial for evaluating barrier function.

In recent years, the development of wearable sensors for body hydration assessment has significantly progressed, driven by the growing interest in personal health monitoring. Companies have developed advanced wearable devices incorporating biosensors to monitor vitals such as heart rate and hydration. For instance, Halo Wearables created the Halo Edge, which combines optical and impedance-based techniques to monitor hydration at a cellular level. Similarly, Nix Biosensors and hDrop

developed devices that use sweat-based biometric sensors to monitor electrolytes and other biomarkers indicative of hydration status.

The chapter also highlighted advancements in multi-modal sensors that integrate optical, electrical, and chemical sensing techniques to provide a comprehensive assessment of hydration status. For example, optical methods like NIRS and OCT are being explored for their potential to non-invasively assess skin hydration. These methods offer high spatial resolution and the ability to penetrate deeper skin layers, making them suitable for continuous monitoring applications.

Recent innovations in sweat-based hydration monitoring were discussed, emphasizing the use of flexible and stretchable sweat sensors integrated into wearable patches. These devices can continuously monitor sweat electrolyte levels and transmit the data wirelessly to smartphone applications, providing real-time feedback on hydration status.

However, the development of wearable hydration sensors faces several challenges. Ensuring accurate and consistent measurements across different users and environments requires rigorous calibration and standardization protocols. The integration and miniaturization of multiple sensing modalities into compact and comfortable wearable devices are essential for user compliance and effectiveness. Data management and privacy are critical considerations, particularly as these devices generate large amounts of sensitive information. Extensive clinical validation is necessary to confirm the efficacy of these sensors in real-world scenarios, and making these technologies affordable and accessible to a broad range of users remains a significant challenge.

In summary, the integration of advanced sensing technologies into wearable devices holds great potential for revolutionizing hydration monitoring. These devices can help individuals maintain optimal hydration levels, improve athletic performance, and enhance overall health and well-being. While significant progress has been made, ongoing research and development are essential to address the existing challenges and fully realize the potential of wearable hydration sensors.

CHAPTER 7:

DESIGN & DEVELOPMENT OF QUAD-WAVELENGTH NEAR-INFRARED OPTICAL SENSOR FOR MEASURING HYDRATION LEVELS

7.1 Selected Optical Properties for Design Consideration

In designing an optical sensor for hydration measurement, it is critical to carefully select wavelengths within the near-infrared (NIR) region that are most sensitive to water content. A thorough literature review of previous studies was conducted to identify these optimal wavelengths. The review highlighted distinct absorption peaks in the water absorption spectra, which are indicative of the behaviour of water molecules and their interaction with light at specific wavelengths[64].

The absorption peaks in the NIR region correspond to the overtone and combination bands of the O-H bonds in water. These peaks occur due to the vibrational transitions of the O-H bonds, which are particularly sensitive to NIR light. When NIR light interacts with water molecules, it induces vibrational energy transitions that result in specific absorption peaks. These peaks are critical for identifying and quantifying water content in various materials, including biological tissues[46].

Through this review, it was determined that the most prominent absorption peaks for water are observed around 975 nm, 1300 nm, and 1450 nm. The peak at 1450 nm is especially strong and is associated with the first overtone of the O-H stretching vibration. This wavelength is highly sensitive to water content and is commonly used in NIR spectroscopy to assess hydration levels in tissues. The 975 nm peak corresponds to a combination band, and although it is less intense than the 1450 nm peak, it is still significant for detecting water. The 1300 nm peak is another important wavelength, representing a balance between absorption sensitivity and penetration depth, making it useful for probing deeper layers of tissue[14], [46].

These wavelengths also differ in their penetration depths, which is a key consideration for measuring water content at different levels within the skin. NIR light at 975 nm and 1300 nm can penetrate deeper into the dermis, making these wavelengths suitable for assessing hydration at deeper tissue levels. In contrast, the 1450 nm wavelength, while highly sensitive to water, has a shallower penetration depth, making it more suitable for measuring water content in the epidermis or other superficial layers. This variability in penetration depth allows for a comprehensive assessment of skin and body water content across different tissue layers[14].

Given these findings, the design of the hydration sensor in this study focused on incorporating LEDs that emit light at these critical wavelengths—975 nm, 1300 nm, and

1450 nm. These wavelengths were selected for their proven sensitivity to water content, which is crucial for accurately assessing hydration levels. Additionally, a fourth LED emitting at 1050 nm was chosen to serve as a reference wavelength. The 1050 nm wavelength is less sensitive to water, making it ideal for baseline measurements and ensuring the accuracy of the sensor's readings.

The LEDs will be paired with a single photodiode capable of detecting the full range of wavelengths emitted by the LEDs. The photodiode will convert the absorbed light into an electrical signal, which can then be analysed to determine the water content in the tissue. The integration of these specific LEDs and the photodiode is central to the sensor's ability to provide reliable, real-time hydration data across different layers of the skin.

The source-detector separation in the device was kept deliberately small to prioritise a compact and wearable form factor. However, it is important to recognise that optimal separation is dependent on the penetration depth of each wavelength, with shorter wavelengths such as 975 nm and 1050 nm probing more superficial layers, while longer wavelengths like 1300 nm and 1450 nm reach deeper tissue. A fixed, small separation across all wavelengths presents a trade-off by limiting the ability to maximise depth sensitivity at longer wavelengths, potentially reducing the contribution of deeper hydration signals, particularly in the dermis, while enhancing signals from more superficial layers.

This design choice, while advantageous for miniaturisation, introduces limitations in the system's capacity to resolve depth-specific hydration dynamics. Future improvements could include optimising the source-detector separation relative to each wavelength's penetration characteristics, or integrating variable or multi-distance configurations to better target different skin layers. Computational modelling, such as Monte Carlo simulations, would support this optimisation by identifying separations that achieve an effective balance between penetration depth, signal strength, and device practicality. This reinforcing support from a Monte Carlo model is described later in Chapter 8.

In summary, the selection of these wavelengths was driven by their distinct optical properties and sensitivity to water, ensuring that the sensor can accurately measure hydration levels in both superficial and deeper tissues. This careful consideration of optical properties is essential for the development of an effective and reliable hydration sensor, with potential applications in both clinical and personal health monitoring.

7.2 Hardware Design

A multi-wavelength optical sensor was designed and developed to accurately measure skin and body hydration levels. The sensor utilizes four specific wavelengths in the near-infrared (NIR) spectrum—975 nm, 1050 nm, 1300 nm, and 1450 nm—each selected based on their distinct interaction with water molecules. The primary components of this sensor include four light-emitting diodes (LEDs) and a single central photodiode, integrated into a compact and wearable design.

The photodiode chosen for this configuration is the LAPD-09-17-LCC, an InGaAs (indium gallium arsenide) large-area photodetector. InGaAs technology is particularly well-suited for NIR applications due to its high sensitivity and responsiveness across the 900 nm to 1700 nm wavelength range. This range encompasses the selected LEDs, ensuring the photodiode can efficiently detect the light reflected from the tissue. The large surface area of the photodiode enhances its ability to capture photons, increasing the signal strength and accuracy of the measurements.

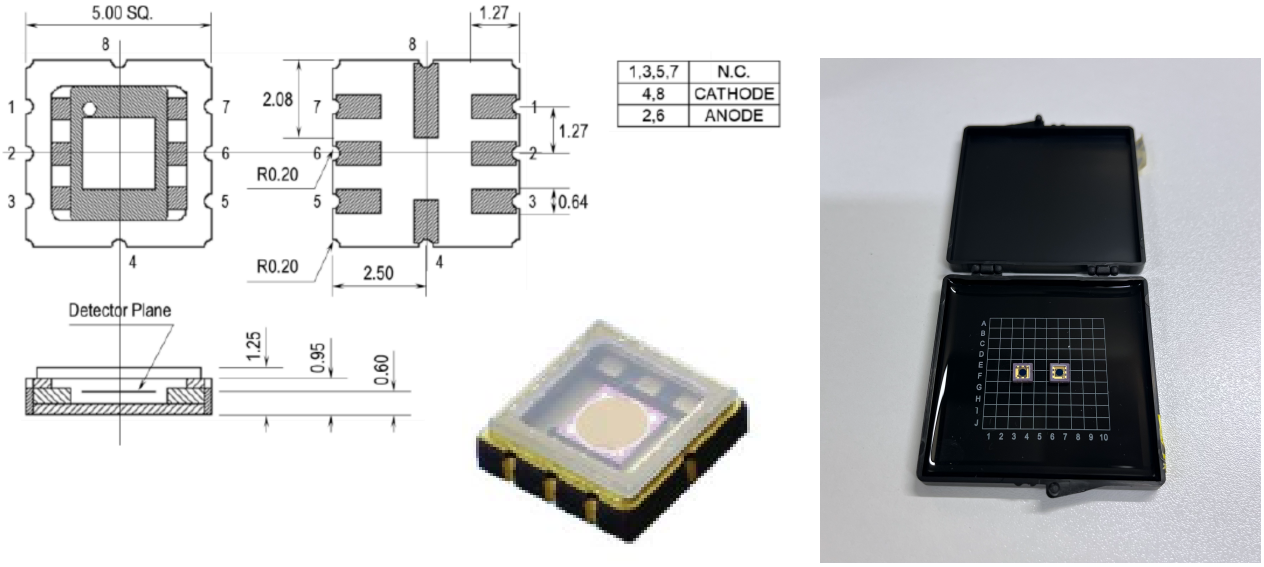


Figure 7.1: LAPD-09-17-LCC InGaAs photodetector

The domed LEDs were selected for their specific wavelengths, each with a unique interaction with water. The 975 nm and 1300 nm wavelengths are known for their deeper tissue penetration, making them ideal for assessing hydration at varying depths within the skin, such as in the dermis. The 1450 nm wavelength is particularly sensitive to water due to the strong absorption by the O-H bonds in water molecules, making it effective for

measuring superficial hydration levels in the epidermis. The 1050 nm LED serves as a reference wavelength, chosen for its minimal interaction with water, allowing for baseline measurements that improve the accuracy of the hydration assessments[14]. Furthermore, the relative sensitivity of the LEDs at the wavelengths of interest (975 nm, 1050 nm, 1300 nm, and 1450 nm) was considered with respect to their peak emission, with sensitivities approximately ranging from 80-90% at 975 nm and 1050 nm, decreasing to around 70-80% at 1300 nm, and further reducing to approximately 60-70% at 1450 nm, which directly informed the optical front-end (OFE) design and the gain settings of the TI amplifiers to ensure adequate signal amplification and noise minimization across all channels.

The placement of these LEDs relative to the photodiode is crucial for optimizing the sensor's performance. The distance between the LEDs and the photodiode influences the penetration depth of the photons into the skin. When NIR light is emitted by the LEDs, it penetrates the skin and interacts with water molecules within different layers. The photons are either absorbed or scattered based on the water content and tissue properties before they reach the photodiode. A carefully controlled distance allows for the differentiation of signals corresponding to different tissue depths, enabling a more comprehensive analysis of skin hydration. This is confirmed using a Monte Carlo simulation later described in Chapter 8.

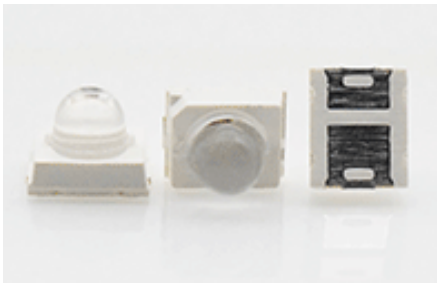


Figure 7.3: Domed-lens Infrared LEDs

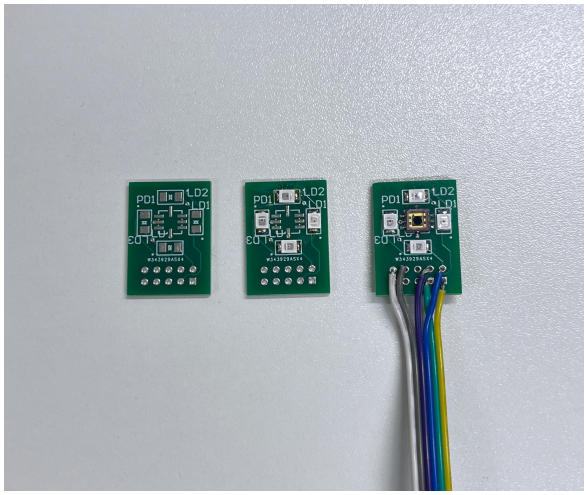


Figure 7.2: PCB with surface-mount LEDs and photodiode soldered

The photodiode is connected to a transimpedance amplifier (TIA) circuit within the ZenPPG processing unit. The TIA converts the small current output from the photodiode, generated by the absorbed photons, into a corresponding voltage signal. The

TIA is a critical component in this system as it amplifies the signal while maintaining a low noise level, ensuring that the small variations in light intensity due to changes in hydration are accurately captured and processed. The amplified voltage signal is then further processed by the microcontroller within the ZenPPG system, which controls the LED operation and manages the data acquisition process[121].

Data from the ZenPPG system is transferred to a computer via a data acquisition card (DAQ), specifically the NI USB-6212. The DAQ card plays a vital role in digitizing the analog signals from the sensor, allowing them to be processed and analysed using LabVIEW software. Within LabVIEW, the signals from the four LEDs are merged and aligned through a DAQ assistant function. The software then filters the signals, extracts the maximum DC voltage values, and plots the output waveforms in real-time. These signals are saved as text files for further analysis in MATLAB, where they can be compared against baseline measurements or other references.

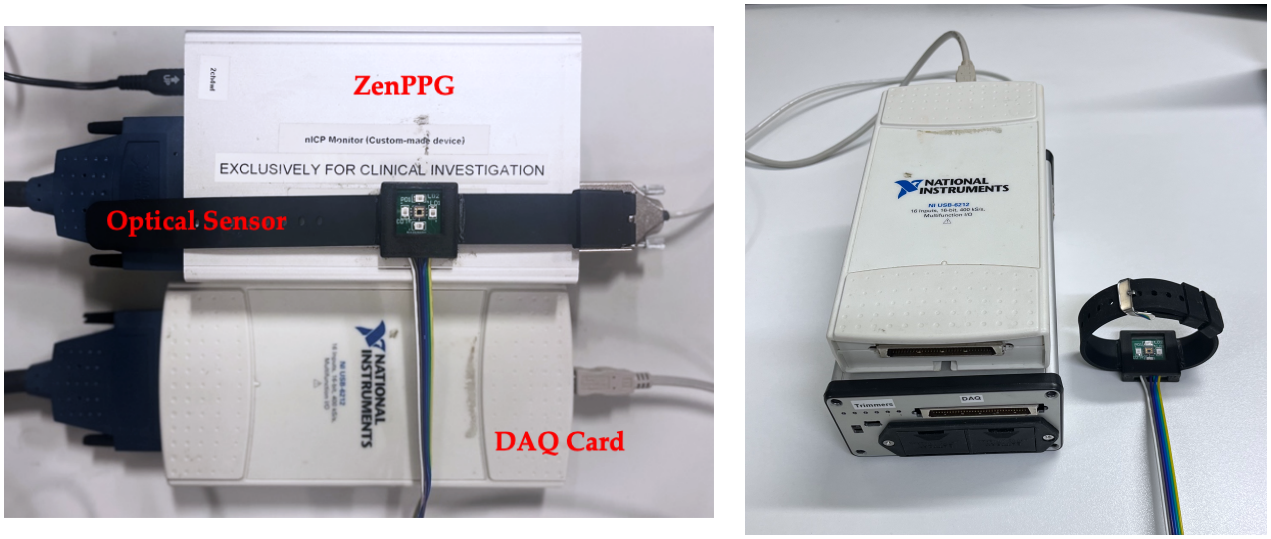


Figure 7.4: Complete device setup: wearable optical sensing device, ZenPPG, and DAQ card

To ensure the accuracy and reliability of the sensor's measurements, the results obtained from the optical sensor via LabVIEW were validated through comparison with gravimetric measurements using a precision weight scale, as well as spectral measurements using an NIR spectrometer (NIRQUEST) and an Avantes light source. The spectral data were pre-processed using Spectragryph software, where reflectance measurements were converted to apparent absorbance spectra, smoothed, and baseline corrected. This data was then analysed alongside the optical sensor recordings in MATLAB, providing a comprehensive validation of the sensor's performance.

Figure 7.7: NIR spectrometer (NIRQUEST)

Figure 7.6: Corneometer CM825 device

The sensor was housed within a 3D-printed casing designed to maintain consistent spacing for the optical window and minimize occlusion effects due to direct skin contact. The optical window, made of Raman-grade calcium fluoride (CaF2), allows NIR wavelengths to pass through with minimal absorption, ensuring that the sensor's readings are not compromised by the material of the casing. This design, coupled with the integration of watch straps, enabled the sensor to be worn comfortably on the wrist, facilitating continuous hydration monitoring as a wearable.

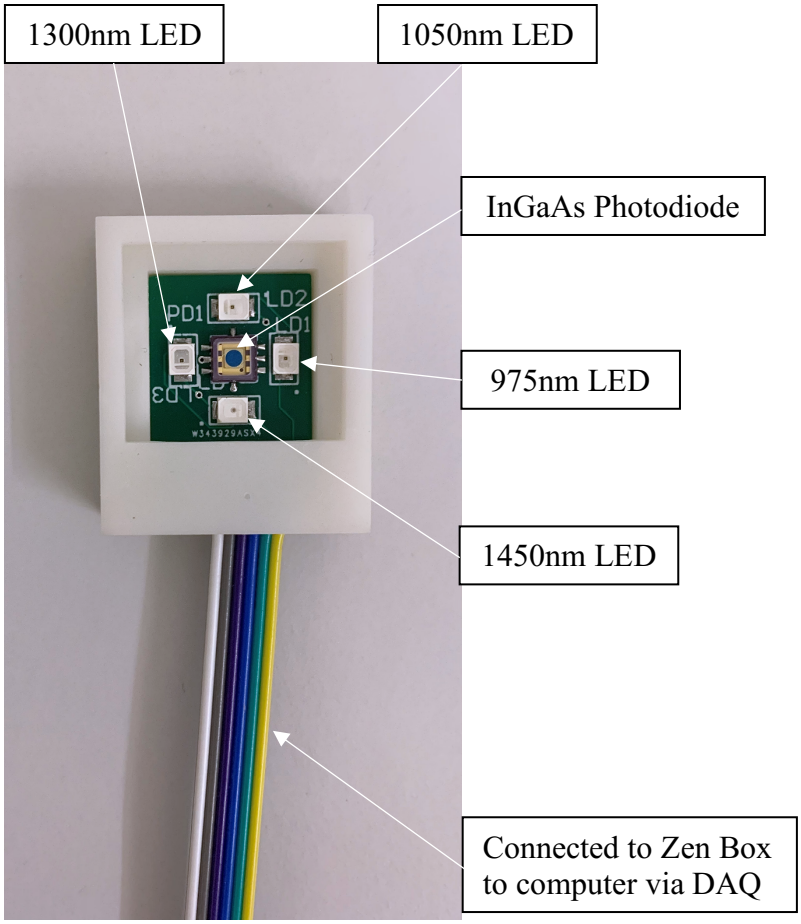


Figure 7.8: Optical sensing device prototype (labelled)

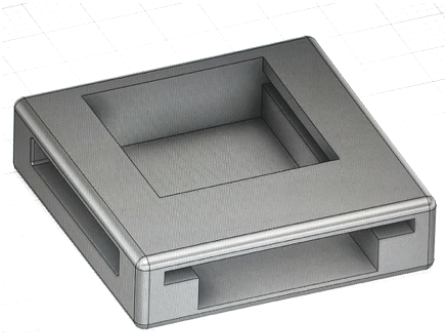


Figure 7.7: 3D-printed casing for PCB and CaF2 optical window in matte black

Figure 7.10: Optical sensing wearable watch

The CaF_2 optical window ensures high NIR transmission and protects the optical path but does not act as an optical barrier to prevent shunting. Optical shunting was mitigated through the careful physical design of the sensor, including emitter-detector spacing and housing geometry, to minimize direct light paths bypassing the skin. However, the absence of a dedicated optical barrier may introduce some level of stray light interference, which could be further reduced in future iterations by incorporating additional optical shielding or collimation techniques.

The ZenPPG system is a multi-wavelength photoplethysmography (PPG) platform designed to drive multiple narrow-band LEDs and sequentially sample the reflected light. It includes programmable current sources to power the LEDs, converting a constant input voltage into a stable drive current. This is critical for maintaining the precision and repeatability of light emission during measurements. The system is equipped with transimpedance amplifiers (TIAs), which convert the small photocurrents produced by the photodetector into time-varying voltage signals suitable for digitisation and further analysis [121].

In this research, a modified version of the standard ZenPPG system was used to accommodate multi-wavelength reflectance monitoring specific to hydration sensing. These modifications included integrating LEDs at the selected near-infrared wavelengths corresponding to water absorption peaks (975 nm, 1050 nm, 1300 nm, and 1450 nm). The current drivers were fine-tuned to deliver up to 200 mA per LED, ensuring adequate penetration depth and signal strength. Switching frequencies were selected in accordance with the Nyquist criterion to allow sequential sampling without temporal aliasing.

To maintain wavelength separation during acquisition, a Sample-and-Hold Amplifier (SHA) was incorporated to isolate and stabilise the voltage response for each LED before conversion. This configuration helped ensure that signal contributions from each wavelength remained distinct and attributable, reducing crosstalk in the time-domain PPG waveforms.

The transimpedance amplifier (TIA) used in the system plays a vital role in ensuring signal fidelity. Its linear response to the wide range of photocurrent levels expected from varying tissue reflectance and light intensities is essential. Preliminary tests of the TIA's output under controlled light exposure demonstrated a linear voltage response up to saturation, confirming its suitability for both low and high reflectance conditions. This linearity is especially important when interpreting subtle changes in

reflectance related to hydration level, where small voltage shifts must correspond proportionally to physiological changes in water content.

However, later comparisons with spectrophotometric measurements in later chapters revealed potential non-linearities in system output, particularly at higher water absorption levels or with probe misalignment. These non-linearities could result from combined effects of LED drive current variation, non-uniform optical coupling, and detector saturation at specific wavelengths such as 1450 nm. As a result, while the ZenPPG system provides reliable relative hydration signals, care must be taken when interpreting absolute values, particularly in high-absorbance scenarios or when comparing across different probe placements.

Overall, the ZenPPG system, with its custom modifications and TIA-based architecture, enables high-resolution multi-wavelength reflectance monitoring. Its performance under variable optical intensities and the inclusion of wavelength-specific signal separation make it well-suited for real-time hydration tracking, with known considerations for potential non-linearity that are accounted for during validation and modelling.

The simplified block diagram for the full optical sensing system is illustrated in the figure below, detailing the flow from the PCB frontend to the acquisition of the signals on the computer.

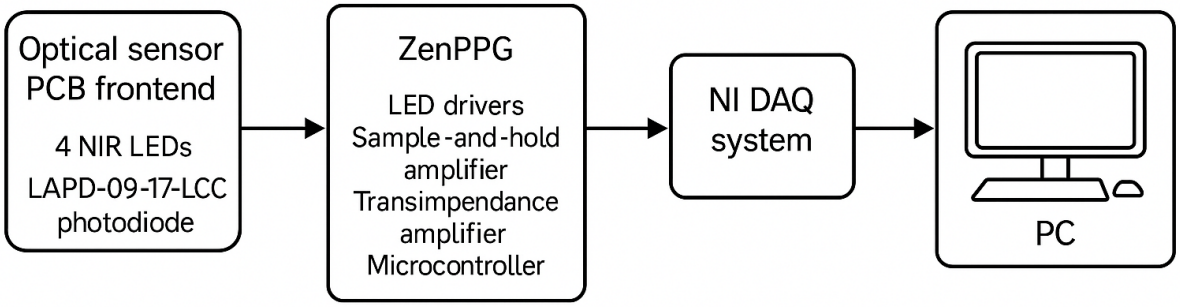


Figure 7.120: Simplified block diagram for full optical sensing system

7.3 LabVIEW Software Design

The preliminary software design for the hydration sensor system is a critical component that enables the real-time acquisition, processing, and visualization of data obtained from the sensor. The developed sensor is interfaced with the ZenPPG system, which includes a transimpedance amplifier to convert the photodiode's current output into a corresponding voltage signal. This voltage signal, representing the intensity of light detected by the photodiode at various NIR wavelengths, is then transmitted to a computer through a Data Acquisition (DAQ) card. The DAQ card, specifically the NI USB-6212, serves as the interface between the analog world of the sensor and the digital realm of the computer, digitizing the analog voltage signals for further processing in LabVIEW.

Within LabVIEW, the DAQ Assistant block plays a crucial role in the data acquisition process. The DAQ Assistant is a built-in feature in LabVIEW that provides a graphical interface for configuring, acquiring, and processing data from DAQ devices. It simplifies the task of setting up the data acquisition parameters, such as sampling rate, input channels, and signal scaling. In this system, the DAQ Assistant is configured to continuously sample the voltage signals at a frequency of 10 Hz, which is sufficient for providing near real-time feedback on hydration levels. The DAQ Assistant collects the signals from the DAQ card and prepares them for further analysis within the LabVIEW environment.

Once the signals are acquired, they are processed by a series of blocks designed to manage and interpret the data. The first step involves splitting the combined signal into separate voltage outputs corresponding to each of the four LEDs (975 nm, 1050 nm, 1300 nm, and 1450 nm). This is achieved using a signal splitting block, which isolates the voltage contributions from each LED based on the input channels configured in the DAQ Assistant. Each of these isolated signals is then analysed to determine its maximum voltage, which is continuously updated and displayed on LabVIEW's front panel as a real-time voltage value. This real-time display is crucial for monitoring the dynamic changes in skin hydration as the sensor interacts with the tissue.

LabVIEW also includes a live signal monitor that visually represents the changes in voltage over time for each LED. This visualization allows users to observe how the detected light intensity varies in response to hydration changes, providing immediate feedback on the sensor's performance. The ability to monitor these signals in real-time is

particularly valuable in experimental settings, where understanding the immediate impact of hydration interventions (such as applying moisturizers or rehydrating) is critical.

In addition to monitoring the LED outputs, the software design includes blocks that manage the power supply to the sensor. Specifically, a monitor displays the voltage provided to the sensor through the ZenPPG system, ensuring that the system is operating within the desired parameters. The software also includes controls for the current sources that power the LEDs, allowing for the adjustment of current levels if necessary. These controls ensure that all LEDs are supplied with a consistent current, which is essential for maintaining the accuracy and reliability of the sensor's measurements.

A preliminary hydration index is also calculated within LabVIEW by summing the outputs of the three water-sensitive LEDs (975 nm, 1300 nm, and 1450 nm) and subtracting the output from the reference LED at 1050 nm. This index provides a quick estimate of the hydration level, which can be useful for immediate assessments without requiring extensive post-processing. The calculation of this index is performed using a simple arithmetic block, which continuously updates the value as new data is acquired.

Finally, the software design includes functionality for recording and saving the acquired data. Users can start, pause, and stop data recording, with the data being saved to a specified file location in a text format. This feature allows for the collection of long-term data, which can be later analysed in other software environments, such as MATLAB, for more detailed studies. The ability to record data is particularly important for validating the sensor against other measurement techniques, such as gravimetric analysis or spectrophotometry, ensuring that the sensor provides accurate and reliable hydration measurements over time.

In conclusion, the software design within LabVIEW plays a crucial role in the operation of the hydration sensor, from data acquisition through the DAQ card to real-time processing and visualization of the signals. The DAQ Assistant simplifies the acquisition process, while the subsequent blocks ensure that the data is accurately processed, monitored, and recorded, providing a comprehensive solution for hydration measurement and analysis. This precursor software design was re-developed, as described next in Chapter 7.4, using a Python model.

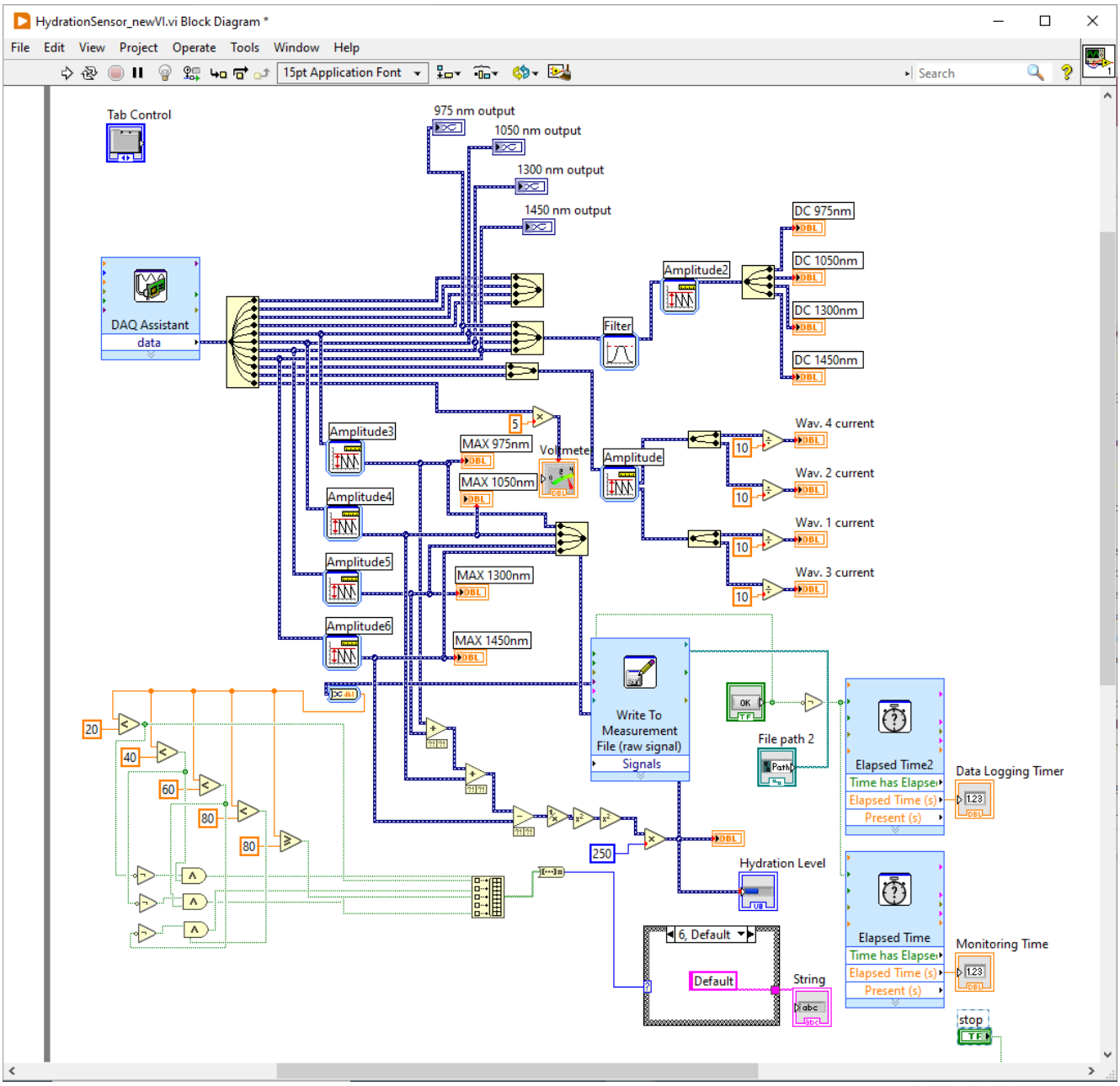


Figure 7.11: LabVIEW block diagram for hydration measurements from optical sensor

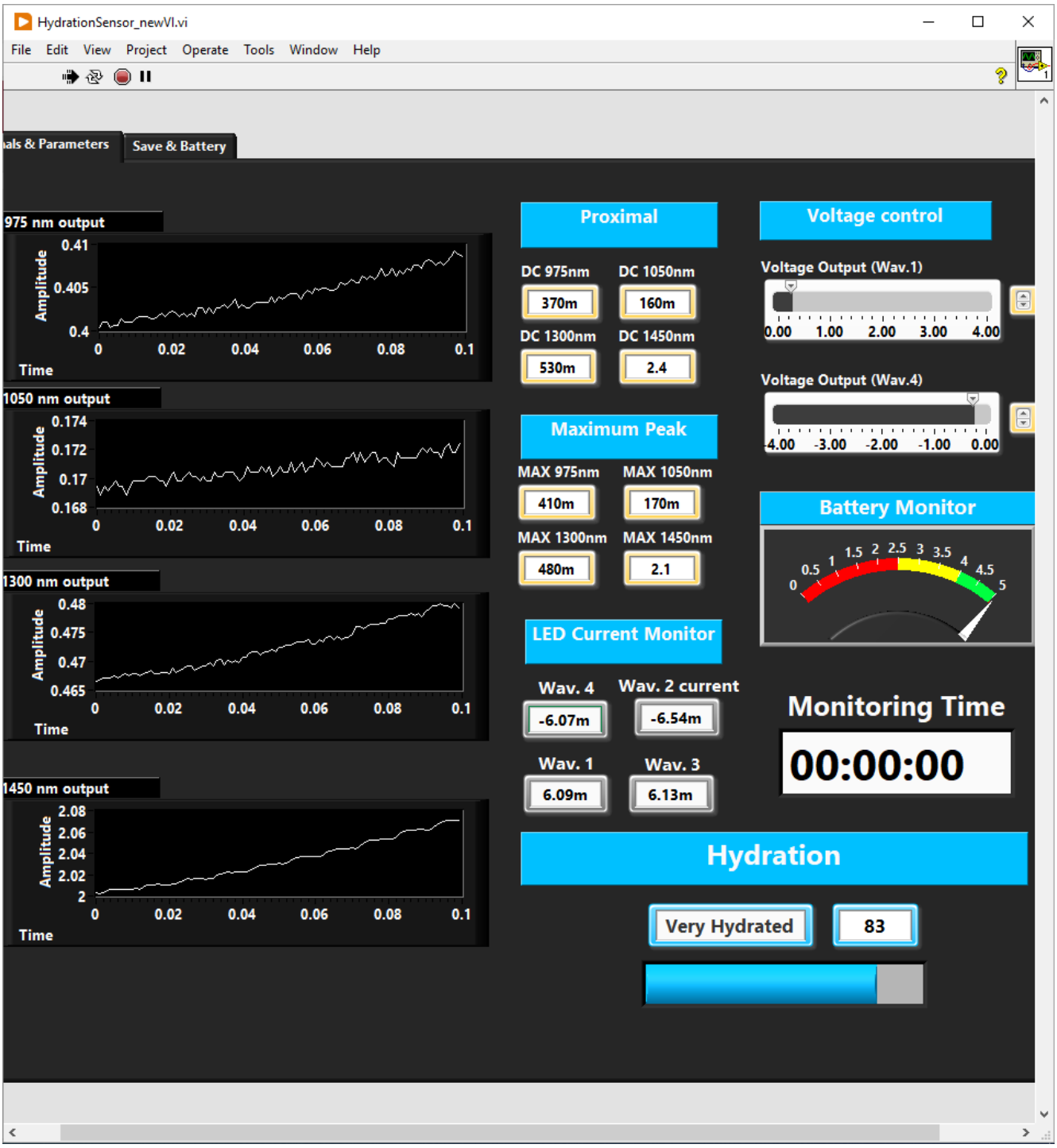


Figure 7.12: LabVIEW front panel for hydration measurements from optical sensor

7.4 Python Baseline Threshold Calibration Model

To determine the baseline hydration level using the output voltages from the four near-infrared LEDs and the photodiode, we can draw parallels with how pulse oximetry works. Pulse oximetry uses ratios of AC (pulsatile) and DC (non-pulsatile) components of light absorption to estimate blood oxygen levels. Similarly, for hydration sensing, the variations in the absorption of light at different wavelengths can be used to infer hydration status. This process involves several phases: calibration, baseline establishment, hydration monitoring, composite hydration index calculation, alert mechanism, and the integration of a mathematical model.

During the calibration phase, voltage readings from the four LEDs are collected continuously over a period when the user is known to be well-hydrated. This phase ensures that the subject maintains optimal hydration levels, providing accurate baseline data. The average voltage for each LED is calculated over the calibration period to establish a baseline, serving as a reference for future hydration assessments.

In the baseline establishment phase, a baseline measurement is taken from the sensor, which provides a voltage for each LED. This is represented mathematically as:

$$\text{Baseline}_i = V_{\text{baseline},i} \qquad \textit{Equation (7.1)}$$

where Baseline_i is the baseline voltage for the i-th LED.

For hydration monitoring, the voltage from each LED is continuously measured and compared to the baseline. This is expressed as:

$$\delta V_i(t) = V_i(t) - V_{\text{baseline},i} \qquad \textit{Equation (7.2)}$$

where $\delta V_i(t)$ is the difference between the current voltage $V_i(t)$ and the baseline voltage $V_{\text{baseline},i}$. The relative change or ratio for each LED is then computed using:

$$R_i(t) = V_i(t) / V_{\text{baseline},i} \qquad \textit{Equation (7.3)}$$

To form a composite hydration index, the ratios from all LEDs are combined using a weighted sum. This is calculated as:

$$H(t) = \text{sum}\ (w\_i * R\_i(t)) \qquad \textit{Equation (7.4)}$$

where H(t) is the composite hydration index at time t, and w_i are the weights assigned to each LED ratio R_i(t).

An alert mechanism is established by setting thresholds based on observed deviations (±10%) from the baseline during the calibration phase or from known hydration levels. An alert is generated if:

$$0.9 * H(t) > H(t) > 1.1 * H(t) \qquad \textit{Equation (7.5)}$$

indicating significant changes in hydration levels that may require attention.

The mathematical model integrates the previously labelled regression-trained model to include user-specific inputs such as ethnicity, age, and other relevant factors. This enhances the accuracy and personalization of the hydration assessment.

Near-infrared (NIR) wavelengths are particularly useful for skin hydration measurement due to their penetration depth and sensitivity to water content. Specific NIR wavelengths correspond to absorption peaks of water, which can be used to accurately assess hydration levels. As mentioned, the wavelengths being used include 975 nm, 1050 nm, 1300 nm, and 1450 nm, each providing different absorption characteristics related to the water content in the skin. NIR light can penetrate several millimetres into the skin, allowing for the assessment of hydration levels in both the superficial and deeper layers of the skin. The penetration depth is influenced by the wavelength of the light and the absorption and scattering properties of the skin. Hydration levels can affect the optical properties of these layers, altering the absorption and scattering of NIR light. Dehydration can lead to increased skin impedance, reduced elasticity, and altered optical properties, which can be detected using NIR spectroscopy. By leveraging the principles of NIR absorption and the developed calibration, this methodology provides a robust framework for continuously monitoring skin hydration levels. The integration of user-specific data

ito the regression model enhances the accuracy and personalization of the hydration assessment, making it a valuable tool for clinical and personal health applications.

This calibration code and hydration measurement process have been integrated into a user-friendly web page interface. Users can input their personal details such as ethnicity, age, and gender, and upload their fixed recorded baseline as well as new readings taken from the sensor at the time. The web page interface provides a seamless way for users to interact with the hydration monitoring system.

To use this system, the user will follow these steps:

1. On the terminal, type: `python hydration_app.py`.
2. The web page will link at `http://127.0.0.1:5000/`.
3. Fill out the user input form and upload measurement files from the hydration wearable saved using LabVIEW VI.
4. Press 'Submit'.

Upon submission, the system processes the input data and displays a hydration index ranging from 0 to 100 on the screen. This result also indicates whether the hydration level is within range, above, or below the user's baseline, providing an immediate and clear assessment of their hydration status.

The flowchart below further illustrates the sequential process followed by the hydration monitoring software, from data acquisition to final hydration assessment. It outlines how sensor readings are collected, processed, and calibrated using machine learning algorithms, incorporating wavelength-specific weighting to extract depth-specific hydration information. The structured pipeline ensures that surface hydration effects are minimized, while personalized calibration enhances accuracy across different users. The final hydration index is computed, analysed against threshold values, and displayed to the user with appropriate alerts for hydration status.

The software architecture for the hydration monitoring system follows a structured pipeline that processes sensor data, applies calibration adjustments, and computes a personalized hydration index. The system begins by collecting voltage signals from four LEDs operating at 975 nm, 1050 nm, 1300 nm, and 1450 nm, which are positioned

equidistantly around a central photodiode. A baseline hydration reading is recorded at the start of each session to normalize subsequent measurements.

To improve accuracy, the sensor readings are combined with user-specific parameters such as age, skin type, and ethnicity. Each wavelength's signal is weighted based on its water absorption coefficient, ensuring that the most relevant signals contribute to the final hydration index. A subtraction method is applied to remove surface hydration effects like sweat and stratum corneum moisture, allowing for a more accurate representation of deeper tissue hydration.

A trained machine learning model further refines the hydration index by adjusting for individual variability. The final hydration status is then compared to predefined thresholds, and alerts are generated if hydration levels fall outside the optimal range. The results are displayed to the user, completing the process while allowing for continuous hydration tracking. This structured approach ensures real-time, depth-specific hydration assessment while maintaining high accuracy through personalized calibration.

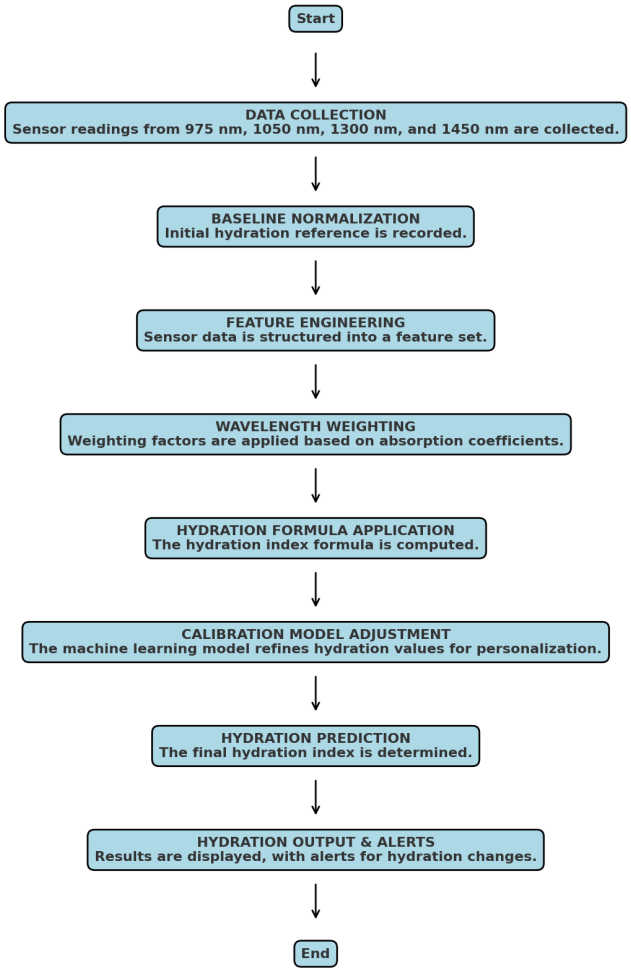
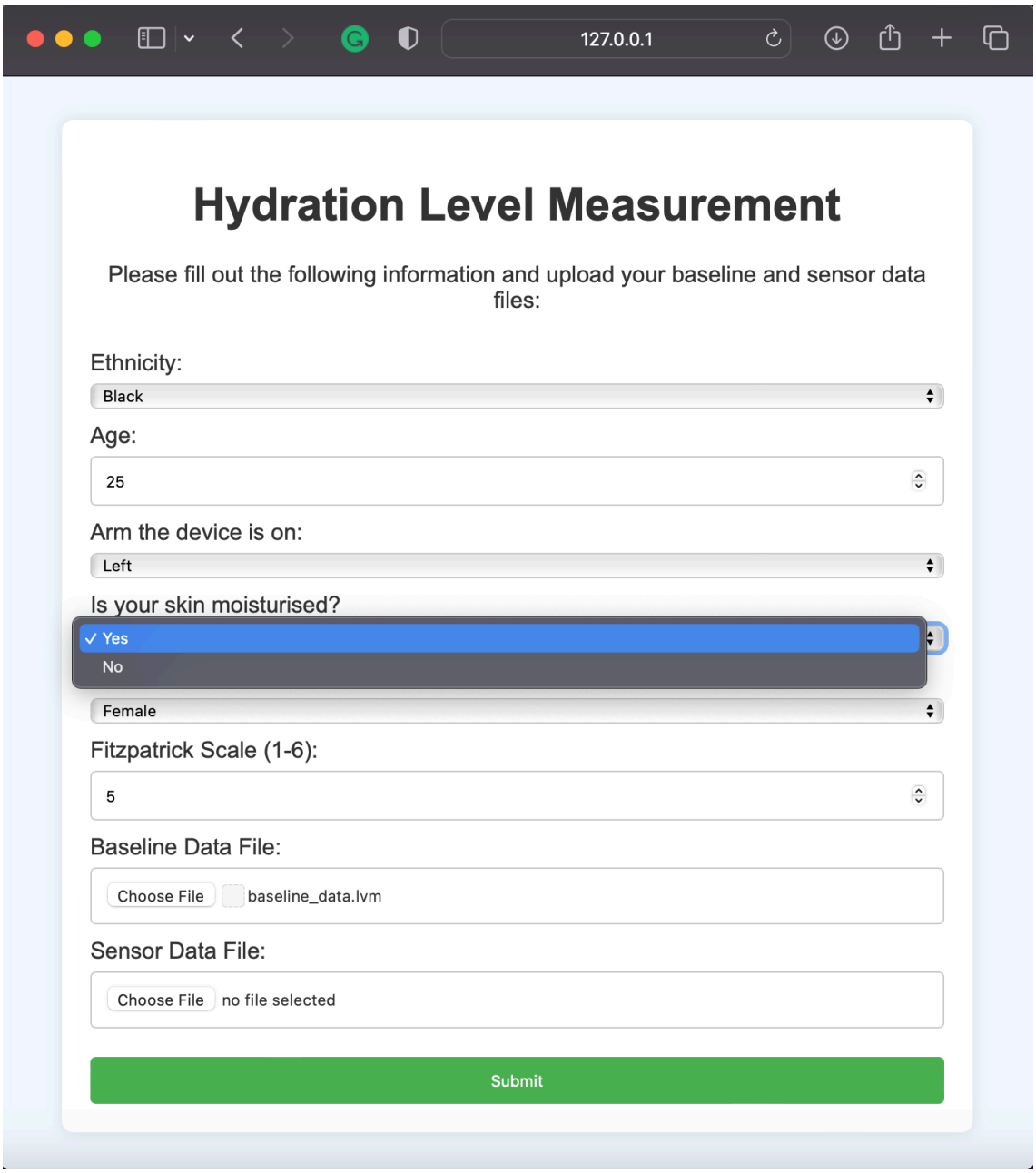


Figure 7.13: Flowchart illustrating hydration monitoring system software process



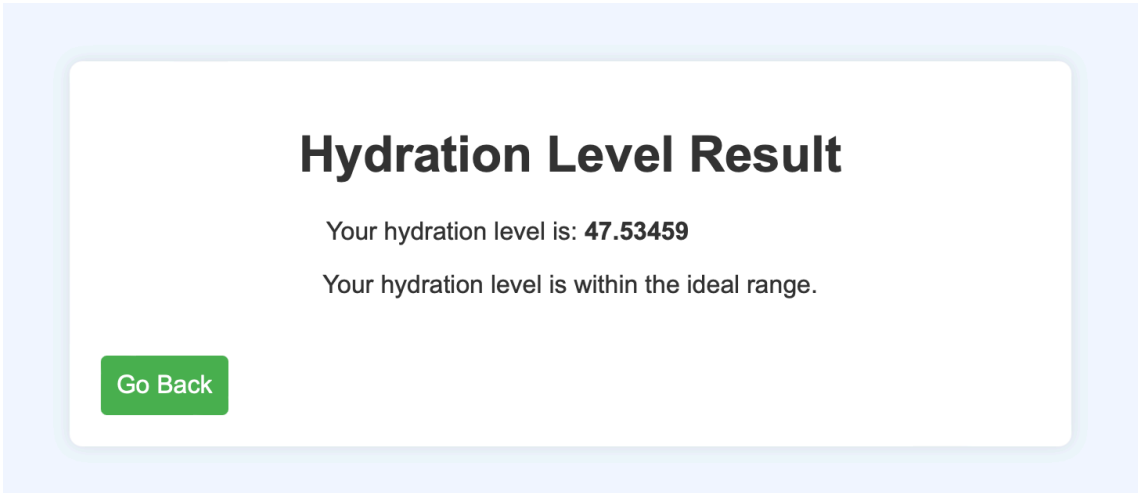


Figure 7.14: Python web page interface for hydration level results

7.5 Summary

This chapter has detailed the design and development of a multi-wavelength optical sensor for non-invasive hydration monitoring, focusing on the selection of optimal wavelengths, hardware configuration, software design, and baseline calibration. The sensor was carefully designed to measure hydration levels across different layers of the skin, leveraging NIR wavelengths that are particularly sensitive to water content. Through an extensive literature review, the wavelengths 975 nm, 1300 nm, and 1450 nm were selected for their distinct absorption peaks in water, while 1050 nm was chosen as a reference due to its minimal interaction with water molecules. These wavelengths, with varying penetration depths, allow for comprehensive hydration assessment from the epidermis to deeper tissue layers.

The hardware design centred on integrating these LEDs with an InGaAs photodiode on to a small PCB. The positioning of the LEDs relative to the photodiode was optimized to ensure accurate detection of light reflected from different skin depths, and the transimpedance amplifier within the ZenPPG system was used to convert the photodiode's current output into a voltage signal. This voltage signal was then processed using a DAQ system and analysed through LabVIEW software, which played a critical role in data acquisition, signal processing, and real-time monitoring.

The software design included the use of LabVIEW's DAQ Assistant block for signal acquisition, along with various processing blocks to isolate and analyse the voltage outputs from the LEDs.

In addition to the hardware and software, a baseline threshold calibration code was developed to establish a reference hydration level, allowing for accurate real-time monitoring and alert generation when hydration deviates from the baseline. This calibration process is crucial for personalizing the hydration monitoring system to individual users, enhancing its accuracy and utility in both clinical and personal health applications.

CHAPTER 8:

VALIDATION OF SENSOR DESIGN VIA IN SILICO AND IN VITRO EXPERIMENTATION

8.1 Monte Carlo Simulation to Validate Wavelengths of Developed Near-Infrared Sensor

8.1.1 Methodology

A Monte Carlo (MC) model is a well-established computational tool used to simulate light transport in turbid media based on probabilistic photon propagation. In this study, an MC model of human skin was developed to emulate the interaction of near-infrared light with multi-layered tissue, aiming to replicate the photon pathway as captured by the custom optical measurement device. The outcomes of this simulation were used to validate experimental data obtained from in vitro measurements on porcine skin, which shares comparable optical properties with human skin and serves as a suitable model for hydration studies.

The simulation utilised a pseudo-random number generator to model the stochastic behaviour of individual photons, assigning each photon an initial weight of 1 and tracking its scattering and absorption events through stratified skin layers. The source-detector configuration used in the simulation mirrored the physical setup of the optical device, with a source-detector separation of 5 mm. This fixed lateral distance was incorporated into the model by placing the virtual photon source at the origin and defining a circular detector area 5 mm away along the tissue surface, corresponding to the photodiode location in the experimental setup.

The numerical aperture (NA) of the system, which defines the angular acceptance of the detector, was also accounted for. Only photons scattered back toward the tissue surface and falling within the angular acceptance range defined by the NA were considered as detected. This angular constraint was simulated by computing the direction cosines (μx, μy, μz) of each exiting photon and evaluating whether the exit angle fell within the acceptance cone determined by the NA. If the angle exceeded this threshold, the photon was excluded from the detected reflectance signal. This ensured that the simulation output accurately reflected the directional sensitivity of the experimental system.

Each tissue layer in the model was assigned distinct optical properties—absorption coefficient (μa), scattering coefficient (μs), anisotropy factor (g), and refractive index—based on values from literature. These were wavelength-dependent and adjusted to reflect different water concentrations. To replicate hydration changes, the absorption

coefficient μa was dynamically varied by changing the volume fraction of water (vW) in the dermal layers. This approach allowed simulation of photon behaviour under both dry and hydrated tissue states.

The human finger is made up of skin, fat, muscle, and bone in reverse order. The skin is made up of six layers, namely the stratum corneum, epidermis, papillary dermis, upper blood net dermis, reticular dermis, and deep blood net dermis. The following MC model was adapted to imitate the light-tissue interactions seen in the porcine skin as the water concentration changes. Firstly, it considered only the behaviour of photons in the skin and fat layers, since the porcine sample used in the in vitro setup was made out of these two layers. Secondly, the model did not account for perfused skin and hence neglected the effects of blood and level of oxygenation. As a result, the absorption coefficients of all the skin layers were equal and denoted by '$\mu_a skin$'. A semi-infinite 3D model, with coordinates (x, y, z) and direction cosines (u_x, u_y, u_z), was developed to represent the spherical nature of photons. The total thickness (magnitude of the z-axis) of the region under investigation was 1.45 mm; the optical properties, which were obtained from the literature, were used to simulate and predict the light-tissue interactions at the source detector separation of 5 mm; the water concentration within the tissue was varied between 0% and 100% in 10% intervals. The simulation was run for 10^7 photons, with each photon assigned a "weight", i.e., intensity, equal to unity before it entered the tissue. Reflectance was quantified by calculating the sum of detected photon intensity, given by the weight of the photons which decreased over time due to surface reflection and fractional absorption at each interaction site. The Monte Carlo (MC) model determines 0% and 100% water concentration by modifying the absorption coefficient (μa) of the skin layers. Specifically, the variable vW represents the water content fraction, which directly influences the absorption coefficients of the skin layers.

As previously noted, the model involved the inclusion of variable inputs for the optical properties of the region of interest at the set wavelengths. These selected values can be seen in Table 8.1 for the scattering coefficients (μs) measured in mm-1, conveying the ability of the light photons to be absorbed and scattered per unit distance, respectively. The anisotropy factor (g) was also included to account for the directionality of the photons when they were scattered. The absorption coefficients (μa) of skin were calculated from the MC model for each wavelength at the different water concentrations[46], [122], [123], [124], [125].

Table 8.1: Optical properties of skin layers at selected wavelengths used in the MC model from the literature.

		975 nm			1300 nm			1450 nm	
	μ_a	μ_s	g	μ_a	μ_s	g	μ_a	μ_s	g
Skin	–	17.5	0.904	–	15.15	0.903	–	11.3	0.873
Fat	1.06	8	0.8625	0.89	8.5	0.9275	1.05	10.5	0.93
Water	0.0514	0	0	0.134	0	0	3.17	0	0

The flowchart below outlines the sequential steps of the Monte Carlo (MC) simulation used to model photon interactions within biological tissue and evaluate hydration effects. The process begins with parameter initialization, where key optical properties such as absorption (μa) and scattering (μs) coefficients, anisotropy factor (g), and refractive indices are defined. Once initialized, photons are emitted from a Gaussian beam source, assigned an initial weight, and allowed to propagate stochastically through the tissue. Their paths are governed by random scattering and absorption events, with energy loss modelled at each interaction site.

As photons travel, boundary conditions are applied to account for reflection and refraction at tissue interfaces. Photons that reach the detector at a fixed separation of 5 mm are recorded, and the reflectance is computed by summing the intensity of detected photons. The simulation was repeated across varying water concentrations (0%–100%) to analyse how hydration affects light propagation. The final results were then compared with experimental reflectance data from porcine skin to validate the model's accuracy.

This structured approach ensures that light behaviour in tissue is accurately modelled at the selected wavelengths. By quantifying changes in reflectance due to hydration, the MC simulation supports the validation of the hydration monitoring system, ensuring reliable depth-specific assessments.

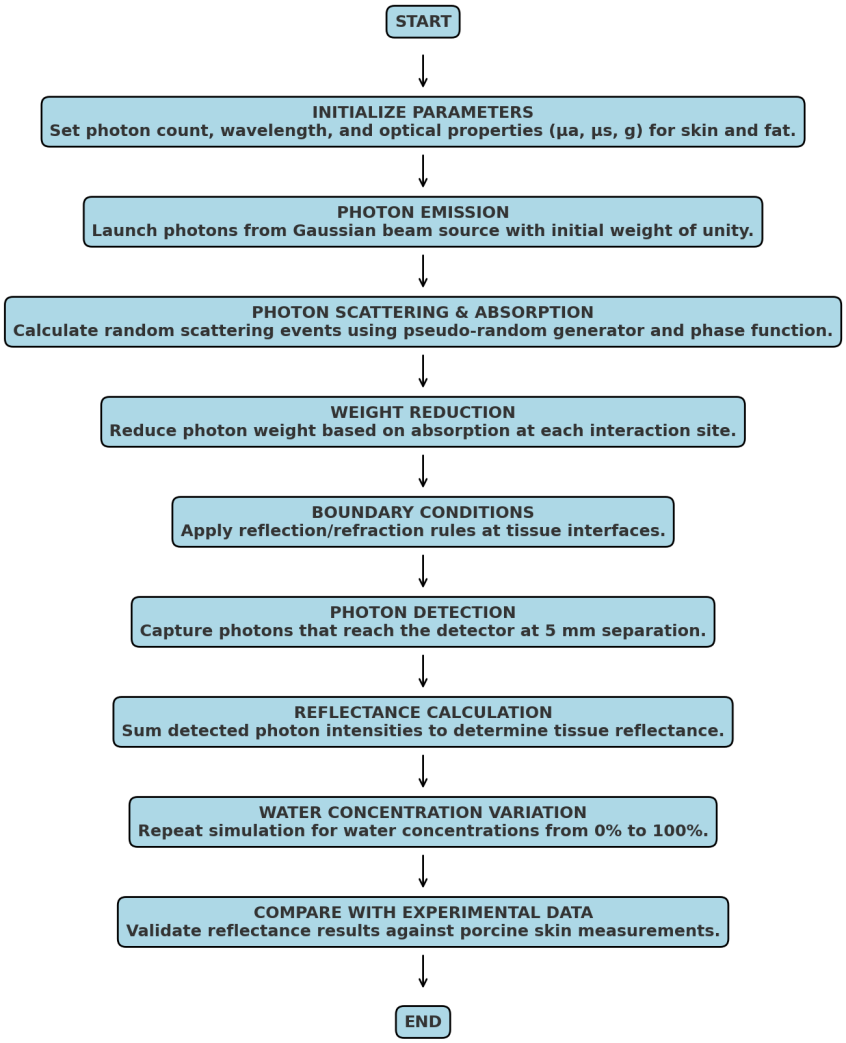


Figure 8.1: Flowchart showing the process of the Monte Carlo simulation and calculation of reflectance via water concentration variation

In addition, preliminary ex vivo experiments were conducted on porcine skin. This is due to porcine skin being the most similar in terms of optical properties to human skin and thus the most accurate to use for skin hydration studies. The experiment is described and analysed later in Chapter 9.

8.1.2 Results

The MC model generated reflectance outputs for all water concentration conditions at three wavelengths (975 nm, 1300 nm, and 1450 nm). Figure 8.2 shows the average reflectance across 0% and 100% water concentrations in increments of 10% at the three selected wavelengths. As expected, the reflectance is seen to increase as the water concentration decreases for all wavelengths, imitating the loss of water due to evaporation from the skin. This is illustrated in Figure 8.2, which plots the average reflectance against decreasing water concentration intervals, showing a positive trend. At the wavelength of 1450 nm, it is known that there is an extremely distinct water band present, thus explaining the increased steepness in its plots at higher water concentrations in comparison to other wavelengths. Consequently, the average reflectance is also plotted using a logarithmic scale for the better data visualisation.

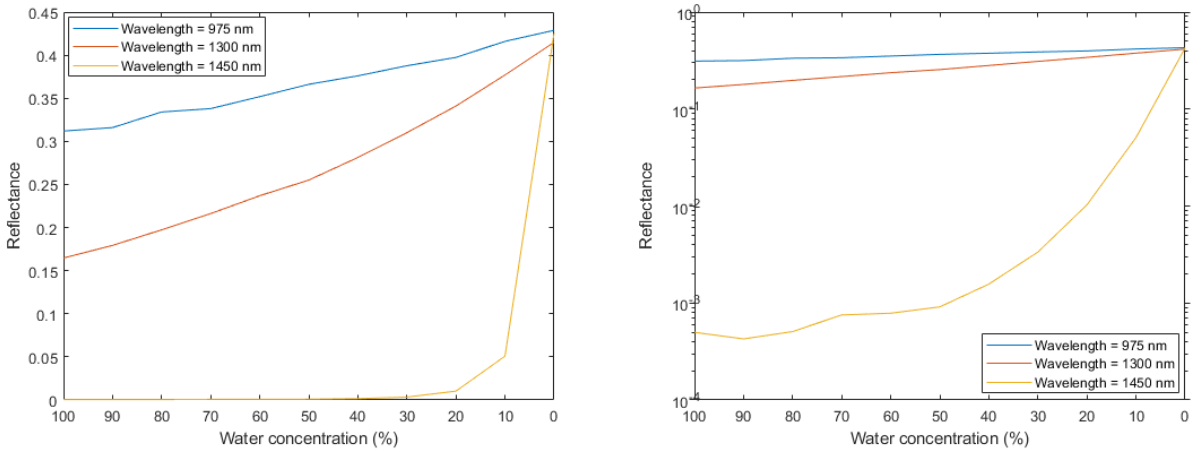


Figure 8.1: Average reflectance output as the water concentration changes at 975 nm, 1300 nm, and 1450 nm: (left) linear scale and (right) logarithmic scale.

In addition, photon profiles were generated from the MC model to illustrate the movement of detected photons between the source and photodetector via backscattering. These profiles, shown in Figure 8.3, compare 0% and 100% water concentrations to highlight optical differences. Water concentration in the model is controlled by adjusting the absorption coefficient (μa), where vW represents the fraction of water present. At 0%, absorption is dominated by baseline tissue properties, while at 100%, water absorption characteristics are maximized, modelling the optical effects of increased hydration rather than measuring pure water.

The profiles show clear differences, with higher water content leading to an increase in detected photons. Variations in scattering events are also evident across wavelengths, particularly at 1450 nm, where stronger water absorption reduces the number of photons reaching the detector, resulting in lower intensity in the profile maps.

Due to the significantly higher absorption coefficient of water at the 1450 nm (3.17 mm–1, Table 8.2), there is an extremely small number of detected photons at the 100% water concentration (Figure 8.3).

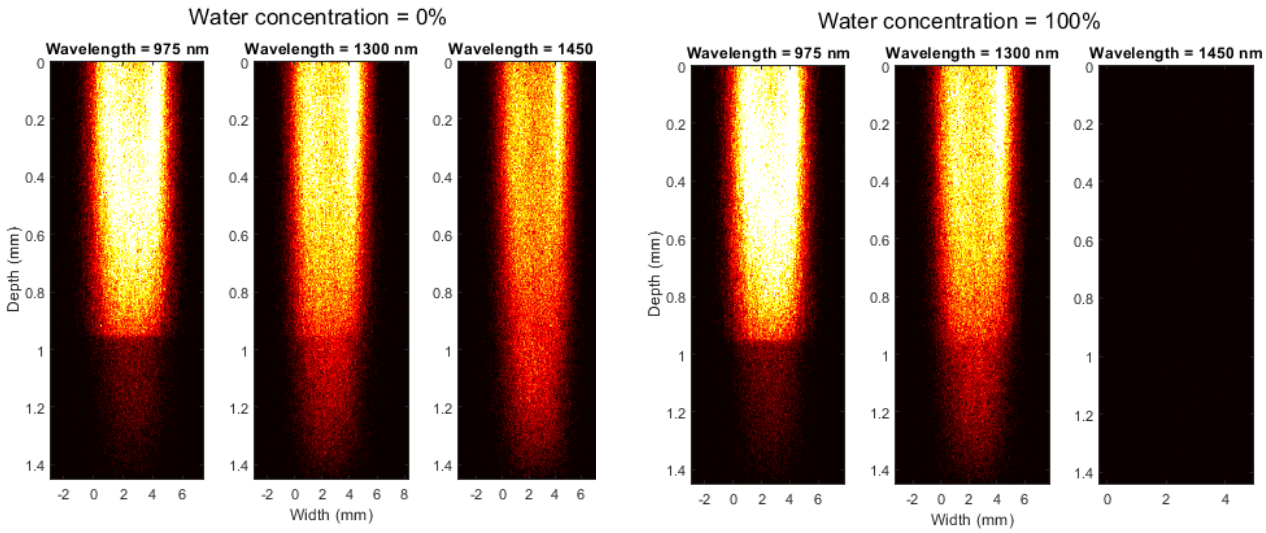


Figure 8.2: Simulated photon profiles for skin at a water concentration of 0% (left) and 100% (right) at 3 selected wavelengths

Table 8.2: Calculated SC absorption coefficients (mm-1) of water concentrations at 975 nm, 1300 nm, and 1450 nm in 10% intervals.

Water Volume (%)	**Wavelength (nm)**		
	975	**1300**	**1450**
0	0.0146	0.0057	0.0040
10	0.0183	0.0186	0.3206
20	0.0220	0.0314	0.6372
30	0.0257	0.0442	0.9538
40	0.0293	0.0570	1.2704
50	0.0330	0.0699	1.5870
60	0.0367	0.0827	1.9036
70	0.0404	0.0955	2.2202
80	0.0440	0.1083	2.5368
90	0.0477	0.1212	2.8534
100	0.0514	0.1340	3.1700

8.1.3 Summary

The need for skin hydration measurement devices based on optical techniques has been shown to offer notable advantages. The acquisition of optical properties within the skin can provide stronger correlations to water content due to the distinct water absorption bands present. These bands are more prominent when using wavelengths in the NIR region, which also allows for deeper penetration into the tissue. A wearable or portable skin hydration measurement device holds potential to enhance the monitoring of individual wellbeing, whether for general health, athletic performance, or dermatological care requiring moisturisation assessments [84], [91], [92].

To evaluate the suitability of the developed sensor design, a MC simulation was conducted, in which the exact source-detector separation and selected wavelengths were defined. The model successfully produced reflectance outputs as expected, along with photon profile visualisations showing the propagation path of detected photons. Within the skin layer—less than 1 mm in thickness—a significantly higher number of scattering events was observed compared to the underlying fat layer. This pattern was evident across all wavelengths, though the 1450 nm wavelength exhibited reduced penetration depth due to its higher water absorption coefficient. Consequently, fewer photons reached the detector, having been absorbed within the superficial skin layers.

Moreover, a higher number of photons were detected under 0% water concentration compared to 100%, particularly at 1450 nm, aligning with the expected absorption behaviour at this wavelength. This indicates that increasing water content in the tissue results in stronger absorption and a corresponding decrease in the detected reflectance signal. These findings complement trends observed in experimental testing and demonstrate that the simulation output is in agreement with theoretical expectations.

The MC simulation findings serve to support the design choices and provide theoretical confirmation of the sensor's sensitivity to hydration changes. The model outputs confirm the coherence between design parameters (e.g., wavelength selection and geometry) and the optical interaction behaviours expected in hydrated biological tissues. These results offer preliminary evidence that the sensor is capable of detecting changes in hydration levels, particularly in superficial layers, though further experimental repetition and comparison with gold standard techniques are necessary to establish definitive claims about measurement accuracy or precision.

CHAPTER 9:

VALIDATION OF DEVELOPED OPTICAL DEVICE VIA EX VIVO EXPERIMENTATION

9.1 Ex Vivo Desorption Test in Porcine Skin by Near Infrared Spectroscopy

9.1.1 Methodology and Materials

In vitro experiments were conducted on porcine skin. This is due to porcine skin being the most similar in terms of optical properties to human skin and thus the most accurate to use for skin hydration studies. The experiment took place in a dark room to eliminate the effect of external light from the environment being detected by the optical sensors during the measurements. The protocol for the in vitro experiment first involved collecting fresh porcine skin from the local butchers at Smithfield market that had been slaughtered on the day of collection. The porcine skin was cut into a 50 x 60 mm sample and was cleaned to remove any excess fat and connective tissue. The thickness of the porcine skin obtained ranged between 0.5 and 5 mm, with the sample being 3 mm thick.

The sample was placed on a petri-dish and into a humidity chamber set at 90% relative humidity (RH) and 25 degrees Celsius for 24 hours to ensure for maximum hydration of the skin sample. The skin sample was placed on an analytical balance with the developed optical sensor positioned on top of the right side of the sample and the reflectance optical fibre suspended over the left side of the sample, ensuring dermal markings were avoided. This set up allows for simultaneous measurements of gravimetric readings as well as reflectance spectra and time series recordings. The total weight of the sample was then recorded with the analytical balance, with its initial wet weight being 33.15 grams, and repeated every 20 minutes. Reflectance measurements from the developed optical sensor was continuously measured whereas the spectral data from the spectrophotometer was recorded every 20 minutes. The experiment continued with these measurements for a duration of 3 hours. The output data was extracted from Oceanview and LabVIEW for pre-processing on Spectragryph software and further comparison and analysis on Matlab. The full set up is shown and labelled in Figure 9.1 below.

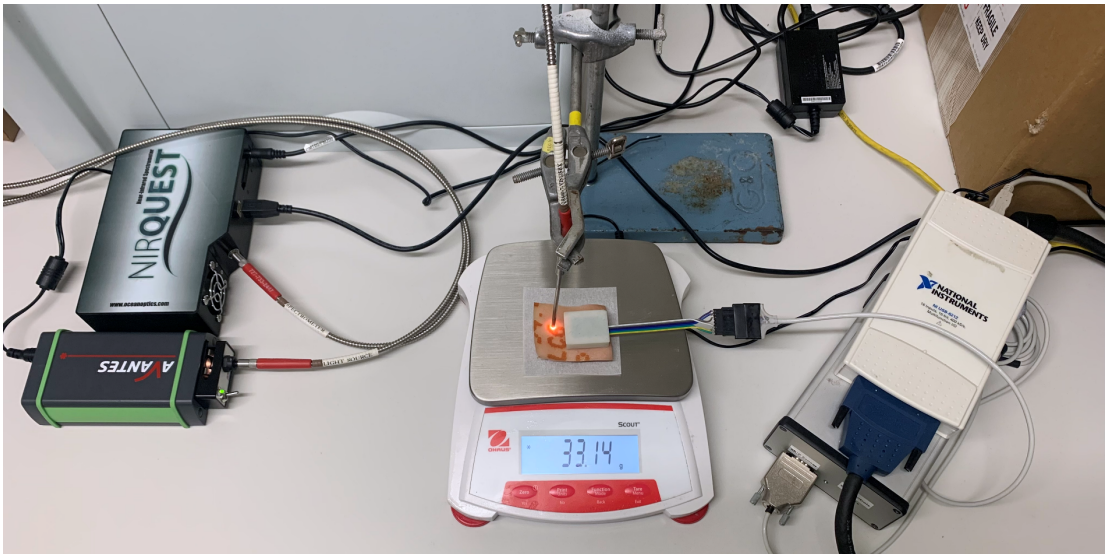


Figure 9.1: Full set up for porcine skin ex vivo experiment, including analytical balance, spectrophotometer and developed optical sensor

9.1.2 Pre-processing of Spectral Data

Pre-processing of the spectral data extracted from the spectrophotometer was performed using Spectragryph software. This involved converting the reflectance measurements to apparent absorbance measurements, using the formula: A = -log(1/R), where A is the apparent absorbance and R is the reflectance. This essentially flips the spectra across the x-axis from 900-1700 nm. The signal is then cleaned using smoothing, specifically using Savitzky-Golay filtering with a set window of 21. The baseline offset is then corrected for to allow for all the readings to be normalised by removing the initial baseline differences.

For the developed optical sensor, a time series was extracted measuring the changes in reflectance values over the duration of the experiment. This data was extracted and imported directly into Matlab for pre-processing. This involved smoothing of the signals at the four different set wavelengths to remove the noise detected by the sensor during readings. As 1050nm was used as the reference wavelength, this signal was subtracted from the signals at the other 3 wavelengths to give the final output spectra for comparisons and further analysis.

9.1.3 Results

The results of this experiment involved a reference of gravimetric measurements indicating the loss of the skin water content over time due to evaporation, via a desorption test, to then compare with the measured spectra. With a decrease in weight measurements of the skin sample, there was seen to be an increase in the reflectance measurement from both sensors. This confirmed that as the porcine skin was becoming more dehydrated, the absorbance was also declining, and thus there was an increase in the reflectance readings due to its inverse relationship to absorption.

The table below presents the gravimetric readings of the porcine skin sample, recorded every 20 minutes over the 3-hour experimental period. A consistent negative linear trend is observed, indicating a progressive reduction in the sample's weight over time. This trend is attributed to the evaporation of water from the skin, known as transepidermal water loss. To ensure the reliability of these measurements and to rule out artefacts such as balance drift or atmospheric interference, the desorption experiment was repeated multiple times under the same conditions. Furthermore, to assess the influence of environmental factors—particularly given that the primary apparatus was not enclosed—comparative tests were conducted using a more sensitive, enclosed analytical balance system. These control trials confirmed that the observed weight loss was indeed due to moisture evaporation from the tissue, and not due to fluctuations in measurement caused by environmental instability or balance calibration drift.

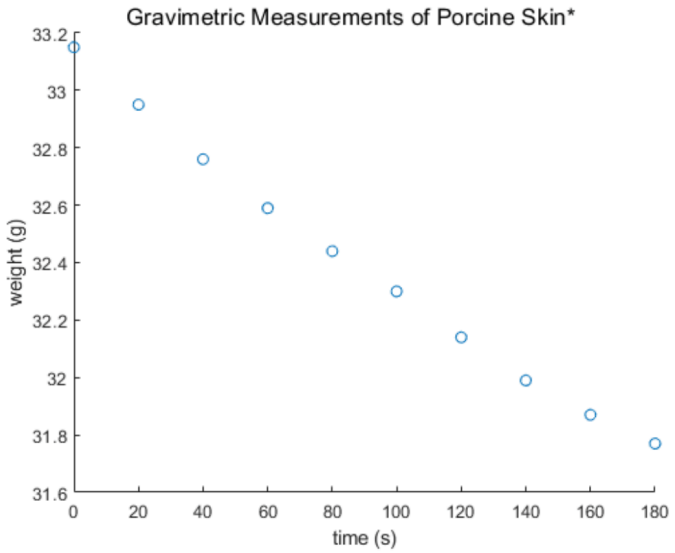


Figure 9.2: Gravimetric measurements of porcine skin ex vivo experiment, including analytical balance, spectrophotometer and developed optical sensor

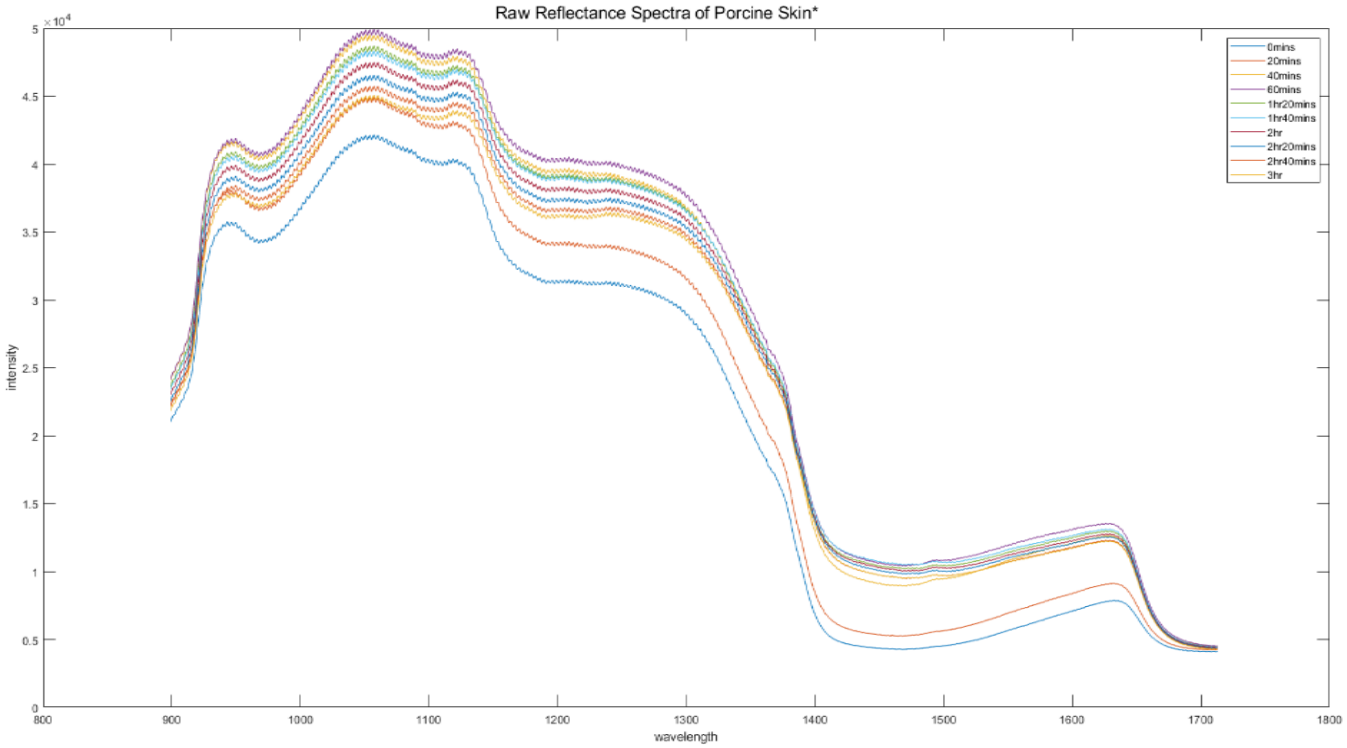


Figure 9.2: Raw reflectance spectra of the porcine skin sample at 20-minute time intervals

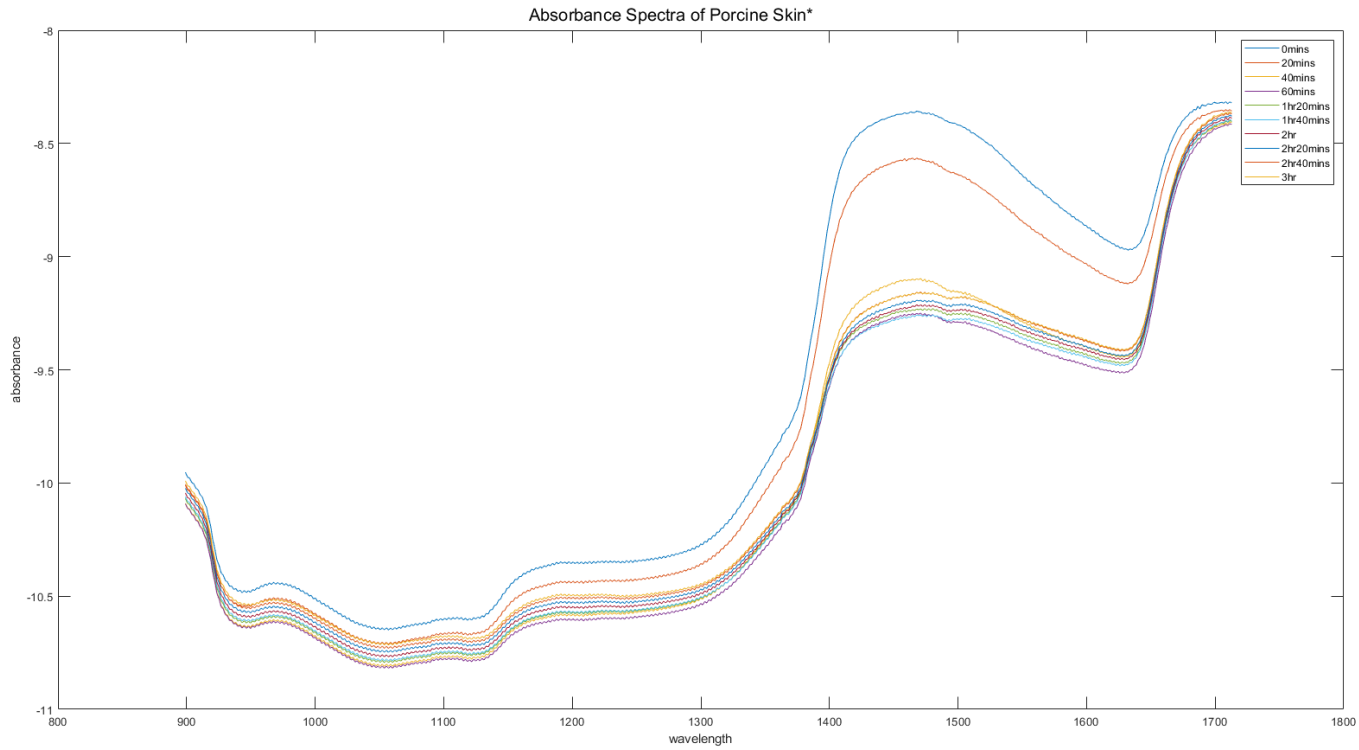


Figure 9.4: Converted absorbance spectra of the porcine skin sample at 20-minute time intervals

The raw reflectance spectra as displayed on the Oceanview software interface were extracted as data points for each 20-minute reading. The graphs plotted on Matlab are illustrated in Figure 9.3 above.

This raw reflectance (R) spectra were also converted to absorption spectra via the application of Beer-Lambert's law, as seen in the figure below. The formula states that the apparent absorbance (A) can be calculated using A = -1/log(R). Moreover, as mentioned in section 8.1.2, further pre-processing of the spectral data was performed that involved smoothing of the signal using Savitzky-Golay filtering followed by the removal of baseline differences. The spectra are illustrated in Figure 9.4 above.

The reflectance spectra obtained from the spectrophotometer in this in vitro experiment was converted to absorbance spectra in order to analyse the peaks present. The absorbance spectra above show the most distinct peaks at 975 nm and 1450 nm. These peaks are due to the water bands that are apparent at these wavelengths. Chemical analysis of the peaks suggests that at these particular wavelengths, there is a sufficient amount of energy provided by the LED that allows for the excitation of certain chemical bonds, for example the OH bonds present in water. This high energy level breaks these bonds and produces the output that is seen as an increased peak in the absorption spectra.

On the other hand, the reflectance spectra have an inverse relationship to the absorption spectra. Therefore, it is presenting the amount of light that has not been absorbed by the sample being measured but reflected back to the detector. These reflectance values in particular will be used further for comparisons to the developed optical sensor as it utilises the same optical methodology, yet as a time series of discrete wavelengths that are known to be sensitive to producing water bands in spectra.

The output from the developed optical sensor gives reflectance data points over time for each of the 4 independent wavelengths. Therefore, 4 time series across the 3-hour period are extracted and plotted on Matlab. As 1050 nm is the wavelength with the least distinct water peaks present, it is used as the baseline reference wavelength. The signal produced by the 1050 nm LED is subtracted from the signals of the other 3 wavelengths. This avoids for the application of further post-processing techniques by correcting shifts from external effects and allows for better graphical visualisation.

1050 nm Reference Subtracted – Optical Sensor Timeseries on Porcine Skin

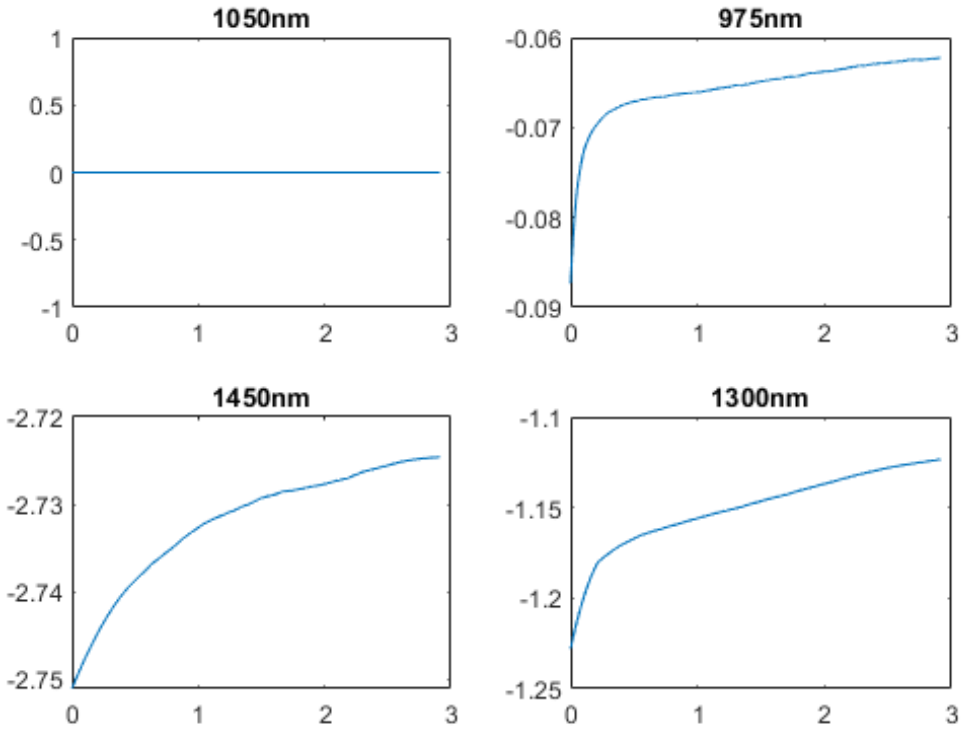


Figure 9.6: Time series of developed optical sensor on porcine skin for the 4 LED wavelengths over time after the subtraction of the 1050nm reference wavelength

Comparison of Timeseries of Reflectance Spectra of Porcine Skin – 975 nm

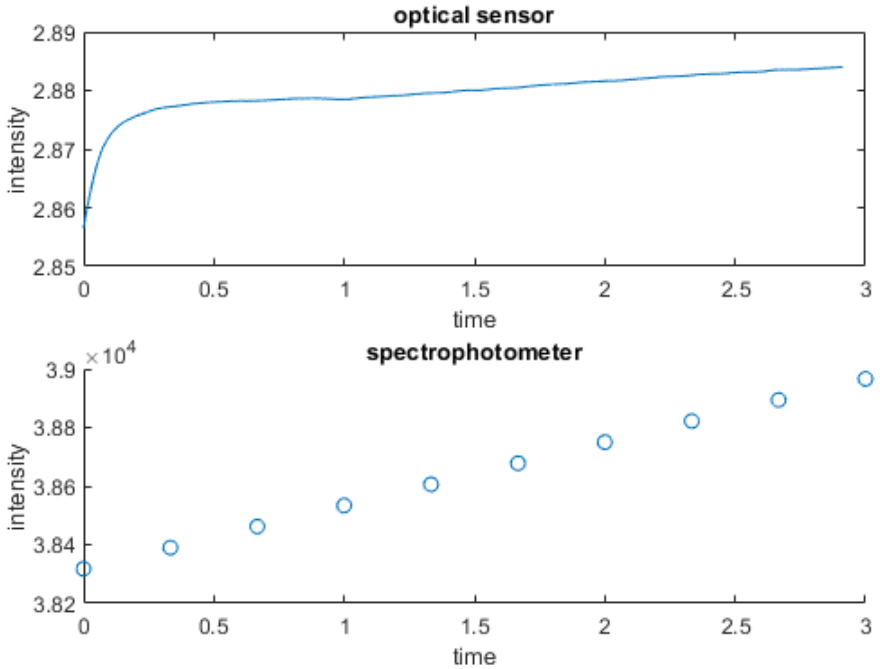


Figure 9.6 : Comparison of reflectance spectra time series of developed optical sensor at 975 nm and the spectrophotometer on porcine skin over time

Comparison of Timeseries of Reflectance Spectra of Porcine Skin – 1300 nm

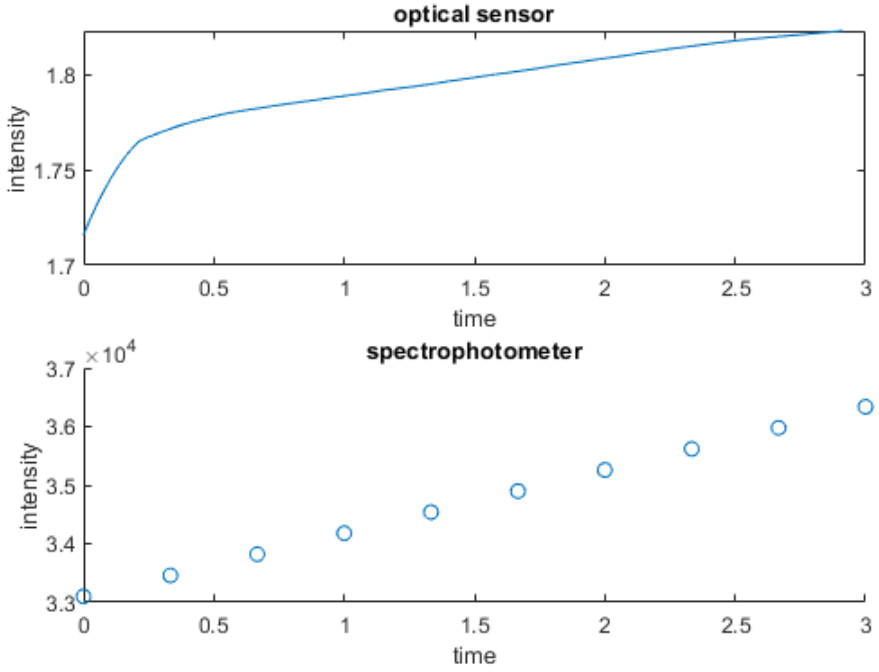


Figure 9.7: Comparison of reflectance spectra time series of developed optical sensor at 1300 nm and the spectrophotometer on porcine skin over time

Comparison of Timeseries of Reflectance Spectra of Porcine Skin – 1450 nm

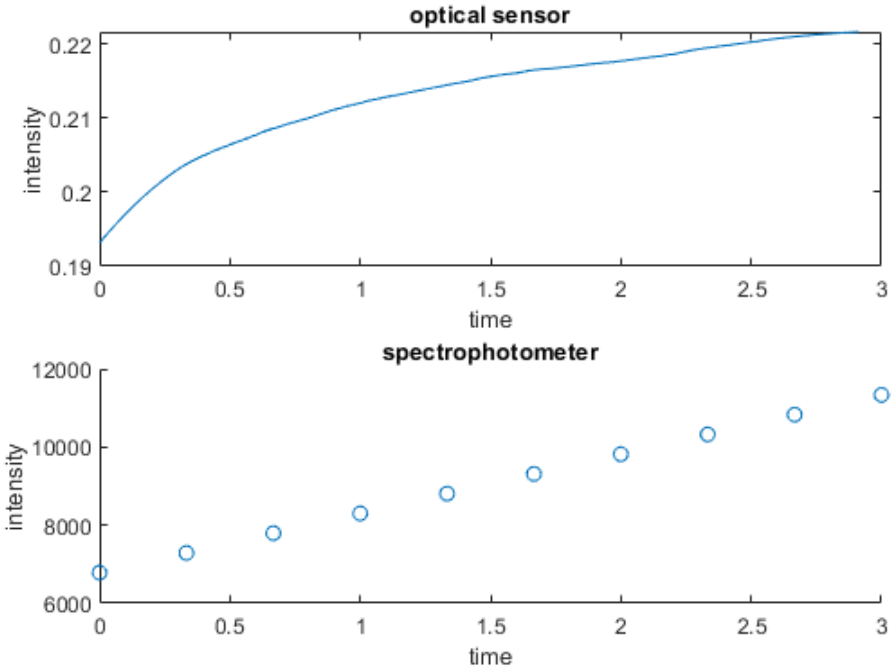


Figure 9.8: Comparison of reflectance spectra time series of developed optical sensor at 1450 nm and the spectrophotometer on porcine skin over time

The output reflectance spectra obtained from the spectrophotometer are converted to time series plots by extracted the discrete reflectance values at the different

time points for each of the 3 wavelengths (975 nm, 1300 nm and 1450 nm). This is shown on the bottom plots of each of the figures above (Figures 9.6, 9.7 and 9.8) as 10 graphical time points. This spectrophotometer reflectance time series is used as a reference to compare to the reflectance time series of the developed optical sensor. It is conveyed from the graphs that there is a similar trend presenting an increase in reflectance values over time due to evaporation of water from the sample. This supports suggestions that as water content decreases, absorption decreases and thus reflectance increases. Moreover, this confirms that the developed sensor works similarly to the spectrophotometer.

When comparing the gravimetric weight measurements of the porcine skin sample over time, its linear relationship can be portrayed to complement the trend of both optical techniques. Both the developed sensor and the spectrophotometer displayed a positive trend in their time series, suggesting an increase in absorbance and relatively a decrease in reflectance due to the inverse relationship between these spectral measures. Furthermore, when solely comparing the spectral time series of the 2 sensors a positive correlation can be visualised, which confirms the similarities of the readings obtained from the developed sensor to that of a gold standard spectrophotometer and increases its validity as a measure.

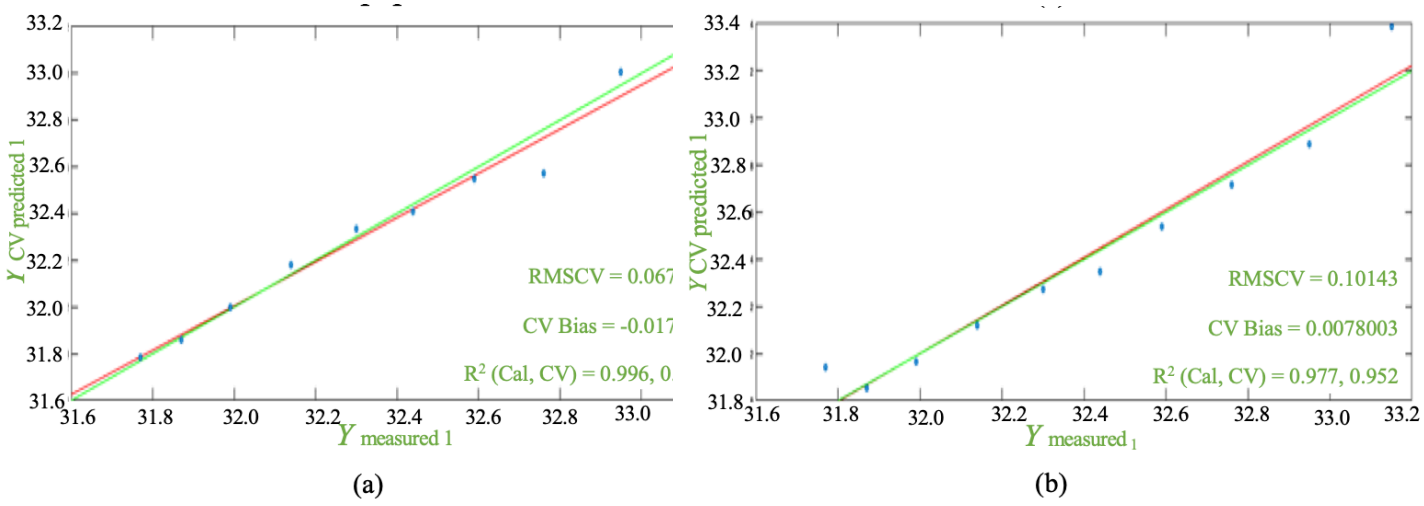


Figure 9.9: left) Output scores extracted from regression techniques: (a) partial least squares (PLS) regression output of spectrophotometer (red) and gravimetric (green) measurements. right) Output scores extracted from regression techniques: (b) multiple linear regression (MLR) output correlation plot of the developed optical sensor (red) and gravimetric (green) measurements.

9.1.4 Analysis of Results

In order to evaluate the correlations between the measured results and the reference, regression techniques are required. For comparisons of the independent

wavelengths chosen within the developed optical sensor to the gravimetric readings, the calculation of the Pearson correlation coefficient is established. This correlation coefficient, R, measures the linear relationship between 2 variables and the strength of this association. The p-value is used to convey the statistical significance of the correlation if the obtained R values conclude a difference from 0 in population. In this experiment, it was expected that there would be a negative correlation present, since a decrease in the weight and water content of the porcine skin sample related to an increase in the measured reflectance readings. Table 9.1 displays the calculated R and p-values at these individual LED wavelengths. It can be seen in the table that 975 nm has a weaker correlation, which could also be seen by it having the highest p-value out of the 3 wavelengths. However, all R values have presented the expected strong negative correlation to the gravimetric reference. This means that there is a high tendency for X variable scores, being the device measurements, to go with low Y variable scores, being the gravimetric readings (and vice versa). Furthermore, these significantly strong correlation coefficients seen for the developed sensor, signifies the high measurement accuracy and reliability. Moreover, the strong negative correlations alongside extremely low p-values is more prominent at the wavelengths of 1300 nm and 1450 nm, signifying the presence of more distinct water bands at these wavelengths and the increased sensitivity of the bands with the developed sensor, which has been proved[64].

Table 9.1: Pearson correlation coefficients of sensor's independent wavelengths vs gravimetric measurements.

	R	*p*-value
975 nm LED vs gravimetric measurements	−0.848 6	0.001 936
1300 nm LED vs gravimetric measurements	−0.989 9	<.0.000 01
1450 nm LED vs gravimetric measurements	−0.928 5	0.000 105

The PLS regression was also used to model and analyse the covariance between the entire spectral output of the spectrophotometer and the reference gravimetric measurements. The PLS has the ability to quantify the relationship between 2 sets of data by constructing linear predictive models. The PLS regression is useful to use in this case due to the predictors being collinear and the spectral measurements containing a large number of variables. Unlike principal components regression (PCR), the PLS takes into

account variance and measures the predictors with errors, increasing the robustness and measurement accuracy of the model. Eigenvector's PLS_Toolbox (R9.2, Eigenvector Research, Washington, USA) was installed on to MATLAB, where the data were inputted into the PLS model, involving cross validation via Venetian blinds with 5 splits, in order to assess the predictive capability of the model. The output data extracted from this model were the calibration coefficient of determination (R2 Cal.), cross-validation coefficient of determination (R2 CV), bias, CV bias, RMSEC, and RMSECV, and graphical scores[126].

On the other hand, the MLR was the regression technique used for analysing the relationship of the developed optical sensor with the reference gravimetric measurements. The MLR correlates 2 datasets by calculating the distance to a fully linear relationship. This regression model was used as the data obtained from the developed data were single intensity points denoting the reflectance value at a specific point in time, as oppose to a spectral output requiring PLS. Three sets of these reflectance data recordings were obtained, as there were 3 LED wavelengths acting as the model's independent variables. The same outputs were extracted for the MLR as the PLS which allowed for comparative analysis purposes.

The results displayed in Table 9.2 conveyed a noticeably high R^2 for the cross-validation of 0.952 with the developed optical sensor, which was an extremely strong correlation coefficient to the reference variables. In comparison, the spectrophotometer had an R^2 value of 0.979, with a difference of only 0.027 to the developed sensor, enhancing its ability to perform similarly to a gold-standard device. Additionally, the likelihood of error affecting the results was low in both cases for the RMSECV, but relatively lower for the developed sensor in cross-validation alongside a significantly lower bias than the spectrophotometer. This enhanced the measurement accuracy of the developed sensor and its ability to outperform a standard spectrophotometer. The differences in bias and RMSECV can be explained by the fact that the developed sensor uses independent wavelengths that are most sensitive to water bands within the skin and therefore experience less interference from other skin constituents.

Although the regression values between the two devices were very close, it was expected that the spectrophotometer would perform slightly better, though the difference was small. However, the developed sensor, with three wavelengths specifically targeting water absorption bands, was shown to sufficiently detect variations in water content

almost equally to a full-range measurement device, suggesting additional wavelengths not directly related to water bands may not be necessary for hydration measurement.

When considering other applications, such as moisturizer assessments, the developed sensor's sensitivity to water-specific wavelengths would still be relevant. Moisturizers primarily aim to alter skin water content by either enhancing water retention or creating a barrier to reduce water loss. In this context, the ability of the developed sensor to detect changes in hydration levels should remain applicable, as the fundamental measurement is based on changes in skin water content. However, the dynamics of water absorption in the presence of moisturizers might introduce additional variables, such as the interaction between skin surface lipids and water, which may affect measurements differently than in experiments focused on TEWL. Further testing would be required to confirm the sensor's performance in such scenarios, but its demonstrated sensitivity to water content changes makes it a promising tool for moisturizer assessments as well.

In addition, the results displayed in Figure 9.9 above present the scores extracted from the regression models. The figure shows the plots for the measured and predicted outcomes against a directly correlated line. It can be seen on the right side, being the MLR scores for the developed sensor, the plotted points are extremely close to the expected line, signifying the strong predictive accuracy, which can also be similarly portrayed from the scores from the spectrophotometer output on the left side of the figure.

These scores allow for the correlation between the x and y score variables to be established as well as the strength of the linear agreement. This positive linear correlation consisting of very few outliers from the predicted correlation line indicates that as one variable changes, the mean of the other variable has almost the exact trend, which can be suggested in engineering that the variables can be considered from the same system.

Table 9.2: MLR and PLS regression outputs of the developed sensor and spectrophotometer to gravimetric measurements.

	R^2 Cal.	R^2 CV	Bias	CV bias	RMSEC	RMSECV
Reference spectrophotometer PLS	0.9963	0.9790	0.0001039	–0.01787	0.02679	0.06794
Developed optical sensor MLR	0.9773	0.9523	0.0000369	0.0078	0.06665	0.1014

9.1.5 Summary

An ex vivo experiment was conducted using a custom-built optical sensor to assess its potential for monitoring skin hydration levels. Porcine skin was selected due to its optical similarity to human skin, making it a suitable model for hydration studies. The experiment was performed in a controlled setting to minimise ambient light interference, and skin samples were hydrated in an environmental chamber before undergoing reflectance and gravimetric measurements.

The gravimetric reference data captured the progressive water loss due to evaporation, showing a clear negative correlation between sample weight and time. Reflectance data from the developed sensor and a standard spectrophotometer showed increased reflectance as dehydration progressed, consistent with the known inverse relationship between absorbance and reflectance.

A comparative analysis between the developed sensor and spectrophotometer showed a consistent positive correlation, with both systems capturing similar temporal trends in water loss. Enhanced signal fluctuation was observed at 1450 nm, consistent with the high water absorption at this wavelength, reinforcing the sensitivity of the sensor to hydration changes in superficial tissue.

To characterise the sensor's performance, regression models using multiple linear regression and partial least squares techniques were applied. Both devices achieved high R^2 values during cross-validation, with the spectrophotometer showing only a marginal increase. Notably, the developed sensor produced lower bias and error estimates, indicating promising agreement with the reference gravimetric measurements.

While these results demonstrate that the sensor performs consistently with established techniques, and shows clear responsiveness to hydration changes, further repeated experimentation under both controlled and variable conditions would be needed to assess long-term precision and robustness. These findings provide supporting evidence that the sensor is capable of detecting hydration-related optical changes in skin tissue and may be useful in non-invasive hydration tracking applications.

9.2 Multimodal Spectroscopic Assessment of Skin Hydration: Optical and Electrical Techniques

9.2.1 Background

The typical techniques used in measuring the hydration of the skin and its barrier function are electrical-based sensors, with the gold standard device being the corneometer. However, although such devices are widely used and considered accurate, they may exhibit a range of errors that can introduce significant measurement discrepancies which have not been adequately addressed in commercial devices. The major errors can be grouped into two main categories: (i) errors that relate to the electrode–skin contact properties, and (ii) errors due to the nature of the measurand, given that the corneometer does not directly measure water content[64], [127], [128], [129], [130], [131].

Errors in category (i) include variability as a function of contact area and geometry, electrode material, and applied pressure. These errors can be associated with the fact that such instruments feature a two-electrode (bipolar) configuration. The Corneometer's principle of operation is described by the wider theory of bioimpedance measurements. Bioelectrical impedance (or bioimpedance) sensing is applied in a broad range of biomedical applications, including body composition estimation, cardiac output measurements, cell culture monitoring, and monitoring of skin hydration, to name a few. The electrical properties of biological tissues are influenced by factors including their morphology, physiology, and pathological conditions[127], [129], [130].

The primary source of errors in bipolar bioimpedance systems, including typical corneometers, relates to the electrode–electrolyte double layer introducing a contact impedance that can significantly interfere with the measured value. As measurements are taken across a range of frequencies (typically up to 1 MHz), the detrimental effect of the contact impedance varies significantly, especially when small electrodes and/or low frequencies are employed[131], [132].

A commonly employed method for mitigating the impact of contact impedance on measurements involves utilizing four electrodes. Referred to as tetrapolar electrical impedance measurement (TEIM), this technique effectively reduces the adverse effects of the double layer on the measurement. The appropriateness of the tetrapolar approach for

measuring hydration has been demonstrated in the literature. It is, therefore, desirable to compare the effectiveness of tetrapolar versus bipolar hydration measurements to assess the significance of double layer errors in electrical sensing methods[133], [134], [135].

Bioimpedance hydration sensing relates to the skin's electrical properties, like permittivity and conductivity, whose combined values vary with different water content levels. As the measurements reflect water content only indirectly, they are expressed in arbitrary values. This reliance on electrical properties results in errors in category (ii), including decreased sensitivity at high hydration levels, susceptibility to environmental variations, and fluctuations with electrolyte concentrations. These errors can be minimised by sensing water content directly, which can be made possible through optical sensing based on the capability to directly detect water-related bands present within the spectral response[136].

As previously mentioned, there has recently been a surge in the research on optical sensing methods aimed at obtaining parameters for skin properties. With this methodology, the sensitivity of recording water presence is increased at and has been observed to display a positive correlation to absorbance levels. Furthermore, as optical methods are sensitive to OH and HOH bands, it can be argued that this is a more direct measurement of water content. However, this is not necessarily hydration, which is based on multiple aspects. This presents the need for techniques to focus on different aspects of skin to provide a more holistic approach[64], [97], [137], [138].

In this section, we will investigate both of the main sensing modalities that can potentially eliminate the aforementioned common corneometer errors (i) and (ii). The multimodal sensing approach, i.e., the use of both electrical and optical sensing modalities, can allow for increased sensitivity and validity in bodily measurements by providing a multi-layer approach to assessment. Combining both these techniques can pave the way for the next-generation skin hydration assessment devices[138].

This experimental section outlines the ex vivo and proof-of-concept in vivo testing of custom-designed optical and electrical-based sensing devices, assessing their precision and reliability individually and evaluating their combined impact. The ex vivo investigation involved employing the desorption of porcine skin over a 6 h duration at room temperature, with measurements taken at fixed intervals. Furthermore, an in vivo indicative study was conducted to ascertain the sensors' accuracy when applied to human skin by examining their raw outputs, considering various affecting parameters. Analysis

of both methodologies facilitated comprehensive insights into the efficacy and advantages of employing multi-modal approaches for skin hydration measurement. The comparative assessment of the individual techniques highlights their potential complementarity in enhancing the precision of skin hydration measurement.

9.2.2 Methodology and Materials

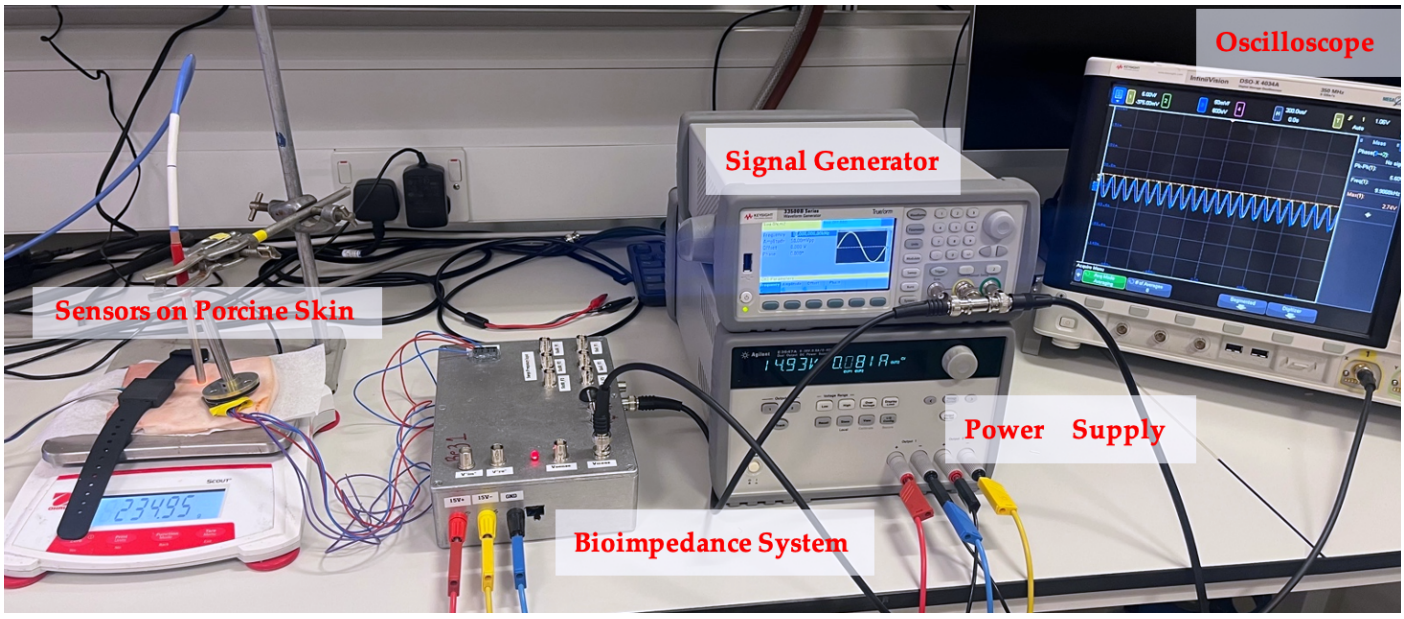


Figure 9.10: Ex vivo experimental setup for bioimpedance system.

A 6-hour ex vivo experiment was conducted using porcine skin to assess the measurement of water content from the developed optical devices during a desorption test. A sample of porcine skin was secured and cut to remove any extraneous adipose or connective tissue. The skin sample had a thickness of 3 mm, consisting of the skin layer and a thin layer of fat. In order to establish optimal hydration, the sample was placed in a humidity chamber set at a controlled relative humidity (RH) of 90%, with the temperature maintained at 25 °C over a duration of 12 hours. Porcine skin is a suitable model for testing skin hydration measurement devices due to its high similarity to human skin in terms of optical properties and tissue layer composition. It offers reproducibility and consistency, making it appropriate for use in controlled ex vivo experiments[139].

Once the sample had achieved its augmented hydration level, it was placed directly onto an analytical scale, where its initial gravimetric measurement was recorded.

Positioned on the left side of the sample was the developed optical sensor, connected to its external ZenPPG system and DAQ card (National Instruments, Austin, Texas, U.S.A.). In the centre of the sample, a reflectance fibre optic probe was suspended 1 mm from the skin surface and connected to a commercial benchtop spectrophotometer. On the right side of the sample was the bioimpedance sensing device, held in place with a small weight to prevent movement of the lightweight sensor.

In terms of measuring the impedance (Z), it can be represented by a simplified lumped impedance representation of the human skin, shown in Figure 9.11, where Re and Ce represent the epidermis and Rd represents the dermis and underlying subcutaneous tissues. From this perspective, porcine skin is a relatively accurate representation of human skin. Both bipolar and tetrapolar electrode configurations were used in the experimental procedure. The tetrapolar measurements were minimally unaffected by the presence of the two R1-C1 impedances, unlike the bipolar measurement. The latter was implemented by shorting the voltage measurement and current injection electrodes to form a single pair by switching S1 and S2 in their on-positions. For each setup of electrode configuration, an output constant current of 25 μA was generated and injected into the porcine skin[140].

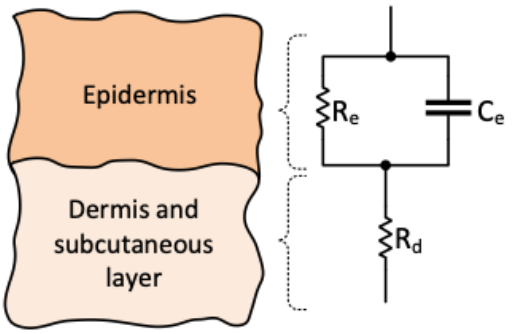


Figure 9.11: Expanded model of the tissue at the target skin site and its equivalent circuit model.

The initial mass after setup was recorded to be 242.35 g, where the initial readings from the 3 measurement devices were also taken. Although the developed optical sensor measurements were recorded in a continuous manner, the other 3 measurement techniques required recordings to be taken simultaneously in 20-minute intervals

throughout the 6 hour experimental timeline. The overall setup for this experiment can be seen as a flow chart, shown in Figure 9.12.

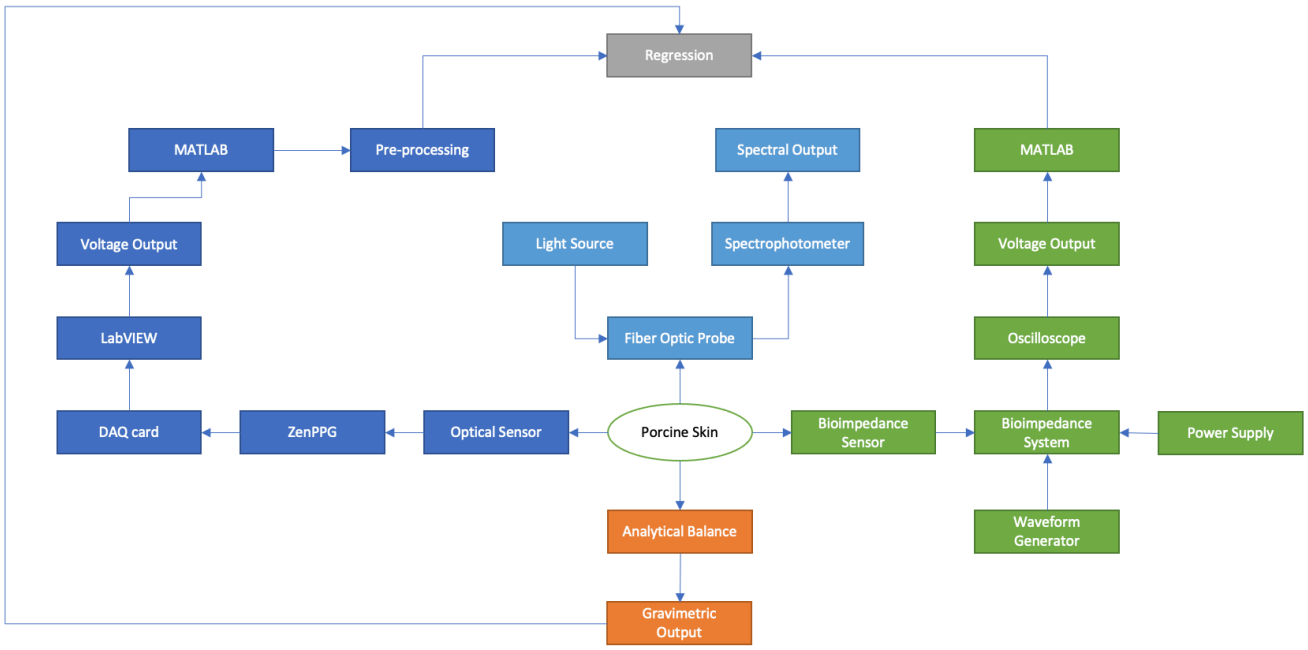


Figure 9.12: Flowing illustrating complete setup and process for ex vivo experiment illustrated as a flow chart (blue = optical, green = bioimpedance, orange = gravimetric measurements).

A small-scale in vivo indicative case study was also conducted to further validate the developed sensors and to confirm their accuracy when used on human skin. This case study is fundamental for several reasons. Firstly, the use of in vivo testing involving human participants provides a more practical applied assessment of the sensors' performance and validity. While ex vivo studies can offer valuable insights, the application in humans reflects their effectiveness when used in the real world, taking into account the complexities and distinctions of skin and inter-individual differences. These measurements were correlated against the gold-standard corneometer device.

The indicative case study involved mean averaged measurements taken from both forearms of a single participant under two conditions. The first condition involved a dry forearm, serving as a baseline reference. This condition represents the natural state of the skin, with no external applications or influences. It allows for the assessment of the sensors' ability to capture hydration of the skin in its typical state. The second condition involved a damp forearm with water applied using a soaked pad, emulating a common scenario when the skin is exposed to external elements or is indicated to be well hydrated in terms of internal water intake. This allowed for the sensors' capacity to accurately

respond to changes in hydration levels to be tested. Figure 9.13 below display the sensors' placement for in vivo measurements on human skin, with the bioimpedance sensor in place in a tetrapolar configuration.

Ethical approval for both porcine skin and human forearm measurement experiments was obtained from the City, University of London Research Ethics Committee. Informed consent was obtained from the human participant prior to commencing the study.

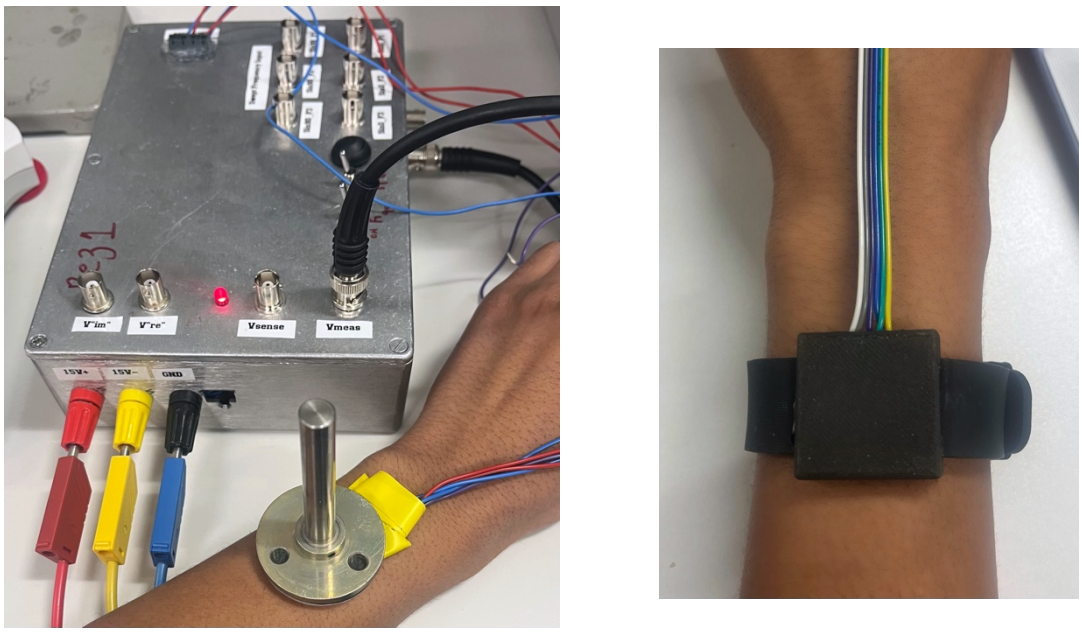


Figure 9.13: left) Bioimpedance sensor positioned on human skin with weight to hold electrodes in place. right) Optical sensing wearable positioned on wrist.

9.2.3 Results

Reference Gravimetric Measurements:

Gravimetric measurements are considered the gold standard for the measurement of water content. This method involves determining the change in the weight of a sample before and after the removal of water. The difference in weight is attributed to the water content. Gravimetric measurements were recorded every 20 minutes for a total experimental duration of 6 h. These results can be seen in Table 9.3. There was an extremely noticeable, linear downward trend in weight over time, from 242 g to 192 g, representing the expected decrease in water content as a result of desorption. As mentioned previously in Section 9.1.3., these gravimetric changes were en

Table 9.3: Recorded gravimetric measurements during desorption test of the ex vivo experiment.

Time (minutes)	0	20	40	60	80	100	120	140	160	180	200	220	240	330	360
Gravimetric (grams)	242	233	230	228	226	225	224	222	221	219	217	215	214	193	192

Spectral Experimental Results from Optical Sensor:

The output from the developed optical sensor generated DC voltage data points over time for each of the four independent wavelengths. Consequently, four distinct time series spanning the six-hour experimental duration were extracted from LabVIEW and visualized using Matlab. To establish a baseline reference wavelength, 1050 nm was chosen due to the minimal presence of distinct water peaks. Subsequently, the signal produced by the 1050 nm LED could be subtracted from the signals originating from the other three wavelengths in later analysis stages. This subtraction process not only negated the need for additional post-processing techniques to correct for external effects, but also enhanced graphical visualization for improved evaluation. The raw data and filtered outputs for the four wavelengths can be seen plotted in Figure 9.14.

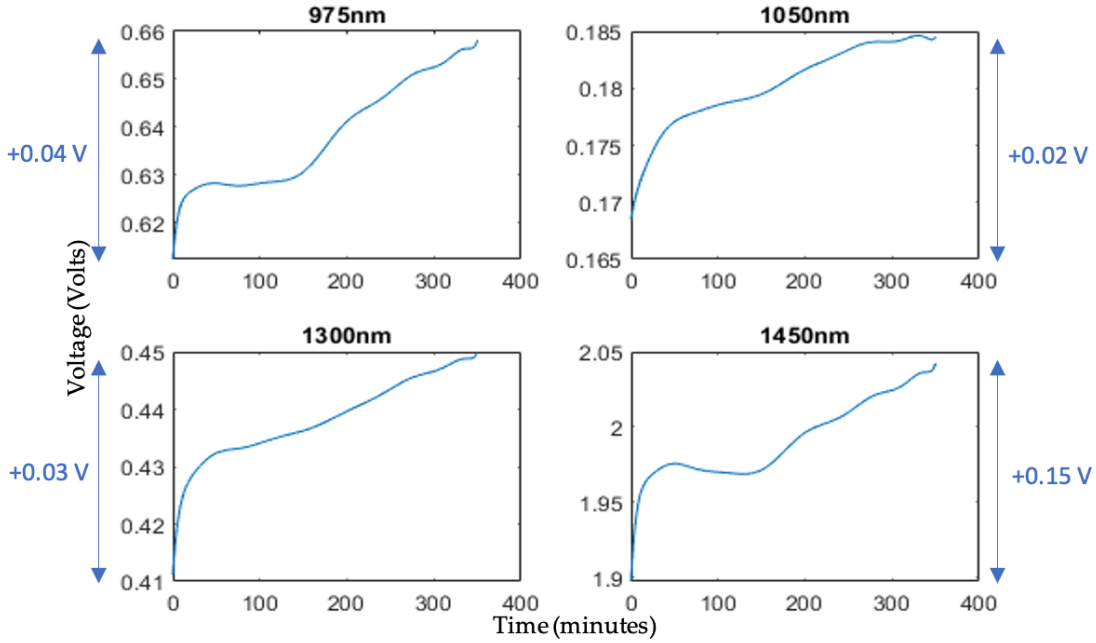


Figure 9.14: Raw voltage output from optical sensor over 6 h experimental time for selected LED wavelengths, with total incremental increases during ex vivo desorption test.

As anticipated, an upward-sloping correlation was evident in the voltage measurements over the total experimental duration. This was consistently observable across all four distinct wavelengths, with the least perceptible variation recorded at 1050 nm, indicating a 0.02 V increment, while the most significant alteration was at 1450 nm, reflecting a pronounced 0.15 V increase, and, thus, an increased rate of change. This empirical observation is consistent with the established literature, which suggests the presence of a distinct and significant water absorption peak at 1450 nm. This results in a heightened sensitivity to alterations in the water concentration within the skin. As the water concentration within the skin decreases, due to the desorption process, the reduced water content results in a diminished attraction for water molecules under incident light at 1450 nm. Consequently, this reduced water concentration exhibits an increase in reflectance values, thereby causing the notable 0.15 V rise which was observed.

The performance of the developed optical sensor was also compared against a conventional spectrophotometer interfaced with a reflectance fibre optic probe. The results provided validation for the reliability of the data obtained with the developed sensor, primarily attributed to the observed linear relationship between the two instruments. This parallelism was established through their concurrent use in a comparable desorption test also employing 975 nm, 1300 nm, and 1450 nm wavelengths. Consequently, this finding significantly enhanced the credibility of the developed sensor as a valid measurement instrument for characterizing alterations in water concentration. The spectrum in Figure 9.15 shows the changes in the water from the porcine skin as time progressed during the TEWL.

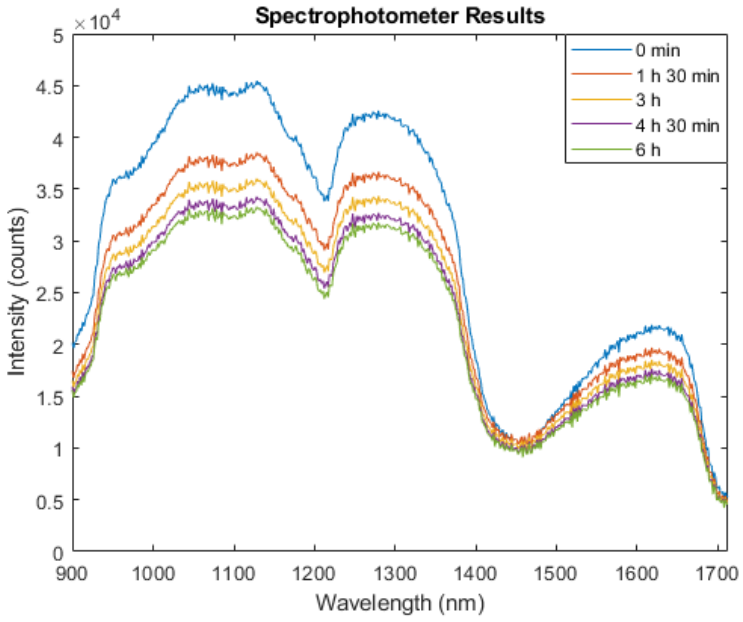


Figure 9.15: Spectrophotometer output at different timepoints during ex vivo experiment.

Experimental Results from Bioimpedance Sensor:

In the initial 120 minutes of the experiment, porcine skin bioimpedance measurements from bipolar and tetrapolar electrode configurations were taken, as shown in Figure 9.16, respectively, where data recorded at 20-minute intervals are indicated by the markers and a spline interpolant was used to generate the data trend traces. Measurements were carried out for frequencies of 1 kHz, 5 kHz, and 10 kHz. The selection of frequencies was influenced by the bandwidth constraints of the conductivity sensor and were lower than the typical maximum bandwidth of corneometers (about 0.9–1.2 MHz), but within range of the lower end of their reported bandwidth, making the chosen values relevant[141].

As anticipated, in all cases, impedance increased over time, reflecting transepidermal water loss from the surface of the porcine skin. The impedance values at 5 kHz and 10 kHz were lower, which was attributed to the decreased reactance of the tissue's reactive impedance component at higher frequencies. Notably, the bipolar configuration (denoted as B in the legend) exhibited higher impedance baseline values compared to the tetrapolar measurements (denoted as T), indicating the influence of electrode/electrolyte interface impedance. The impedance variation over the course of the experiments seemed more pronounced in the tetrapolar results.

The overall impedance variation in each case was better illustrated through the plotting of the percentage impedance magnitude increase, shown in Figure 9.17 for the bipolar and tetrapolar configurations, respectively. The percentage increase for each data set was calculated using the formula in the equation below:

$$\text{Percentage Increase} = \frac{\text{Final Value} - \text{Starting Value}}{\text{Starting Value}} \times 100 \qquad \textit{Equation (9.1)}$$

A percentage increase of >21% in impedance magnitude was observed in both cases of electrode configuration. However, the bipolar configuration exhibited a lower percentage increase in impedance at lower frequencies.

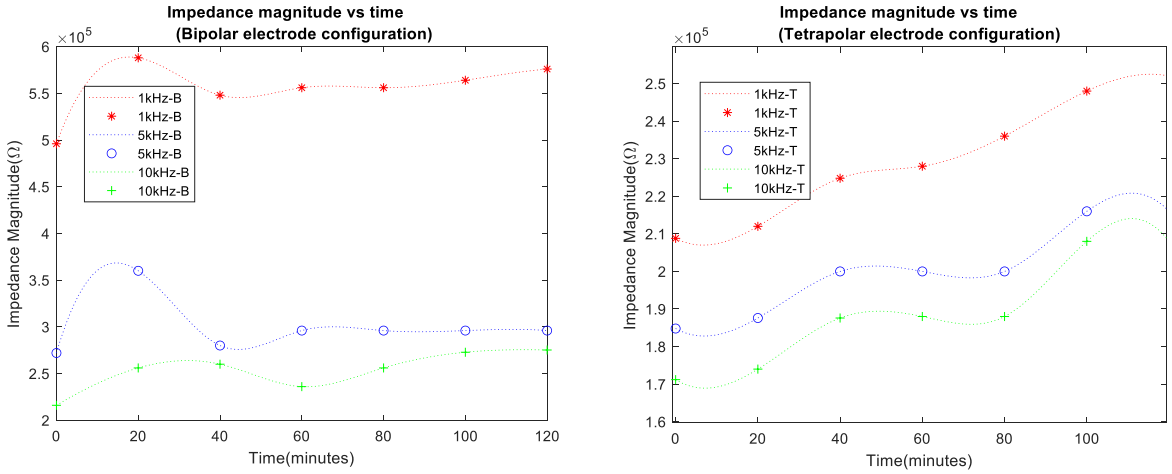


Figure 9.16: left) Impedance magnitude vs. time for bipolar electrode configuration. right) Impedance magnitude vs. time for tetrapolar electrode configuration.

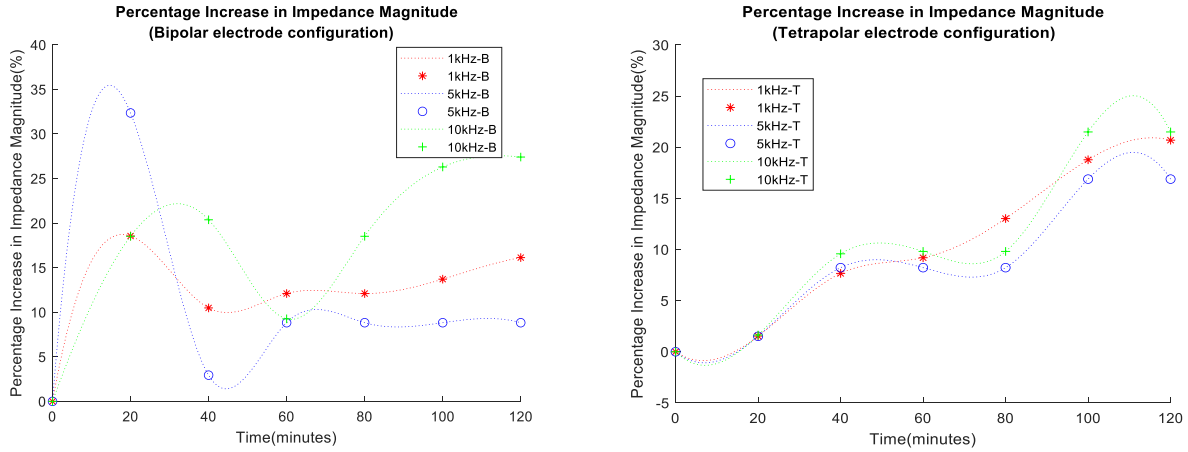


Figure 9.17: left) Percentage increase in impedance magnitude for bipolar electrode configuration. right) Percentage increase in impedance magnitude for tetrapolar electrode configuration.

In addition, as seen in Figure 9.18 below, the corneometer was used to assess the hydration status of the porcine skin sample at the start and end of the experiment. These results provide valuable confirmation of the sensor's accuracy and reliability in assessing changes in skin hydration.

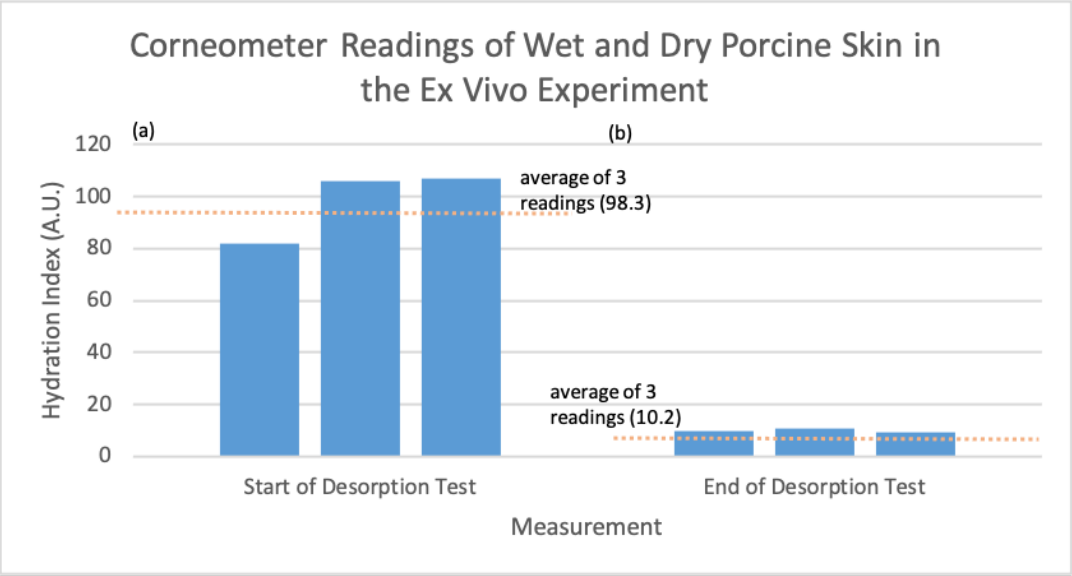


Figure 9.18: Corneometer output for porcine skin before/after desorption experiment (Number = Number of Measurements, Corneometer Value = Hydration Index). (a) Hydration index of porcine skin before desorption, (b) hydration index of porcine skin after desorption

In Vivo Experiments:

The in vivo indicative case study was an investigation aimed at further validating the efficacy and precision of both developed sensors when applied to human skin. The transition to in vivo experimentation represents a fundamental step towards ascertaining the sensors' practical utility and reliability for real-world applications. Human skin exhibits various complexities and inter-individual differences that can influence measurements, rendering this study an essential link between laboratory and practical usage. To establish a robust benchmark, the data obtained from the developed sensors were correlated against the gold-standard corneometer device. Two conditions were examined on the forearms of the participants: dry skin (baseline) and damp skin (simulating external influences and changes in water content).

Table 9.4 displays the results obtained from in vivo experiments, including measurements obtained using the bioimpedance sensing device, optical sensor, and corneometer, under both dry and damped skin conditions. Bioimpedance measurements were performed at 1 kHz, while optical sensor measurements were also taken at the wavelength of 1450 nm. Through Kruskal–Wallis testing, it was found that the *p*-value was 0.0073; thus, we are able to reject the null hypothesis and assume statistical significance with these results.

The introduction of ionized water on the skin resulted in a reduction in the measured impedance magnitude. This can be attributed to the substantial influence of dissolved ions on the electrical conductivity of water, providing it with the capacity to conduct electricity. In the skin impedance measurements, the incorporation of ionic water augmented conductivity by enhancing the mobility of ions. Consequently, this augmentation led to a decrease in impedance levels at the measurement point. The anticipated decrease in impedance magnitude when applying ionized water to the skin aligned with corneometer measurements, indicating an increase in hydration levels following water application. Trends from the optical sensor output presented a decline in measured voltage with increased hydration, analogous to the bioimpedance sensor.

Figure 9.19 presents a bar chart illustrating measurements from the three systems detailed in Table 9.4. Notably, under dry skin conditions, optical measurements exhibited higher reflected voltage compared to damp skin, indicating reduced light absorption. Bioimpedance measurements displayed a higher impedance value for dry skin than for damp skin, which was attributed to decreased ionic presence on the skin surface due to reduced water content. Corneometer values were higher for damp skin than dry skin, as anticipated due to increased capacitance from elevated water levels on the skin surface. While all systems demonstrated agreement, the Corneometer exhibited a higher rate of change between the two states, from dry to damp skin, with the optical technique being the least reactive. This figure is intended solely to visualise the direction and relative magnitude of change between timepoints across the different modalities. Although each system operates in distinct units (impedance, voltage, and arbitrary units), the grouped display is used for illustrative comparison rather than for direct quantitative equivalence.

Table 1.4: Output sensor results for in vivo experiment over 2 conditions on the forearm.

Condition	**Dry**	**Damp**
Raw Optical @1450 nm	2.38 V	2.23 V
Bioimpedance @1 kHz	1.6 kΩ	960 Ω
Corneometer (A.U)	38.6	88.2

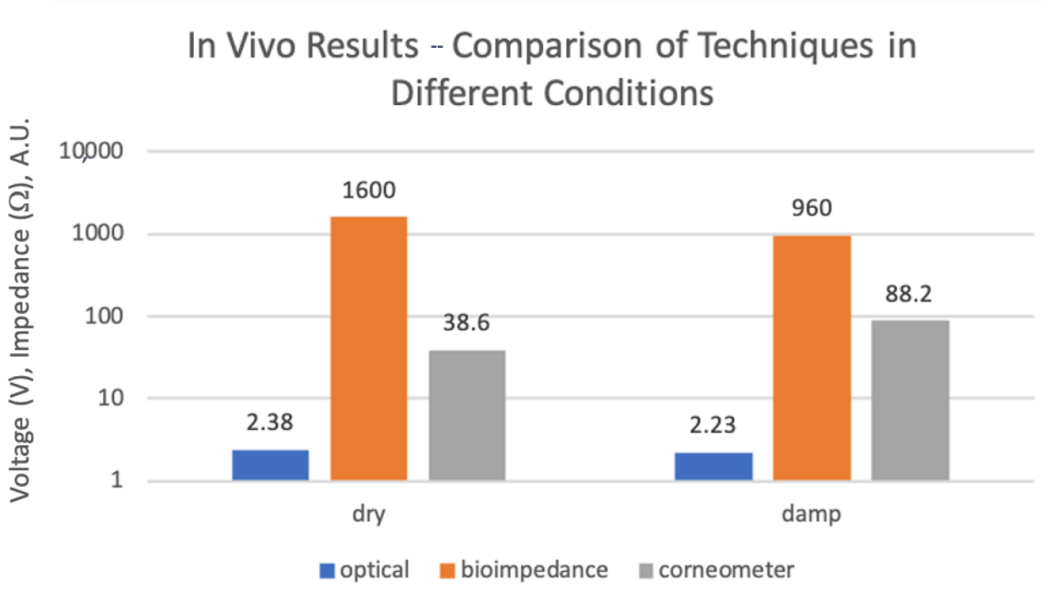


Figure 9.19: Comparison of outputs from all techniques for in vivo experiment over both conditions (logarithmic scale).

9.2.4 Analysis of Results

Multiple linear regression (MLR) was selected as the modelling method for understanding hydration dynamics through various measurement variables due to its ability to handle multiple independent variables influencing a single dependent variable. This method provides interpretable coefficients, allowing for a quantification of the impact of each variable on hydration status, which is crucial for understanding the specific contributions of different measurements. The assumption of linearity aligns with the expectation that the relationships between measurements and hydration are reasonably linear. Additionally, the statistical significance tests offered by multiple linear regression aid in identifying which measurements significantly contribute to predicting the dermal water content.

The results from the MLR model involved three independent variables (x1, x2, and x3), being the optical, bioimpedance, and gravimetric measurements, predicting a dependent variable (y). The model equation is expressed as ($y = \beta0 + \beta1x1 + \beta2x2 + \beta3x3$), where β0 is the intercept and β1, β2, and β3 are the coefficients for the respective variables [142]. The estimated coefficients indicate the contribution of each independent variable to the dependent variable. The intercept (β0) was estimated to be −3657.4, representing the expected value of y when all independent variables are zero. The

coefficients for x1, x2, and x3 were 1411.8, −0.49935, and −0.94791, respectively. These coefficients represent the expected change in y for a one-unit change in the independent variable, holding other variables constant. The statistical significance of each coefficient was assessed through t-statistics and associated p-values. The t-statistics measure the number of standard deviations away from zero that a coefficient is located. The lower p-values obtained, 2.9697 × 10−10, 0.00066671, 6.594 × 10−11, indicate that each variable was statistically significant in predicting y.

The model fit was assessed through various metrics. The R-squared value of 0.996 suggested a very strong fit, while the adjusted R-squared value remained high at 0.995, supporting the model's reliability. The root mean squared error (RMSE) was 0.501, representing the average prediction error. The R-squared value of 0.996 was very high, which can be justified due to the experiments being carried out ex vivo on porcine skin. Typically, there would be an expected decline in this correlation on human skin due to other affecting factors, e.g., skin texture and composition.

The F-statistic tested the overall significance of the regression model. The low p-value (6.45 × 10−21) suggests that at least one variable was significant in predicting y, affirming the overall model's significance. The overall results from this MLR model can be seen in Table 9.5. Figure 9.20 displays the actual vs. predicted plot, which visualizes the model's predictive performance by plotting the actual gravimetric measurements against the predicted values. The R-squared trend over time visualizes how the R-squared changed over time. This provides insights into how the model's performance varied across different periods, as it was indicated to stay consistently at its high significance level of 0.996. The bar plot for p-values visually compares the p-values associated with each variable, being optical, bioimpedance, and gravimetric measurements. This helps to identify which variables had a more significant impact on the dependent variable, which was demonstrated to be the output obtained from the bioimpedance sensor, exhibiting a significantly small p-value. The results can be seen below in Table 9.5, and output plots displayed in Figure 9.20.

It is important to note that the use of multiple measurement techniques inherently involves different unit scales (e.g., voltage for optical measurements, ohms for bioimpedance, and mass for gravimetric measurements). The regression analysis in this context does not imply direct equivalence or sensitivity comparison between these methods. Rather, the MLR model statistically evaluates how each method contributes to

explaining the variance in hydration-related changes. The visual and statistical outputs shown are used solely for assessing relationships and predictive contributions. Although all three methods respond to hydration state changes, each capture different physiological characteristics: gravimetric mass loss as a reference standard for water content change, optical measurements through reflectance variations, and impedance through conductivity alterations. The comparison among them is not made to equate sensitivity, but rather to understand complementary data contributions in modelling hydration dynamics.

Figure 9.22 displays the magnitude of the voltage change (V) for both the optical and bioimpedance results in response to each unit in gravimetric decline. This shows a greater magnitude of voltage change per gravimetric unit in the bioimpedance over the optical measurements. Figure 9.21 presents the actual vs. predicted plots for the optical and bioimpedance output alone against the predicted gravimetric TEWL response. The MLR models were run for each technique, with the R-squared values of 0.888 and 0.97 and p-values of 1.66 × 10−9 and 2.09 × 10−4, respectively, expressing strong correlations for both individually, but more so in the electrical results. In addition, Figure 9.23 displays the box plot outputs for the Kruskal–Wallis statistical analyses for the three measurement techniques, presenting the spread and variance of each technique.

Table 9.5: MLR model results for optical sensor and bioimpedance sensor voltage outputs against gravimetric measurements.

	Estimated Coefficient	**Standard Error**	**T-Statistic**	***p*-Values**	**Root Mean Squared Error**	**0.501**
β_0	−3657.4	267.52	−13.671	7.1367×10^{-10}	R-squared value	0.996
X_1	1411.8	97	14.554	2.9697×10^{-10}		
X_2	−0.49935	0.11685	−4.2733	0.00066671	Adjusted R-squared value	0.995
X_3	−0.94791	0.05856	−16.187	6.5946×10^{-11}		

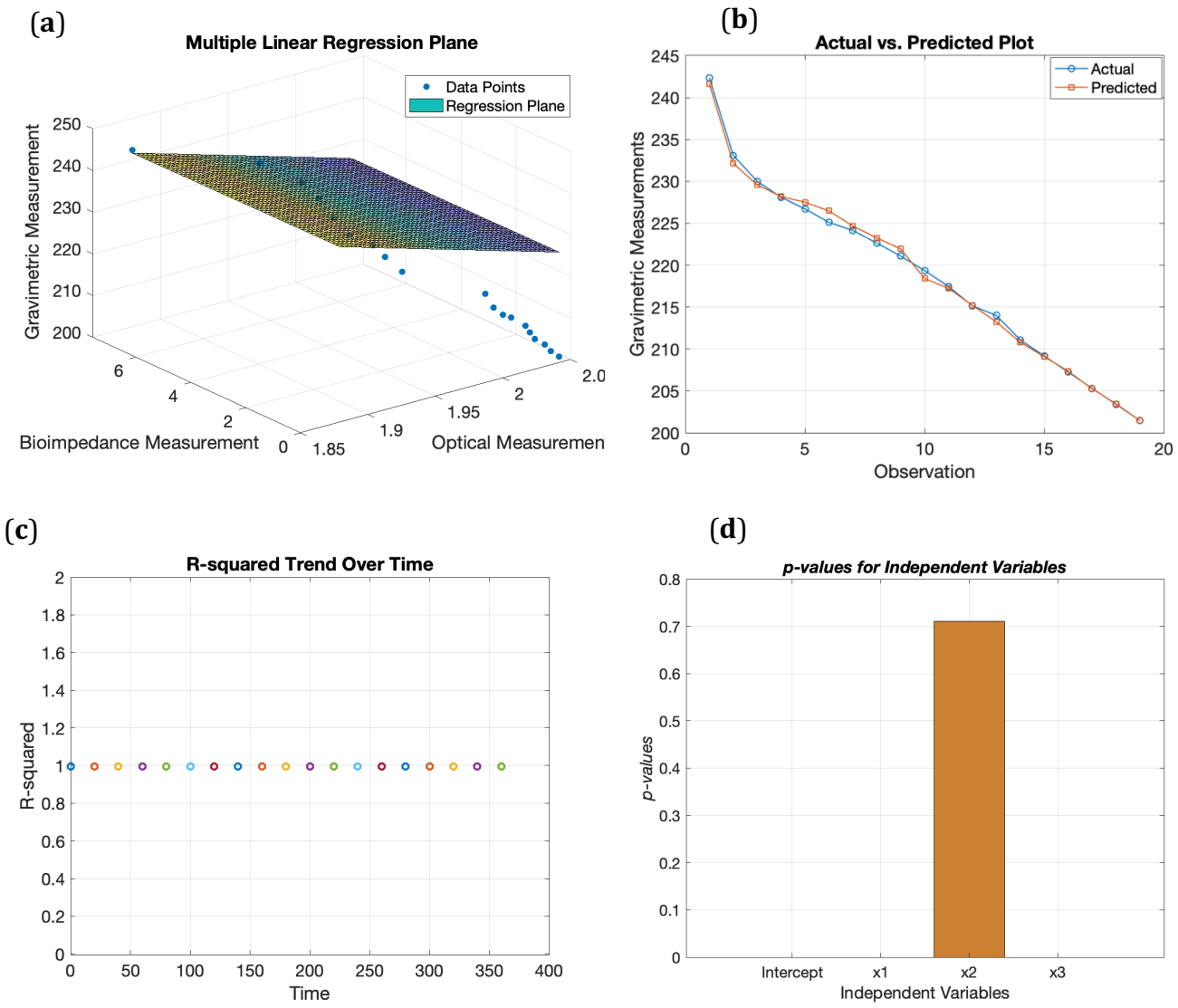


Figure 9.20: MLR model results for optical and bioimpedance sensor outputs against gravimetric measurements for ex vivo experiment: (a) regression coefficients plane, (b) actual v predicted plot, (c) R-squared trend over time, (d) p-values for independent variables.

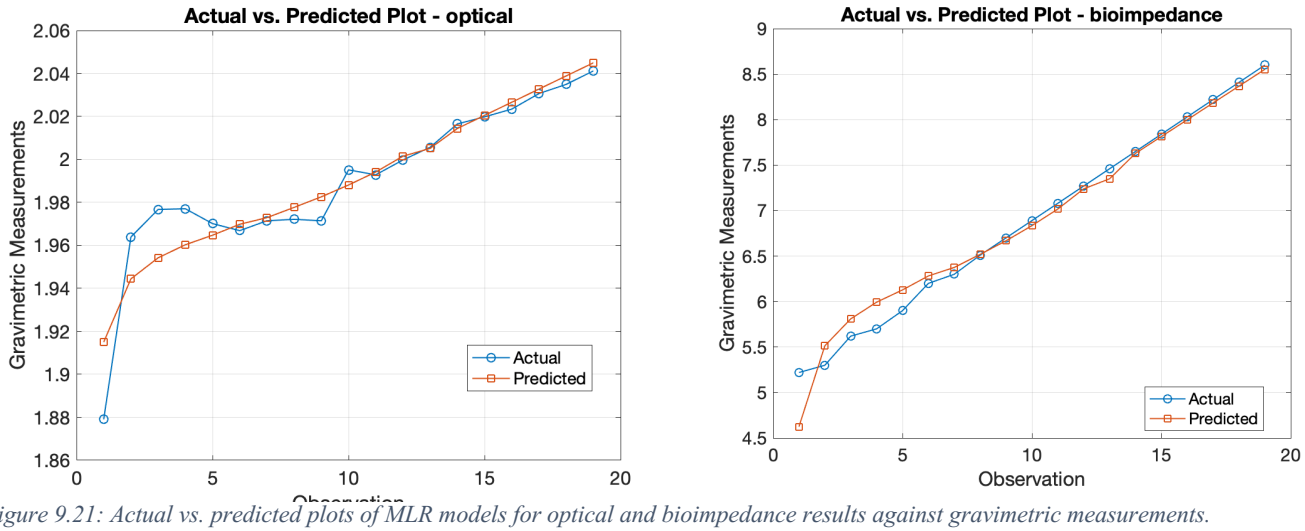


Figure 9.21: Actual vs. predicted plots of MLR models for optical and bioimpedance results against gravimetric measurements.

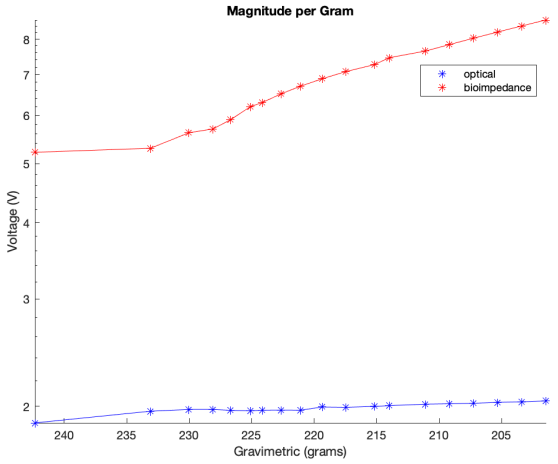


Figure 9.22: Magnitude of voltage change per gravimetric unit change.

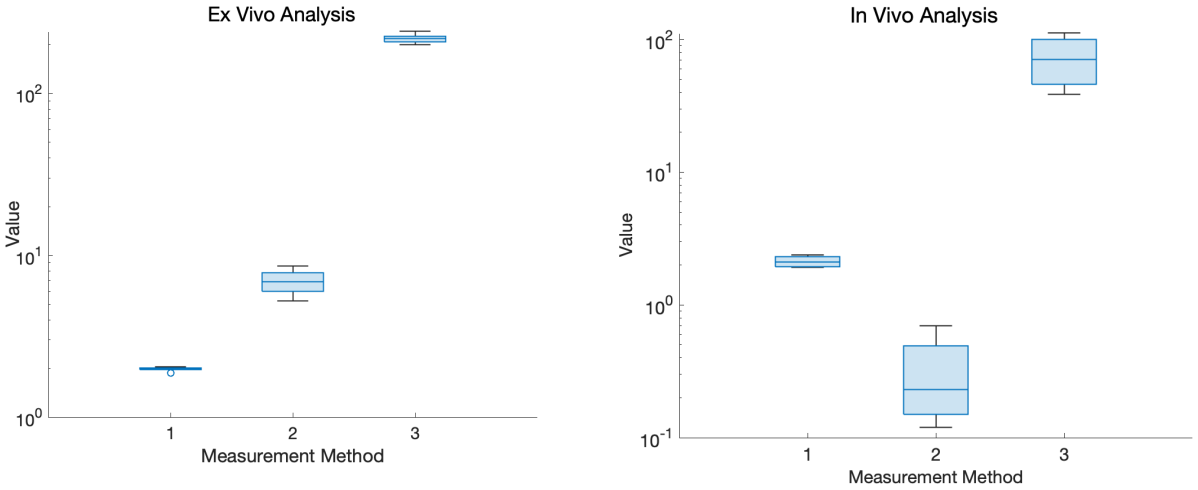


Figure 9.23: Box plot variance outputs for Kruskal–Wallis statistical analysis for 'measurement method 1. optical, 2. bioimpedance, and 3. reference measurements' in ex vivo (left) and in vivo (right) experiments.

9.2.5 Summary

In this section, a second ex vivo experiment was conducted using the same custom-made optical sensor to further explore its responsiveness to hydration changes in biological tissue. The test used porcine skin, prepared in a humidity chamber and monitored over a three-hour period, during which reflectance and gravimetric measurements were captured simultaneously.

The gravimetric data exhibited a progressive reduction in sample mass, consistent with TEWL via evaporation. Correspondingly, the reflectance data from both the developed sensor and a reference spectrophotometer increased as dehydration advanced, confirming the expected inverse relationship between reflectance and tissue water content. A comparative assessment showed a similar trend in both instruments, with the strongest reflectance fluctuation observed at 1450 nm—further affirming the sensitivity of this wavelength to superficial hydration levels.

Regression analysis, including MLR and PLS models, indicated high correlation coefficients between sensor readings and gravimetric reference values. The developed sensor consistently demonstrated lower error and bias compared to the spectrophotometer, offering further support for its capacity to capture meaningful optical signals associated with tissue hydration. However, these initial findings should be interpreted with caution; repeat experiments and broader testing are required before asserting claims of measurement reproducibility or superiority.

The study also evaluated the integration of a bioimpedance sensor using both tetrapolar and bipolar configurations. Tetrapolar setups provided improved signal stability by reducing the influence of electrode–skin interface impedance. Bioimpedance measurements showed an increase in impedance values over time, aligning with the expected loss of water content in the tissue.

Electrical techniques, such as bioimpedance analysis, operate by applying an alternating electrical current through the skin and measuring the resulting voltage to assess the skin's electrical properties. These methods take advantage of the frequency-dependent behaviour of biological tissues, where low frequencies (typically below 10 kHz) primarily travel through extracellular fluid because the cell membranes act as capacitive barriers, whereas higher frequencies can penetrate cell membranes and access intracellular compartments. This frequency-dependent response allows discrimination

between extracellular and intracellular water content. In this work, frequencies of 1 kHz, 5 kHz, and 10 kHz were selected to target this range, capturing the transition from extracellular-dominant pathways to partial intracellular contribution, while remaining within the practical limitations of the measurement setup. The electrical techniques provide complementary information to optical measurements by offering insights into the conductive and dielectric properties of the tissue, which are indirectly influenced by water content. However, both techniques ultimately reflect changes in skin hydration, albeit through different physical principles—optical techniques measure water absorption directly, while electrical techniques infer hydration from changes in tissue conductivity and permittivity. While there is some overlap in the physiological information they capture, combining both methods enhances the robustness of hydration assessment by cross-validating hydration-related changes through independent modalities.

Statistical significance was confirmed via MLR and Kruskal–Wallis analysis, supporting the general trend of the data and the feasibility of a multimodal optical-impedance sensing approach. An indicative case study performed on human skin further supported the potential to extend findings from ex vivo to in vivo applications.

It is important to emphasise that comparisons between the optical, impedance, and gravimetric techniques were performed for statistical modelling purposes rather than for direct sensitivity assessment. Each method operates in different units and captures hydration characteristics through different physiological mechanisms—optical reflectance, electrical conductivity, and mass loss, respectively. The combined modelling approach helped identify complementary contributions of each technique to overall hydration tracking, and visualisations were used to assess directional agreement rather than absolute sensitivity across modalities.

Overall, the results highlight that a multimodal sensing method could provide improved resolution of hydration status by capturing complementary features from both optical and electrical domains. Future studies should focus on refining the sensing algorithms and confirming performance through repeated trials in real-world use cases.

CHAPTER 10:

VALIDATION OF DEVELOPED OPTICAL DEVICE VIA IN VIVO EXPERIMENTATION ON HUMAN PARTICIPANTS

10.1 Assessing Skin Hydration Dynamics with a Wearable Optical Sensor: Longitudinal Analysis and Regression Model Approach

10.1.1 Background

Skin hydration is a critical aspect of dermatological health, influencing the skin's appearance, texture, and overall barrier function. The hydration level of the skin is determined by the water content in the stratum corneum, the outermost layer of the epidermis, which is influenced by various factors including moisturization, sebum production, and transepidermal water loss (TEWL). Moisturization refers to the process of applying substances that help to maintain or restore hydration, enhancing the lipid bilayer of the skin and reducing TEWL. Sebum, an oily substance produced by sebaceous glands, plays a vital role in forming a protective barrier that helps retain moisture[143], [144], [145], [146], [147].

The motivation behind measuring skin hydration lies in its significance for both cosmetic and clinical applications. Accurate and continuous monitoring of skin hydration can aid in the development of better skincare products, offer insights into various skin conditions, and improve the management of diseases such as eczema and psoriasis. Traditional methods of assessing skin hydration, like corneometry and TEWL measurement, provide valuable information but are often limited by their invasive nature, the need for direct contact, and the lack of real-time monitoring capabilities.

Optical sensing techniques, particularly Near-Infrared Spectroscopy (NIRS), have emerged as powerful tools for non-invasive skin measurements. NIRS utilizes the interaction of near-infrared light with skin tissues to provide information about the water content and other molecular compositions of the skin. This technique is advantageous due to its non-invasiveness, ability to penetrate deeper layers of the skin, and real-time monitoring potential. The absorption spectra of water and skin components in the near-infrared range are well-documented, allowing for precise analysis of skin hydration levels. It can be seen in the water spectra that there are significant distinct absorption peaks around 975 nm and 1450 nm, which are particularly sensitive to water concentration changes, thus making these wavelengths critical for accurate hydration assessment[48], [64], [91], [94].

Skin characteristics such as colour, thickness, and underlying structure can vary significantly among individuals, influencing the accuracy of hydration measurements. This study will explore these variations to understand their impact on optical sensing outcomes and to improve the robustness of hydration assessments across diverse populations[148], [149].

The primary aim of this experiment is to investigate the longitudinal effects of moisturization on skin hydration using an optical sensing wearable device. This study employs a calibration regression model to analyse the collected data, correcting for individual differences and external factors to provide more accurate and personalized hydration measurements. The device leverages water-sensitive wavelengths of 975 nm and 1450 nm, offering a direct measurement advantage over traditional electrical devices[125].

In summary, this experimental section aims to:

1. Evaluate the effectiveness of a wearable optical sensor in monitoring skin hydration over time.
2. Assess the impact of moisturization on skin hydration levels.
3. Investigate the influence of individual skin differences on hydration measurements.
4. Develop and validate a calibration regression model for improved accuracy of optical sensing data.

10.1.2 Methodology and Materials

Subjects:

This study was conducted with the approval of the City University of London Ethics Committee. Prior to participation, all subjects were provided with an information sheet detailing the study's purpose, procedures, and potential risks. Each subject gave written informed consent by signing a consent form, ensuring they understood and agreed to participate in the research. A total of 20 healthy volunteers, aged between 21 and 35 years, were recruited for the study. The selection aimed to include a diverse sample representing different ethnicities and skin tones to assess the variability in skin hydration

measurements. The volunteers presented a range of skin types, including dry, combination, and normal skin, based on self-reporting.

Inclusion criteria for the study were:

1. Age over 21 years.
2. Healthy skin without any dermatological conditions.
3. Willingness to refrain from using any new skincare products during the study period.
4. Agreement to follow the study protocols and attend all scheduled measurement sessions.

Exclusion criteria included:

1. Presence of any skin disease or condition that could affect hydration measurements.
2. Recent use of topical or systemic treatments that could influence skin hydration.
3. Participation in another similar clinical study within the last three months.

The demographic details of the subjects, including age, ethnicity, and skin type, were recorded at the initial visit. This diverse cohort enabled the investigation of the impact of individual differences on skin hydration levels and the effectiveness of moisturization over time.

Preliminary Baseline Measurements & Participant Questionnaire:

Before the commencement of the experiment, each participant completed a detailed questionnaire. The questionnaire gathered comprehensive demographic and skincare-related information, including age, gender, race, self-reported skin type, Fitzpatrick scale classification, frequency of moisturization, and use of laser treatments, waxing, and tanning treatments. This information was collected to facilitate a thorough analysis of the factors influencing skin hydration and to assess potential confounding variables. This can be seen in the appendix.

Baseline skin hydration measurements were taken prior to the application of any moisturization treatments. These measurements were performed using three different devices to ensure accuracy and reliability: a developed optical sensor, a Corneometer, and

a spectrometer connected to a fibre optic reflectance probe. This also provided standard reference measurements to validate the developed optical sensing device against. The developed optical sensor, an innovative wearable device, was used to measure skin hydration levels non-invasively. This device utilizes near-infrared spectroscopy (NIRS) to assess the water content in the skin. The Corneometer, a widely recognized gold standard device for measuring skin hydration, was employed to provide a quantifiable measure of the stratum corneum's moisture content by assessing the capacitance of the skin surface. Additionally, a spectrometer connected to a fibre optic reflectance probe was used to obtain detailed spectral data from the skin, allowing for the analysis of the interaction between light and skin, providing insights into the skin's optical properties and hydration status.

Measurements were taken at two timepoints: initially (baseline) and at the conclusion of the study (day 7). The combination of these techniques enabled a comprehensive evaluation of skin hydration, ensuring robust and comparative analysis across different methodologies. The results from the preliminary questionnaire and baseline measurements were recorded meticulously, serving as a foundation for subsequent analysis and allowing for the correlation between participant characteristics and changes in skin hydration over the study period.

Experimental Protocol:

Upon signing consent forms, participants proceeded with the experimental protocol, which involved moisturizing a single forearm twice a day for seven consecutive days. The other arm remained untreated to serve as a control throughout the study duration. The moisturizer used in the study was NIVEA 48hr Moisturising Lotion (NIVEA, Germany). This widely available moisturizer was chosen for its common use and effectiveness in hydrating the skin. Three measurement devices were utilized to assess skin hydration levels on both arms on day 0 and day 7 of the study:

1. ***Developed Optical Wearable***: The device recorded output voltage values averaged over 30 seconds, providing continuous monitoring of skin hydration levels.
2. ***Corneometer Device***: This device provided an averaged hydration index value based on three measurement instances, offering a quantitative assessment of skin hydration.

3. ***Spectrometer Connected to Reflectance Fibre Optic Probe***: The spectrometer recorded total absorption spectra from 900 to 1700 nm, allowing for detailed analysis of skin hydration and optical properties.

Each participant yielded 12 sets of measurements, comprising three measurements on each arm at each timepoint (day 0 and day 7). This comprehensive approach allowed for a thorough evaluation of the longitudinal effects of moisturization on skin hydration levels. Moreover, the experimental protocol ensured standardized procedures and data collection methods, facilitating accurate and reliable assessment of the impact of moisturization on skin hydration over the study period.

10.1.3 Results

Optical Sensor Measurements:

Figure 10.1 presents a bar chart illustrating the differences in the optical sensor voltage output between the two measurement timepoints for each participant. Each bar colour represents a different LED wavelength on the sensor (975 nm, 1050 nm, 1300 nm, and 1450 nm). The chart displays the output for all 20 participants, with the control arm results on the left side and the treated arm results on the right. This comprehensive visual representation highlights the variations in skin hydration changes measured by the optical sensor across different wavelengths and conditions. Results show that for all cases there was an increase in the magnitude of difference between timepoints for the treated arm, represented by a larger decrease in the voltage output at the study completion measurement. Furthermore, it can be seen that the wavelengths most sensitive to water, being 975 nm and 1450 nm, have higher magnitude in all participants when compared to those lesser. Figure 10.2 displays the plot for the calculated hydration index using the developed optical sensor trained calibration model, taking into account individual differences reported from the participant questionnaire as well as the optical measurements. The same pattern can be visualised as in Figure 10.1 in which the moisturised arm on the follow up timepoint experienced a greater increase in calculated hydration index, thus a higher moisturisation and hydration level reported in the skin.

Figure 10.3 below focuses specifically on the 1450 nm wavelength, re-plotting solely this data as a bar chart to facilitate easier visual analysis. The 1450 nm wavelength was chosen due to its high sensitivity to water absorption, making it more responsive to smaller changes and fluctuations in skin hydration. This wavelength also produces higher magnitude voltage outputs, thus showing a more pronounced change in hydration levels. In this figure, the control arms (blue bars) and treated arms (red bars) for each participant are positioned adjacently, allowing for a clearer comparison of hydration changes between the two arm conditions.

Analysis of the results reveals a distinct pattern: more than half of the participants' control arms exhibited a decrease in skin hydration over the study period, indicating that their skin became drier. In contrast, the remaining participants showed a smaller increase in skin hydration on the control arm compared to a significant increase observed in the treated arm. This disparity suggests that while the treated arms consistently benefited from the twice-daily application of the NIVEA 48hr Moisturising Lotion, the control arms either experienced a natural decline in hydration or failed to maintain hydration levels without treatment. Several factors may explain these observations. Environmental conditions such as temperature differences could have contributed to the drying effect seen in the control arms of some participants. Additionally, individual differences in skin physiology and baseline hydration levels might have influenced how each participant's skin responded to the lack of moisturization. The significant increase in hydration on the treated arms underscores the effectiveness of the moisturizer in enhancing skin hydration, but it also highlights the variability in how untreated skin can respond under different circumstances.

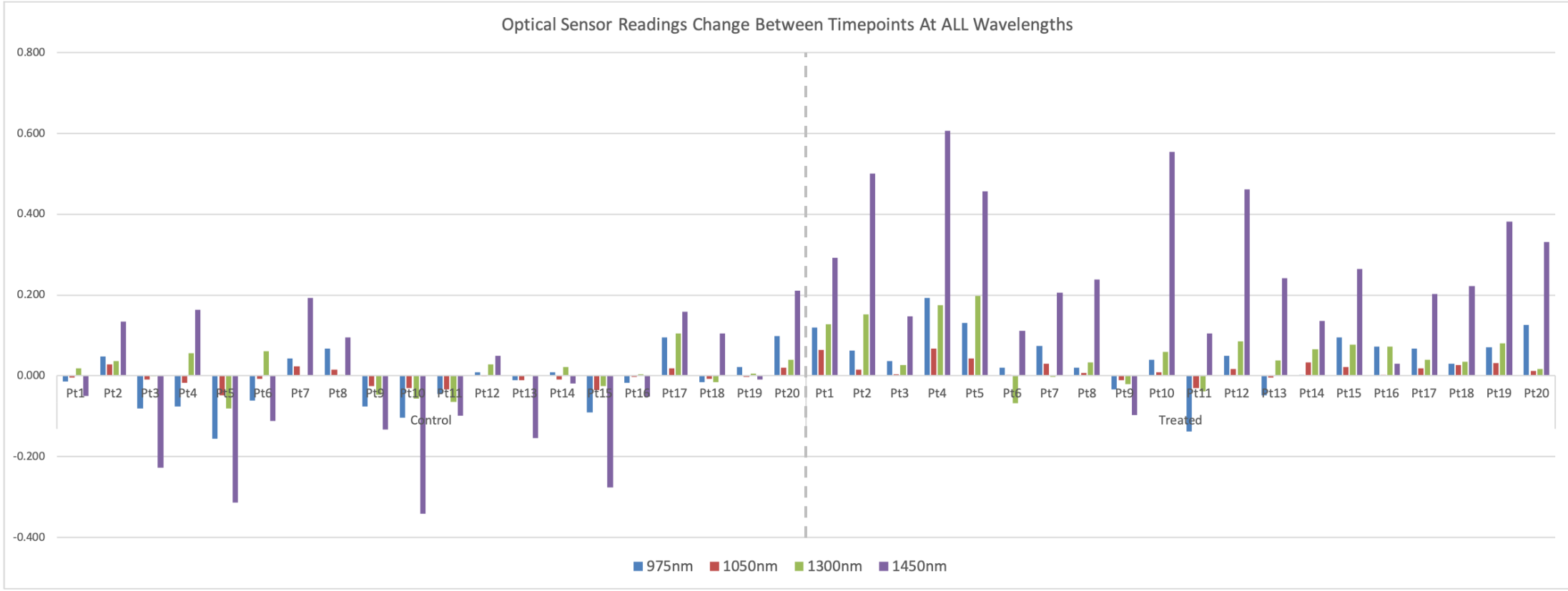


Figure 10.1: Optical Sensor Output Voltage Measurements Between Timepoints for All Participants at All Wavelengths for Both Control and Treated Arms

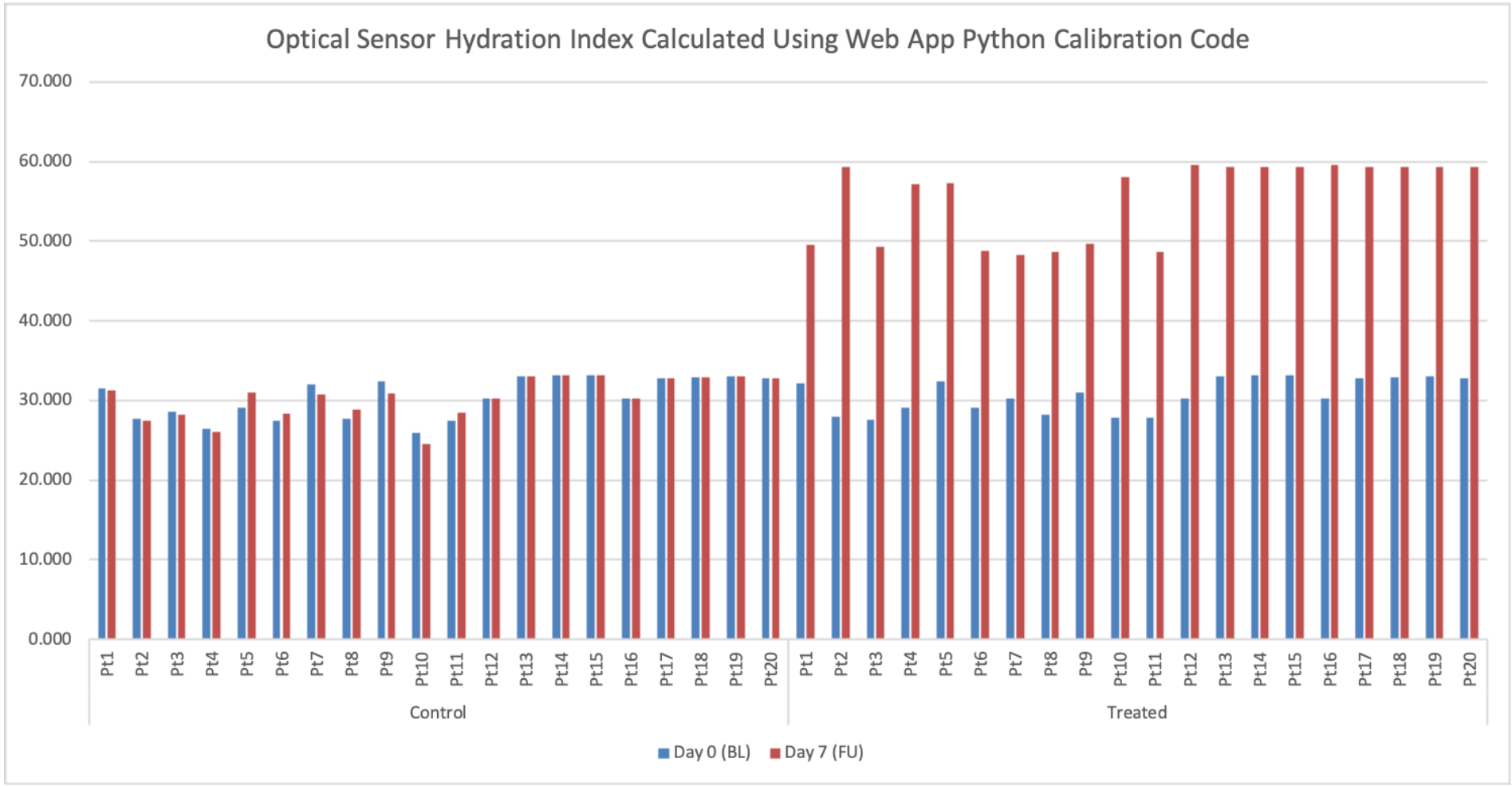
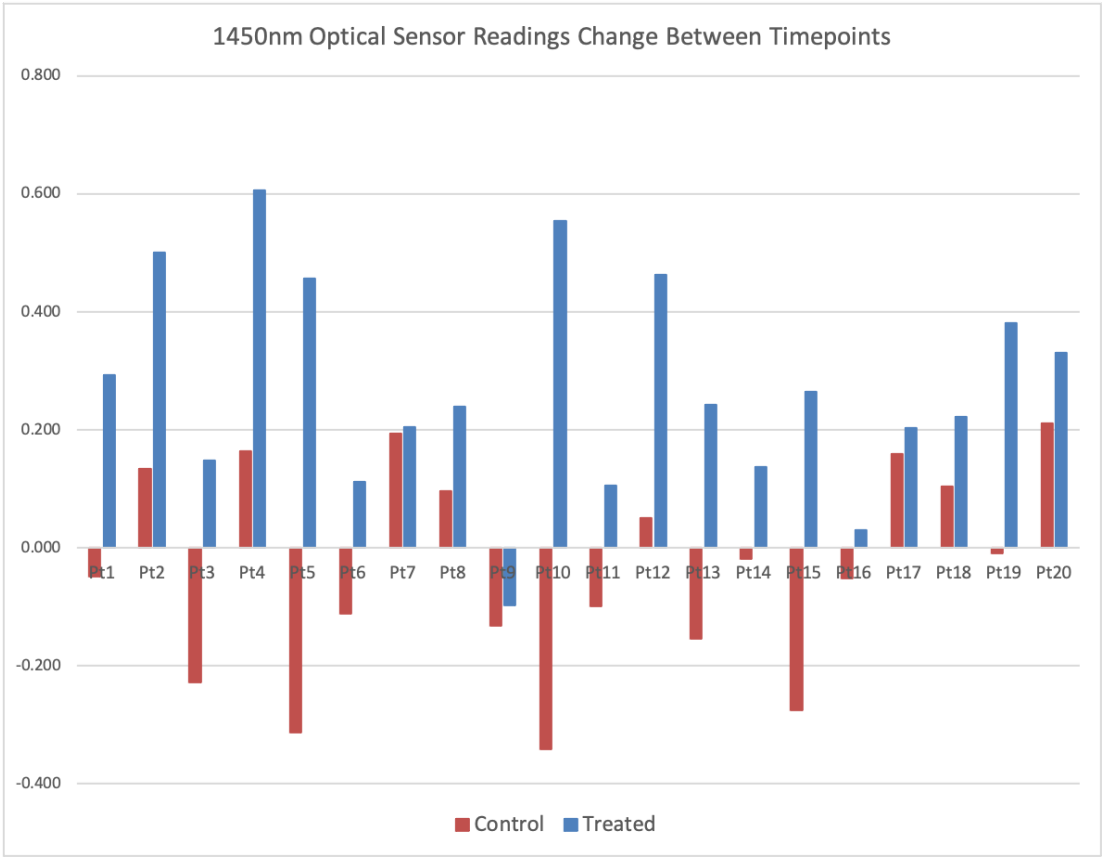


Figure 10.2: Optical Sensor Overall Hydration Index Calculated Using Web App Python Calibration for Both Control and Treated Arms



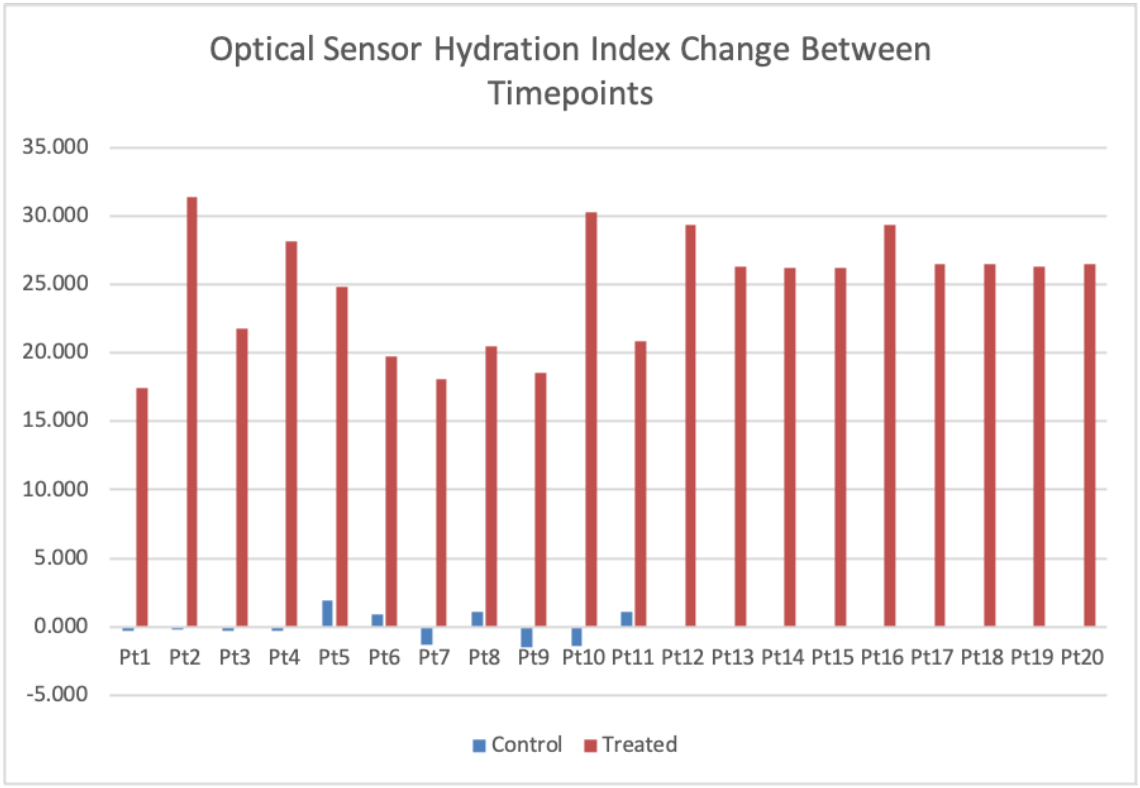


Figure 10.3: top) Extracted 1450 nm optical sensor output voltage measurements between timepoints for control and treated arms. bottom) Converted optical sensor hydration index using web app calibration code between timepoints

Corneometer Measurements:

The Corneometer, regarded as a gold-standard device for skin hydration measurement, provides a hydration index value ranging from 0 to 100. In this study, three single measurements were taken at each timepoint for both the control and treated arms of the participants, and the average of these measurements was used to generate the bar chart displayed in Figure 10.4. The control and treated arms for each of the 20 participants are represented by adjacent bars, facilitating a clearer comparison of the hydration changes over the study period.

The results reveal that all the treated arm bars (blue) exhibit a radically higher change in hydration index between timepoints compared to the corresponding control arms. This pattern holds for all participants except one, where the change was the same for both arms. However, further examination of this participant's optical sensor results at the 1450 nm wavelength shows a very slightly higher change in the treated arm, indicating that the optical sensor was more sensitive to hydration changes for this specific individual. Consistent with expectations, some control arms became drier over time. However, in regard to the control arms, many demonstrated a minor increase in moisturization, albeit significantly smaller than the changes observed in the treated arms. The treated arms consistently showed a substantial increase in hydration, highlighting the effectiveness of the moisturisation.

The bar charts provide a very clear visual distinction between the control and treated arms. The treated arms' marked increase in hydration index underscores the efficacy of moisturisation, while the varying responses in the control arms suggest differences in individual skin physiology and environmental factors. This comparison reinforces the reliability of the Corneometer as a reference measurement and validates the optical sensor's capability to detect changes in skin hydration effectively.

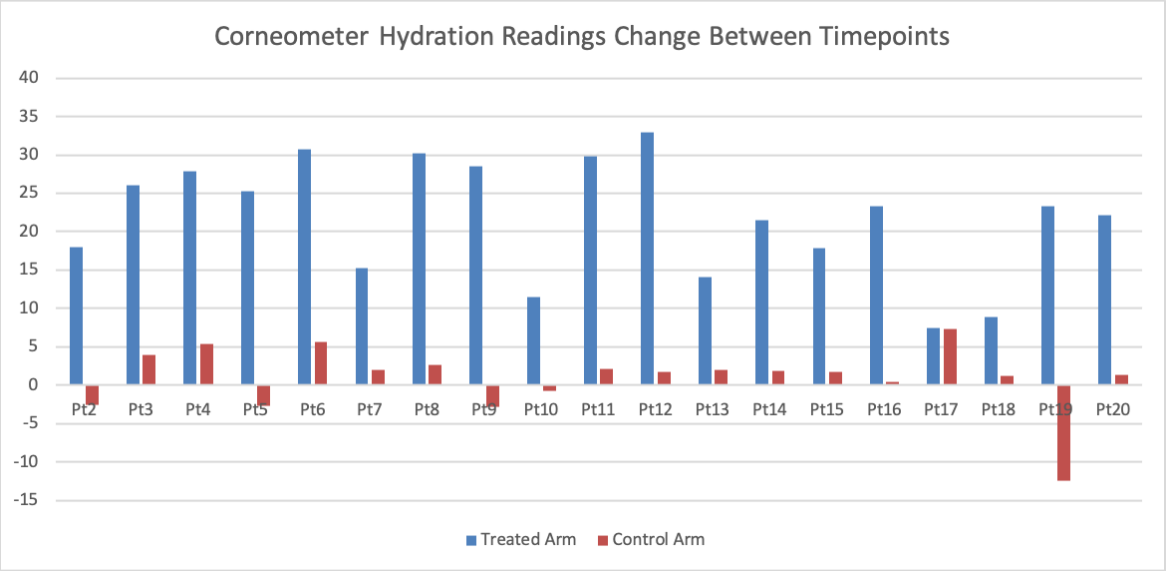


Figure 10.4: Corneometer Measurement Differences Between Timepoints for All Participants for Control and Treated Arms

Spectrophotometer Measurements:

The spectrophotometer provided reflectance spectra from 900 to 1700 nm, with specific values assigned to each nanometre within this range. We focused on four wavelengths (975, 1050, 1300, and 1450 nm) by extracting the corresponding reflectance values at these specific wavelengths for each arm and at each time point.

At 1450 nm, which is highly sensitive to water concentration changes in the skin, we calculated the percentage differences in reflectance between the treated and control arms. A negative percentage change indicates an expected increase in water absorption. Out of 20 participants, 14 showed the expected higher magnitude negative trend in the treated arm, resulting in a 70% effectiveness for the spectrophotometer. These are highlighted in the line graph in Figure 10.5 below, where the treated arm (blue) can be seen to exhibit a greater percentage decrease than the control (red).

In comparison, the developed optical sensor showed a higher accuracy of 95%, referring to the results in section 3.1. This significantly better performance may be attributed to the sensor's specific design for measuring water concentration directly, while the spectrophotometer's broader spectrum requires more complex interpretation. Additionally, the wearable sensor's optimized design for skin measurements likely contributes to its higher sensitivity and accuracy in practical conditions. Further research is needed to understand these discrepancies and confirm the reliability of the developed sensor.

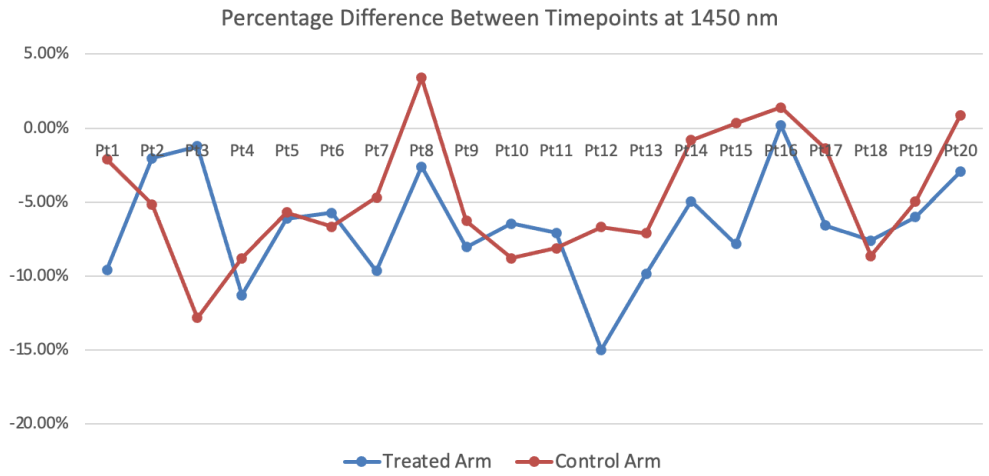


Figure 10.5: Line Graph Showing Percentage Difference Between Timepoints of Spectrophotometer Output at 1450 nm for Each Arm

The benefit of conducting individual-based analysis lies in its ability to capture the variability in skin responses across different individuals. Factors such as baseline hydration levels, skin type, environmental exposure, and individual physiological differences can greatly influence skin hydration and absorption rates. By analysing data at the individual level, we are better equipped to identify trends and patterns that would otherwise be masked in a generalized analysis. This approach allows us to account for outliers, understand personalized responses to moisturization, and develop more accurate and individualized skincare regimens. For example, the variation observed in the 30% of participants who did not show the expected trend with the spectrophotometer highlights the importance of considering individual factors when assessing hydration outcomes. Such analyses are crucial in refining hydration assessment technologies to ensure they work effectively across diverse populations, further supporting personalized skincare solutions.

10.1.4 Correlational Analysis of Results

To evaluate the relationship between the developed optical hydration sensor and the Corneometer, individual hydration index (HI) values from both devices were compared across baseline and follow-up measurements. The correlation analysis (Figure 10.6) produced an R^2 value of 0.36 with a statistically significant p-value of 0.0001, indicating a moderate but meaningful linear relationship. These results suggest that while

the devices are not perfectly aligned, there is a consistent trend between the hydration readings obtained by the optical sensor and the Corneometer.

This R^2 value indicates that approximately 36% of the variation in Corneometer readings can be explained by the optical sensor measurements. Although not indicative of a perfect match, this level of correlation is significant in the context of comparing two fundamentally different sensing modalities—optical reflectance versus electrical capacitance. Moreover, the presence of a statistically significant p-value confirms that this relationship is unlikely to be due to random chance and reinforces the validity of the optical sensor's measurement capability. The scatter of data points around the regression line highlights some individual variability, which is to be expected given differences in skin physiology, environmental influences, and the sensing depth of each device.

A Bland-Altman analysis (Figure 10.7) was also conducted to assess agreement between the two devices. The mean difference was -1.68, indicating that, on average, the optical sensor reports slightly lower hydration values than the Corneometer. The limits of agreement ranged from -20.37 to 17.01, which reflects the variability between measurements but still supports the sensor's ability to track hydration changes in a clinically relevant range. The distribution of the differences did not indicate any systematic bias across the hydration range, suggesting that the optical sensor's deviation is relatively stable and predictable.

Importantly, the optical sensor's design allows it to penetrate deeper into the skin, capturing water content changes not only in the outer stratum corneum (as measured by the Corneometer) but also in the viable epidermis and superficial dermis. These deeper layers are known to hold a more stable and physiologically representative hydration state, especially relevant for chronic conditions or cosmetic interventions. As such, the moderate correlation observed may reflect the complementary—rather than identical—information each device provides. The Corneometer focuses on surface moisture, which can be more susceptible to short-term environmental effects, while the optical sensor incorporates signals from multiple depths, capturing broader hydration dynamics influenced by individual physiology, skin thickness, and structure.

These results validate the optical sensor's performance while also highlighting the unique insights it offers into hydration profiles, supporting its use in real-time, non-invasive skin health monitoring. Future work will further refine its calibration models to enhance consistency across different skin types and hydration states.

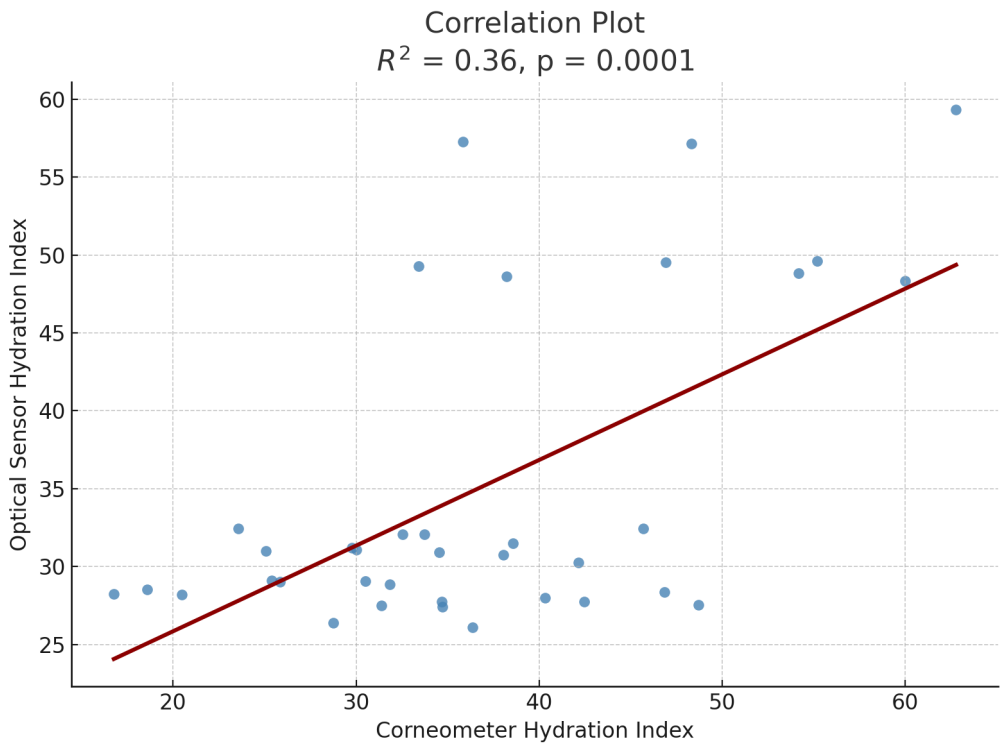


Figure 10.6: Correlation analysis plot showing R^2 between the calculated optical sensor hydration index and Corneometer hydration index

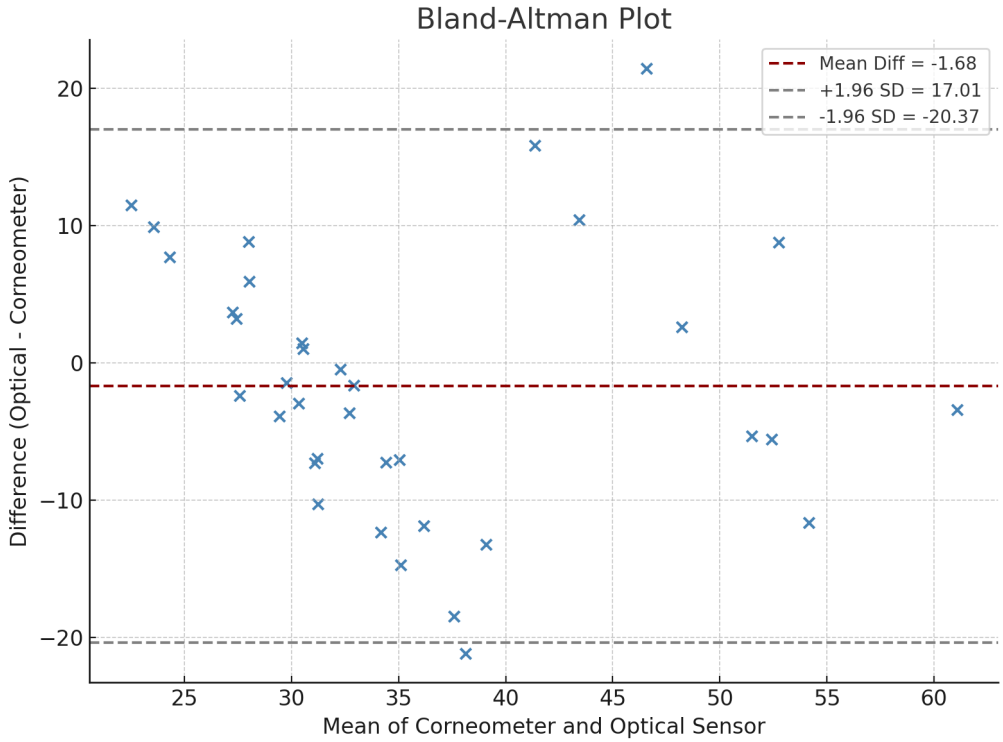


Figure 10.7: Bland-Altman analysis plot to assess agreement between the calculated optical sensor hydration index and Corneometer hydration index

10.1.5 Calibration Regression Model & Statistical Analysis

For the statistical analysis, a regression model was developed in Python using packages such as `scikit-learn` and `pandas`. The model was trained using the 80 sets of data collected from all 20 participants during the in vivo study, encompassing both arms and both timepoints. Each participant's responses from the personal and skin health questionnaire provided additional input variables, including ethnicity, age, sex, moisturization habits, and Fitzpatrick scale number. These, along with the output voltages from the four wavelengths (975 nm, 1050 nm, 1300 nm, and 1450 nm) of the optical sensor, constituted the `X` data for the model. The corresponding Corneometer value, representing the skin hydration index (0-100), served as the `Y` prediction output. The model training process involved a test-train split of 0.3, meaning 70% of the data was used for training the model, while the remaining 30% was used for testing its performance. The regression model aimed to predict the hydration index based on the input variables.

The actual vs. predicted plot extracted from the model, seen in Figure 10.8 below, showed an R-squared value of 0.76, indicating that 76% of the variance in the Corneometer measurements could be explained by the model. This high R-squared value suggests a strong correlation between the predicted and actual hydration index values. Additionally, the mean absolute error (MAE) was calculated to be 5.7, indicating that the model's predictions were, on average, within 5.7 units of the actual hydration index values. This low error further supports the model's accuracy and reliability in predicting skin hydration as well as the developed optical device. In Figure 10.8, the x-axis of the plot represents the data point index (i.e., the individual test samples), ranging from 0 to approximately 25, corresponding to the 30% test split of the total 80 samples. The y-axis shows the actual hydration index values measured by the Corneometer on a scale from 0 to 100.

The regression model also produced plots illustrating the relationships between each input variable and the predicted hydration index (Figure 10.9). These plots provide visual insights into how different factors, such as ethnicity, age, sex, and moisturization habits, influence skin hydration as measured by both the optical sensor and the Corneometer.

Future steps involve continuously updating the input CSV file with new recorded 'X' data, which will enhance the model's accuracy over time. By appending each new set of

data to the training set, the model can be retrained periodically, improving its predictive performance and robustness. This approach ensures that the model remains up-to-date and capable of providing accurate hydration assessments as more data becomes available.

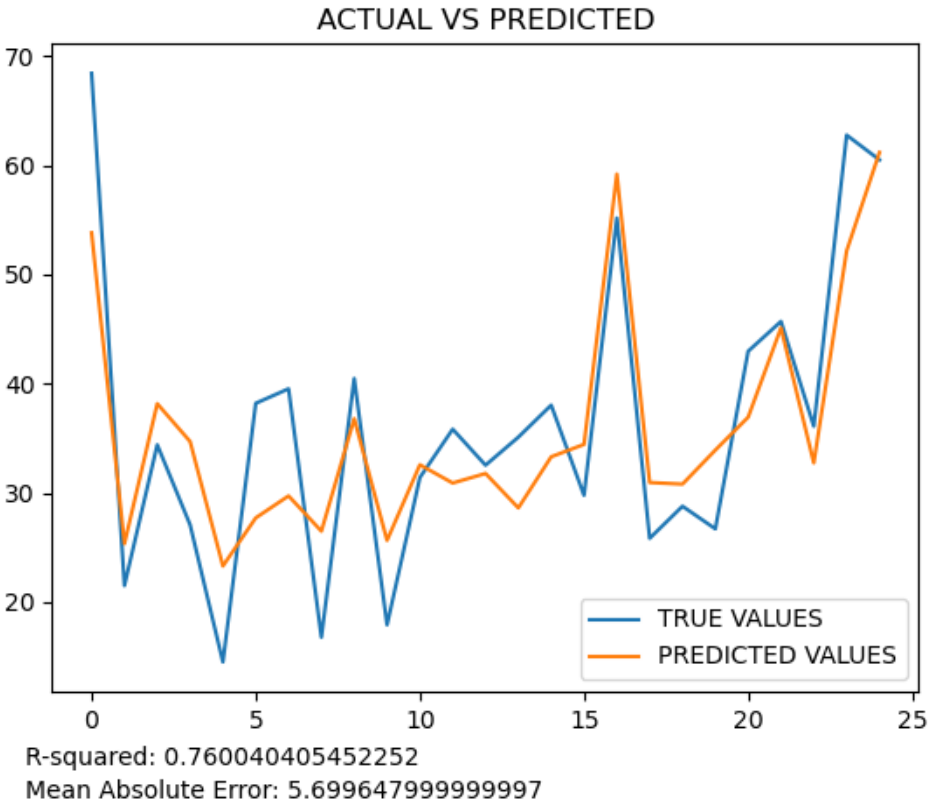


Figure 10.*8*: Actual vs Predicted Plots for Regression Model Output Using Subject Data with a Test-Train Split of 0.3

The statistical analysis also involved plotting the output voltages from the optical sensor against the Corneometer hydration index values for all four wavelengths (975 nm, 1050 nm, 1300 nm, and 1450 nm), as displayed in Figure 10.9(a)-(d). The results indicated negative correlations across all wavelengths, as anticipated. This is because higher hydration levels, as measured by the Corneometer, typically result in lower reflectance and, consequently, lower voltage outputs from the optical sensor due to increased light absorption by water. While the expected trend is encouraging, it is clear that not all correlations are equally strong, nor are they of the same strength. The variation in correlation strength can be attributed to the different principles and sensitivities of the two measurement methods. Optical sensors, including the one used in this study, primarily measure water content through light absorption characteristics at specific wavelengths. However, as mentioned, their readings can be influenced by various skin factors such as skin thickness, melanin content, and other structural properties of the skin. On the other

hand, the Corneometer measures skin hydration by assessing the dielectric constant of the skin, which is affected by surface moisture, sebum levels, and skin smoothness, rather than water concentration directly.

The plots in Figure 10.9(e)-(i) separated the outputs into moisturized and non-moisturized skin from participants, which further highlighted these differences. As expected, the moisturized skin generally showed higher hydration levels, with the directional trend clearly indicated by the red arrows on the figure plots. This trend underscores the optical sensor's sensitivity to changes in skin hydration due to moisturization. However, the variability in the strength of the correlations across different wavelengths also reflects the complex interaction between light and skin tissue, which includes absorption and scattering effects influenced by various skin properties beyond just water content. Since the Corneometer does not take into account individual differences in skin type, the moisturisation split on the plot in Figure 10.9(i) is more substantial than the optical sensor plots. These findings highlight the complexity of measuring skin hydration accurately. While optical sensors offer a direct measure of water content, they are influenced by deeper skin structures and overall skin health. In contrast, the Corneometer provides a surface-level hydration measure that can be affected by immediate skin conditions like oiliness and texture. Understanding these differences is crucial for interpreting the results accurately and improving the calibration models to account for the various factors affecting skin hydration measurements.

Additionally, the statistical analysis included plotting the regression model outputs against various input variables, with a focus on ethnicity and Fitzpatrick scale number in relation to the output voltages from the optical sensor. The analysis revealed a clear trend: participants with darker skin tones, particularly Black and Asian individuals, exhibited significantly lower voltage outputs across all wavelengths compared to those with lighter skin tones. This result is evident in the labelled plots within Figure 10.10. One primary explanation for this finding is the well-documented differences in skin features between various ethnicities. Darker skin tones have higher levels of melanin, the pigment responsible for skin colour. Melanin is known to absorb light, particularly in the near-infrared range used by the optical sensor’s LEDs. Therefore, the increased melanin content in darker skin tones would result in greater absorption of the emitted light, leading to lower reflected signals and subsequently lower voltage outputs from the sensor. In addition, studies have shown that darker skin tones tend to have a thicker epidermis and

a more compact stratum corneum, which can further influence light absorption and reflection properties. The increased thickness can cause more scattering and absorption of light within the skin layers, reducing the amount of light that returns to the sensor's photodiode. These structural differences in skin composition and morphology between ethnicities contribute to the observed variations in the sensor's voltage outputs.

Moreover, the differences in skin features among ethnicities highlight the importance of considering individual skin characteristics when developing and calibrating optical sensing devices. Ensuring that these devices are accurate across diverse populations requires comprehensive data and careful adjustment of the models to account for such variations. This study underscores the necessity of including a wide range of skin tones and types in research to develop universally applicable and reliable health monitoring technologies.

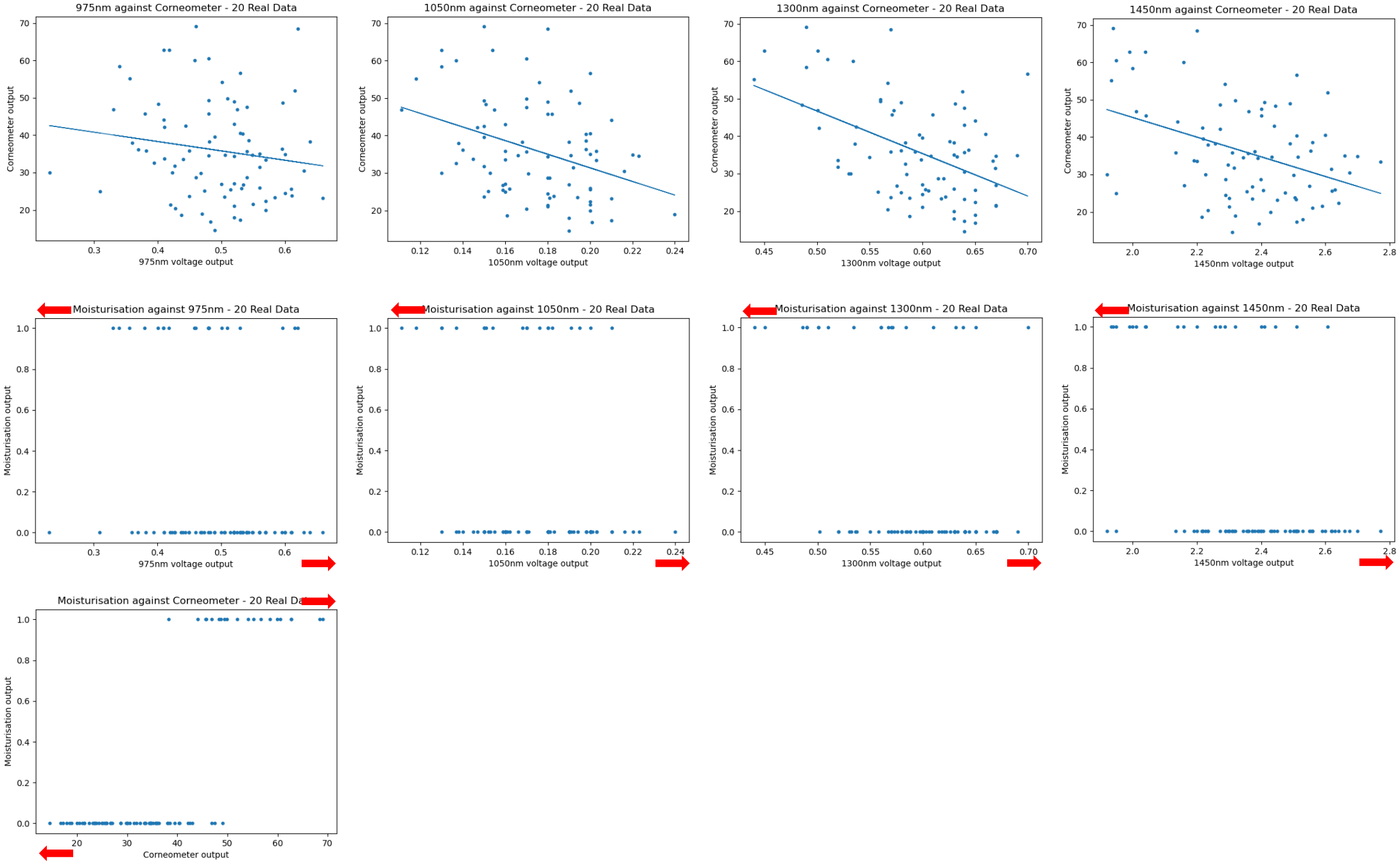


Figure 10.9: Regression model outputs. (a)-(d) Output between optical sensor voltages at 4 wavelengths against the Corneometer output. (e)-(h) Output between moisturisation and voltages at each wavelength. (i) Output between moisturisation and Corneometer output.

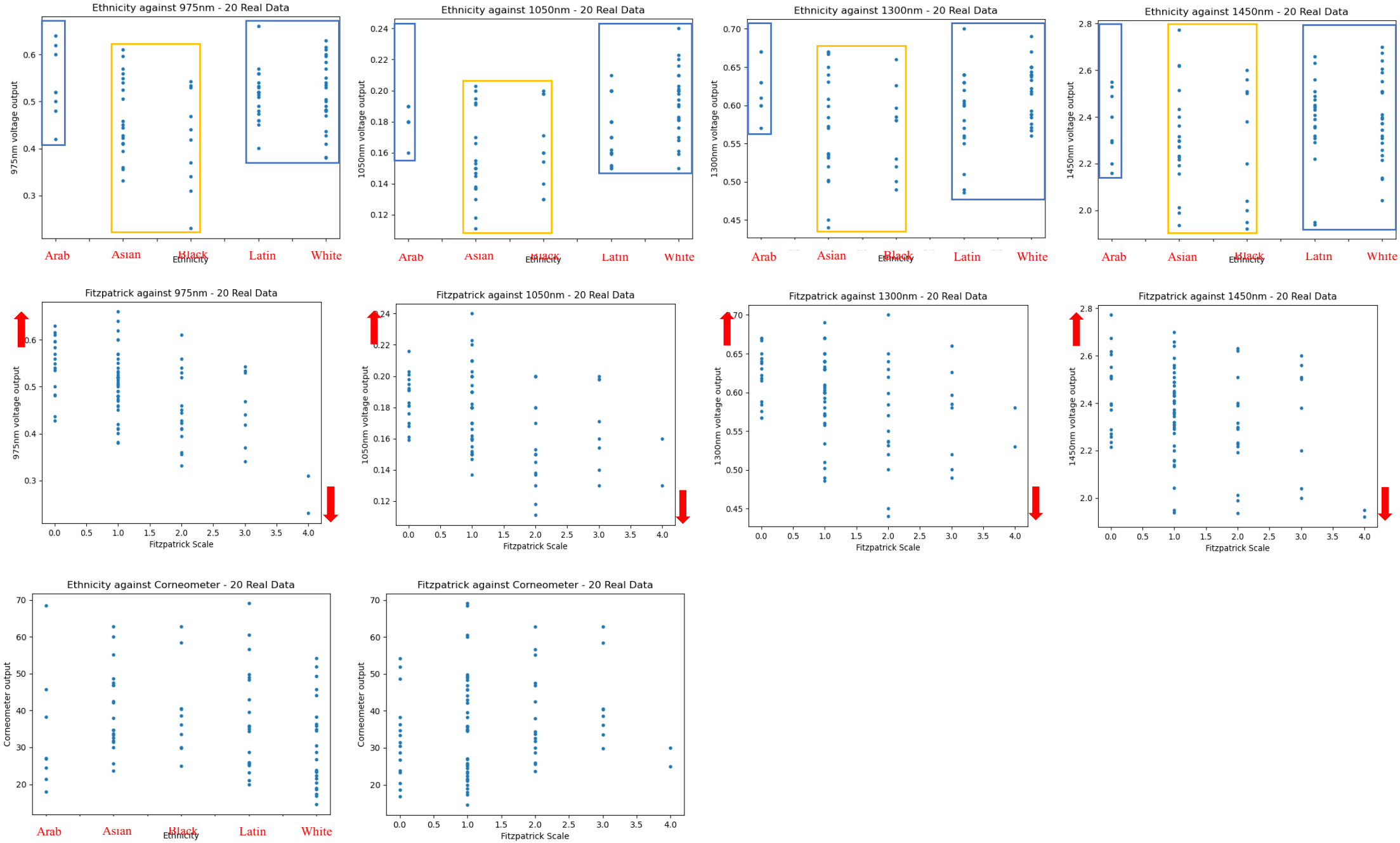


*Figure 10.*10*: Regression model outputs. (a)-(d) Voltages vs ethnicity - yellow is 'Black' & 'Asian', blue is 'White', 'Latin' & 'Arab'. (e)-(h) Voltages vs Fitzpatrick scale (I-V) - red arrow is expected directional trend. (i)-(j) Corneometer against both variables.*

To capitalize on the calibration model and training data derived from the study, a model-based Graphical User Interface (GUI) was developed using Python. This GUI serves as an intuitive, interactive tool for users to assess their skin hydration levels with the optical wearable device developed in this research. The GUI is designed to guide users through a seamless process, ensuring ease of use and accuracy in measurement.

The process begins with the GUI instructing the user to properly position the optical wearable on their wrist. Once secured, the user completes the recording process using the LabVIEW Virtual Instrument (VI) developed for this study. After capturing the optical data, it is automatically saved and made available for analysis by the Python-based GUI. The user is then prompted to input personal details, such as ethnicity, age, sex, moisturization habits, and Fitzpatrick scale number. These inputs are provided through intuitive number inputs and dropdown menus, ensuring that the interface is user-friendly and accessible to individuals with varying levels of technical expertise.

Once the user's inputs are collected, the GUI employs the previously trained regression model, which was developed using 80 sets of data from the study participants. This model integrates the user's input data with the recorded optical sensor voltages to predict the user's current hydration index value. The prediction is then displayed as a hydration index value ranging from 0 to 100, offering users an easy-to-understand estimate of their skin hydration based on the trained model's outputs.

The developed GUI not only provides a straightforward way to assess skin hydration but also lays the groundwork for future enhancements. Potential developments include real-time integration with LabVIEW, allowing for immediate feedback on hydration levels, and the transformation of the code into an app-style program. Such advancements would significantly enhance the accessibility and convenience of the tool, enabling users to monitor their skin hydration effortlessly via smart devices. This continuous improvement approach aims to refine the user experience while maintaining the accuracy and reliability of hydration assessments provided by the developed optical wearable device.

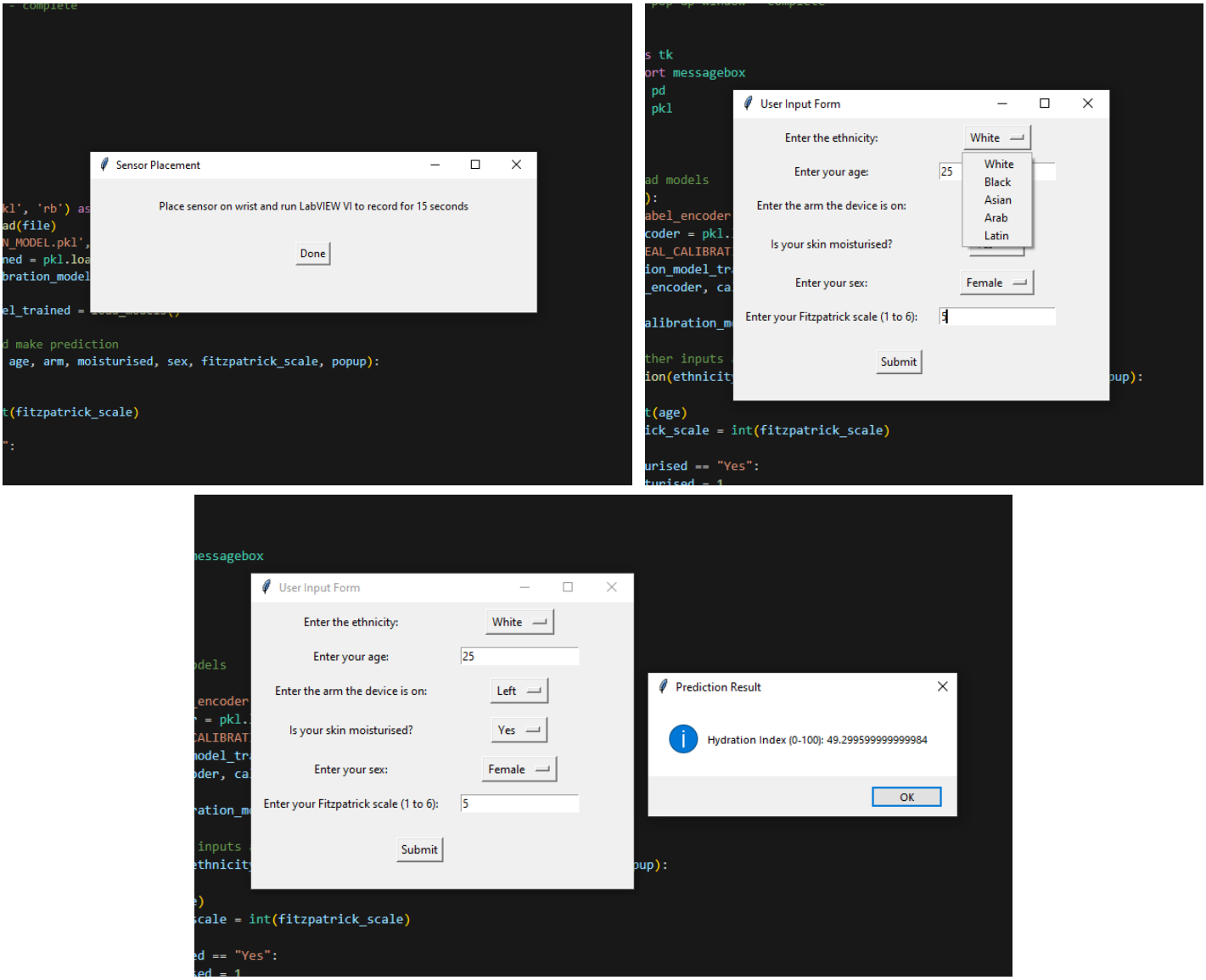


Figure 10.11: Developed Python GUI Using Participant Data Trained Regression Model. (a) Sensor Placement Window. (b) User Input Form. (c) Prediction Result of Hydration Index (0-100)

10.1.6 Summary

Skin hydration refers to the water content within the skin, which is essential for maintaining elasticity, smoothness, and the skin's barrier function. Moisturisation helps improve hydration by reducing trans-epidermal water loss (TEWL) and reinforcing the lipid barrier. This in vivo study examined the effects of a seven-day moisturisation regimen using a wearable optical hydration sensor, benchmarked against the Corneometer—a widely used reference instrument.

Participants applied NIVEA 48hr Moisturising Lotion to one forearm while leaving the other untreated [149]. Measurements taken at baseline and after the treatment period showed consistent increases in hydration in the treated arms. The optical sensor registered corresponding changes in voltage outputs at each wavelength, with the most pronounced shifts at 1450 nm, indicating sensitivity to water content changes in the superficial tissue layers. These findings were consistent with Corneometer readings and further supported by comparative spectrophotometer data [92], [150].

The spectrophotometer showed the expected trend in 14 out of 20 participants (70%), whereas the optical sensor showed consistent detection in 19 out of 20 participants (95%). This enhanced responsiveness may be attributed to the sensor's targeted wavelength configuration and real-time measurement capabilities, which offer greater practical sensitivity than broadband spectrophotometry in dynamic skin conditions [92].

A quantitative comparison between the optical sensor and the Corneometer showed a moderate but statistically significant correlation (R^2 = 0.36, p = 0.0001), demonstrating the sensor's ability to track hydration changes in agreement with a gold-standard reference. Bland-Altman analysis further supported this relationship, showing stable differences across the hydration range. The slightly lower readings from the optical sensor may reflect its capacity to assess water content not only in the stratum corneum but also in deeper layers of the skin, offering a more comprehensive hydration profile. This suggests that the moderate correlation is partly due to each device capturing distinct but complementary physiological information—surface moisture versus broader dermal hydration—highlighting the added value of the optical sensor in future personalised skincare and health monitoring applications.

The optical sensor's integration into a wearable form factor provides added benefit by enabling non-invasive, longitudinal monitoring, reducing operator-dependent variability. Its wavelength-specific approach (975 nm, 1050 nm, 1300 nm, and 1450 nm) allows depth-specific hydration analysis, particularly given the superficial sensitivity of 1450 nm. This enhances the contextual value of hydration readings across skin layers, offering potential clinical and cosmetic use cases.

The study also explored population variability. Regression model analysis demonstrated that participants with darker skin tones tended to exhibit lower voltage readings, consistent with higher melanin content and known differences in optical tissue properties. These findings underline the importance of inclusive datasets in device calibration [150], [151].

In summary, the developed optical hydration sensor showed promising performance in detecting skin hydration changes during the moisturisation intervention. While these results demonstrate good agreement with reference methods, including the Corneometer, future work would aim to further quantify measurement reproducibility and assess performance across broader demographic and environmental conditions. These findings support the sensor’s applicability in wearable skincare technologies and open up opportunities for personalised health monitoring tools based on optical hydration sensing [152].

10.2 Effect of Dehydration on Optical Skin Hydration Status – A Case Study

10.2.1 Methodology and Materials

A case study was designed to investigate the effects of exercise-induced dehydration on body hydration status using a developed optical sensing device. The study aimed to explore the relationship between changes in body hydration as measured by the sensor, with a focus on understanding how dehydration affects the optical properties of the body cells.

The study involved five participants (Pt 1-5), all of whom provided informed consent after reviewing and signing the information sheet and risk assessment forms. Ethics approval was obtained from the City University of London's Ethics Committee, ensuring that the study adhered to ethical standards and participant safety. To ensure the reliability and validity of the results, participants were instructed to abstain from food and drink for 12 hours before the study. This fasting period was critical for stabilizing baseline hydration levels and eliminating potential variables that could influence the measurements.

At the start of the study, participants' height and baseline body weight in pounds (lbs) were recorded. Weight is a crucial parameter in dehydration studies, as it helps quantify fluid loss during exercise. The study protocol involved a 45-minute cycling session on an exercise bike (Wattbike) in a pre-heated room maintained at 25 degrees Celsius. The controlled temperature environment was chosen to enhance sweating and accelerate dehydration, allowing for more pronounced changes in body hydration to be measured. Measurements were taken at 15-minute intervals during the exercise session using both the developed optical sensor and a digital weight scale. The optical sensor, positioned on the participant’s forearm, recorded the hydration status while the scale measured the weight loss due to water loss via sweat.

After the completion of the cycling session, participants were given a one-hour rest period to allow their bodies to return to equilibrium and cool down to normal temperature. This rest period is essential to observe the body's natural recovery process and its effect on hydration. After this period, additional measurements were taken to assess the immediate post-exercise hydration status. Participants were then instructed to drink 500 ml of water, followed by a final set of measurements one hour later to determine the rehydration effects[153].

Figure 10.12: Wattbike and electronic scale used for exercise-induced dehydration study

The optical sensor measurements were taken from the forearm, with participants instructed to wipe away any sweat with a tissue before each reading. This precaution was necessary to reduce the occlusion effect caused by sweat accumulating on the sensor's glass surface, which could interfere with the readings. The sensor recorded data for 30 seconds at each time point, and the output was averaged to provide a consistent measurement.

As the developed sensor operates at multiple wavelengths within the near-infrared (NIR) spectrum, each with different penetration depths and sensitivities to water. The 1450 nm wavelength is known for its high sensitivity to water and its shallow penetration depth, making it particularly susceptible to surface-level hydration changes,

such as those caused by sweat. Conversely, the other wavelengths penetrate deeper into the dermis, providing more accurate measurements of internal hydration levels[14].

To isolate the effects of internal body hydration from surface-level sweat, the 1450 nm output was subtracted from the outputs of the other wavelengths. This subtraction technique is based on findings from literature that suggest the 1450 nm wavelength primarily reflects surface hydration, while deeper wavelengths like 975 nm, 1050 nm, and 1300 nm provide more accurate representations of internal hydration. This signal processing step significantly improved the validity of the results by mitigating the impact of sweat on the sensor's readings.

Sweat significantly affects the sensor's measurements due to its water content, which can block and absorb photons emitted by the LED, leading to a significant reduction in the real output voltage. This occlusion effect was evident in the initial results, where raw data showed a large decrease in voltage due to sweat accumulation. However, after applying the signal processing technique of subtracting the 1450 nm output, the results aligned more closely with expected hydration patterns, confirming the effectiveness of this approach in eliminating surface-level interference[120], [154].

The processed results were then plotted against the time points, and analysis was conducted using MATLAB. This allowed for a detailed examination of the relationship between dehydration, rehydration, and skin hydration status, providing insights into the dynamics of hydration as measured by the optical sensor under varying conditions. This methodology highlights the importance of accounting for surface-level effects in optical hydration measurements and demonstrates the potential of the developed sensor in monitoring hydration status during and after exercise.

10.2.2 Results

In this case study, the effect of dehydration on hydration status was monitored using an optical sensor, with measurements taken at baseline, every 15 minutes during exercise, one-hour post-exercise, and a further hour after the participants consumed a fixed amount of water. The optical sensor provided 30-second voltage readings at each time point, which were averaged to yield single voltage values, while weight measurements were recorded using a digital scale.

Consistently across all five participants, there was a noticeable decline in weight throughout the course of the study, reflecting fluid loss due to sweating during exercise. The decline in weight is a well-documented indicator of dehydration, as the body loses water through sweat to regulate temperature during physical activity. This weight loss directly correlates with reduced hydration levels in the body, which was also observed through changes in the optical sensor readings.

Table 10.1: Weight measurements in pounds (lbs) for each participant at the experimental time intervals

	0 min	**15 min**	**30 min**	**45 min**	**1 hr 45 min**
Pt 1	136	135.8	135.6	135.2	135
Pt 2	137.4	136.8	136.6	136.2	136
Pt 3	136.2	135.8	135.6	135.2	135
Pt 4	190.4	189.4	189.2	188.4	188
Pt 5	191.6	190.4	190.2	189.4	189

The averaged voltage outputs from the optical sensor were plotted against each time point, revealing several key trends. Notably, there was a steep decline in voltage between the baseline reading and the first 15-minute reading during exercise. This sharp decrease can be attributed to the transition from dry skin at baseline to moist skin due to sweating induced by exercise. Sweat, primarily composed of water, causes a significant reduction in voltage readings from the optical sensor. This occurs because the sensor detects water on the skin's surface, which absorbs the emitted light, leading to lower reflectance and consequently lower voltage readings[155].

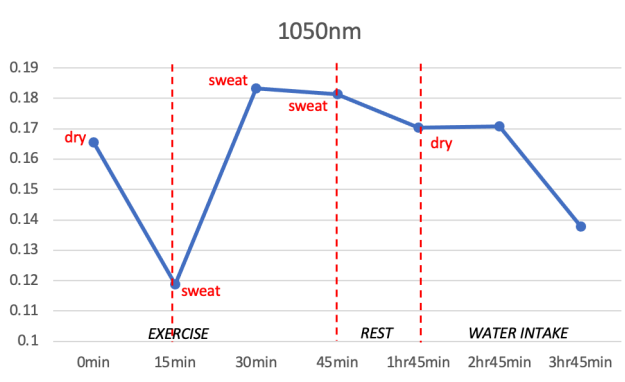
Following this initial decline, the subsequent voltage readings exhibited fluctuations, which can be attributed to the lack of baseline correction. Without correcting for the surface-level effects of sweat, the raw signal from the sensor included noise and variability that obscured the true changes in internal hydration status. To address this issue, signal processing was performed by subtracting the 1450 nm output from the outputs of the other wavelengths. The 1450 nm wavelength, known for its shallow penetration depth, primarily reflects surface hydration levels due to sweat. By subtracting this, the resulting signals from the other wavelengths, which penetrate deeper into the skin, provided a more accurate representation of internal hydration status. This approach effectively smoothed the signal, showing a more consistent increase in voltage as dehydration progressed, indicating that as cells lost water, the absorption decreased and reflectance increased[14].

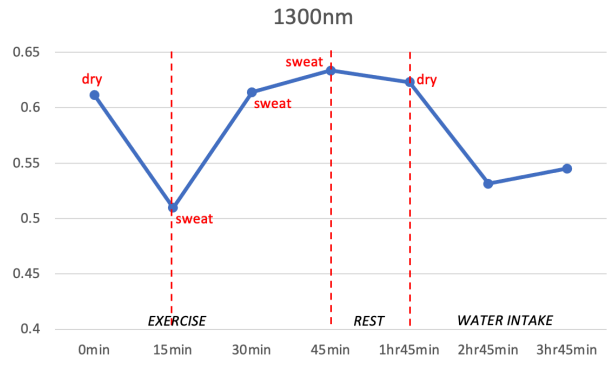
All five participants exhibited the same trend of increasing voltage in response to exercise-induced dehydration, demonstrating the sensor's capability to reliably detect changes in hydration status across different individuals. This consistency is promising for the development of a reliable tool for monitoring dehydration using optical methods.

Future work should focus on further refining the signal processing methods to address the significant impact of sweat on the measurements. One approach could involve developing an algorithm to automatically detect and correct for the steep decreases in voltage that occur due to sweat, perhaps by dynamically adjusting the baseline or incorporating real-time calibration methods. Another potential solution could be the integration of multi-wavelength analysis to distinguish between signals from different skin layers, improving the accuracy of hydration assessments despite the presence of surface moisture.

Pt 1


975nm
0.55
0.5
0.45
0.4
0.35
0.3
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min
3hr45min

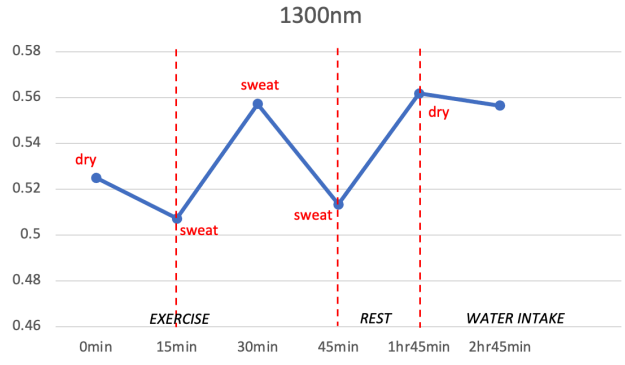

1050nm
0.19
0.18
0.17
0.16
0.15
0.14
0.13
0.12
0.11
0.1
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min
3hr45min


1300nm
0.65
0.6
0.55
0.5
0.45
0.4
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min
3hr45min

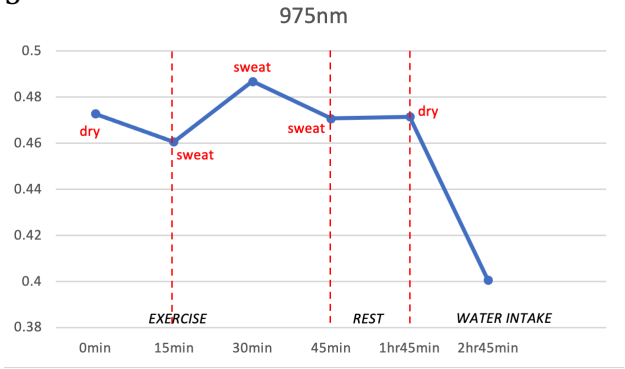
Pt 2


975nm
0.46
0.44
0.42
0.4
0.38
0.36
0.34
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min

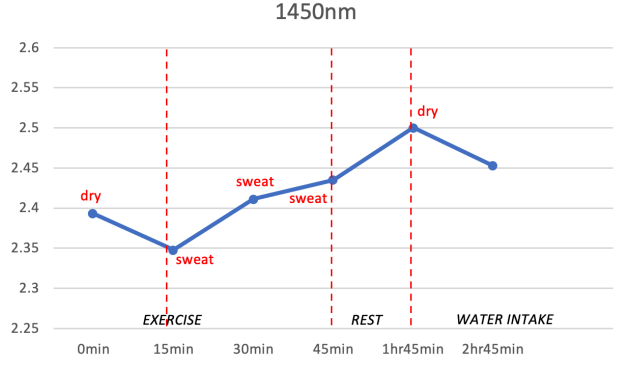

1050nm
0.16
0.155
0.15
0.145
0.14
0.135
0.13
0.125
0.12
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min


1300nm
0.58
0.56
0.54
0.52
0.5
0.48
0.46
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min

Pt 3


975nm
0.5
0.48
0.46
0.44
0.42
0.4
0.38
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min


1050nm
0.195
0.19
0.185
0.18
0.175
0.17
0.165
0.16
0.155
0.15
0.145
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min


1450nm
2.6
2.55
2.5
2.45
2.4
2.35
2.3
2.25
dry
sweat
sweat
sweat
dry
EXERCISE
REST
WATER INTAKE
0min
15min
30min
45min
1hr45min
2hr45min

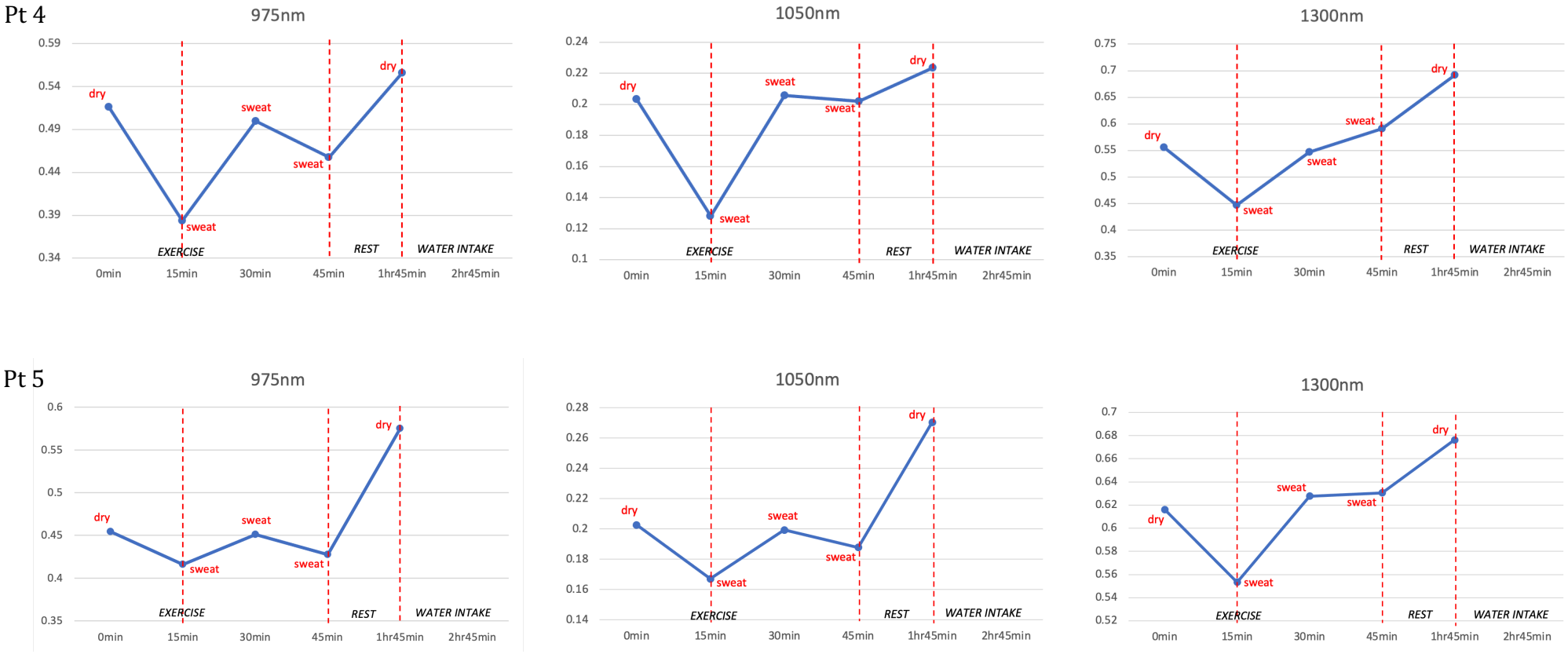
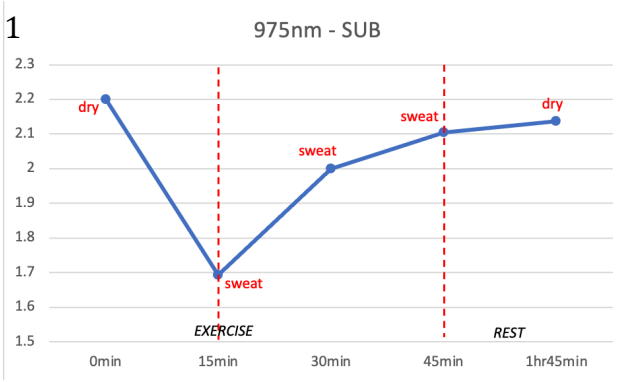
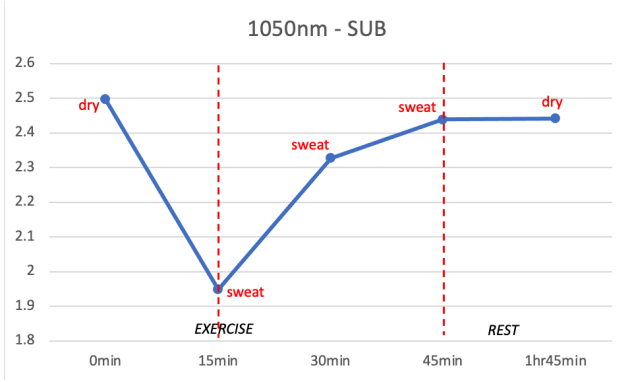
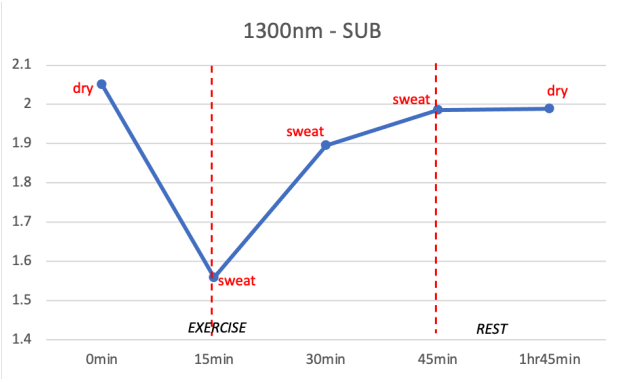


Figure 10.13: Optical sensor raw voltage output of 3 wavelengths across the 5 participants

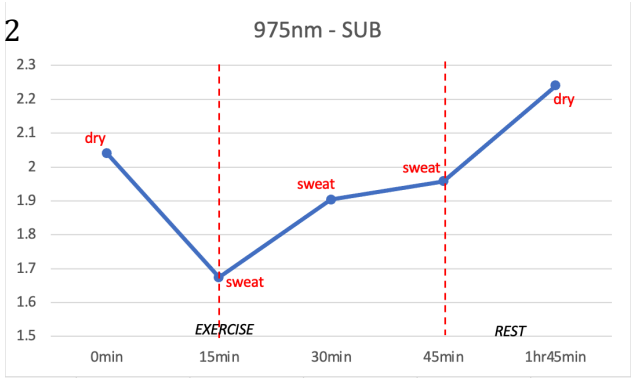
Pt 1


975nm - SUB
2.3
2.2
2.1
2
1.9
1.8
1.7
1.6
1.5
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min

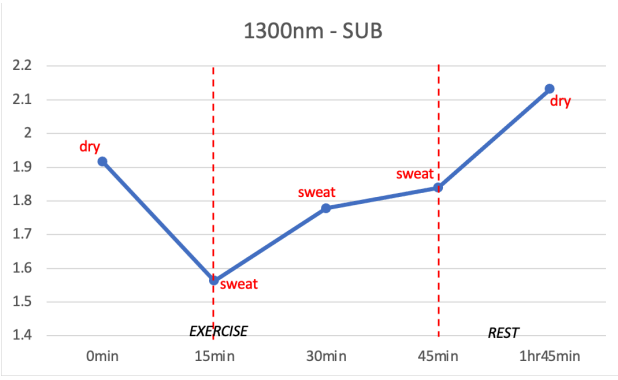

1050nm - SUB
2.6
2.5
2.4
2.3
2.2
2.1
2
1.9
1.8
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min


1300nm - SUB
2.1
2
1.9
1.8
1.7
1.6
1.5
1.4
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min

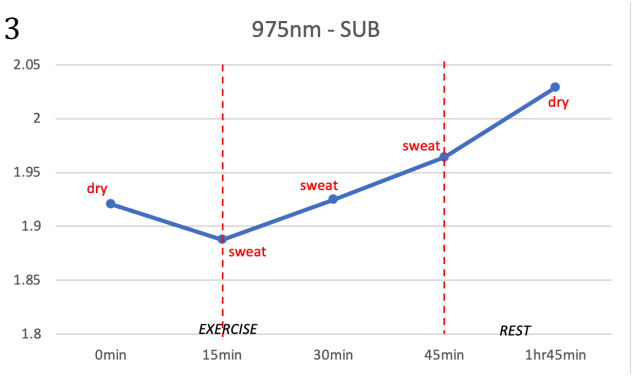
Pt 2


975nm - SUB
2.3
2.2
2.1
2
1.9
1.8
1.7
1.6
1.5
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min


1050nm - SUB
2.6
2.5
2.4
2.3
2.2
2.1
2
1.9
1.8
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min


1300nm - SUB
2.2
2.1
2
1.9
1.8
1.7
1.6
1.5
1.4
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min

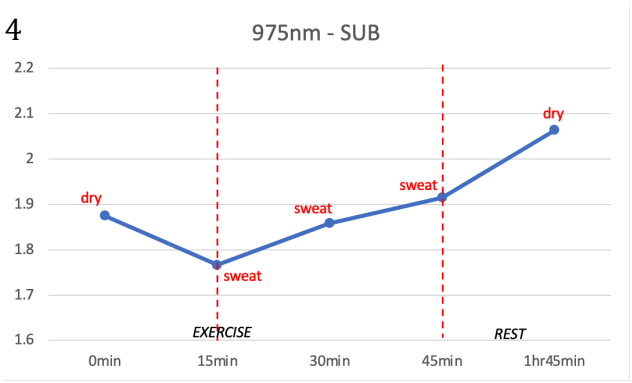
Pt 3

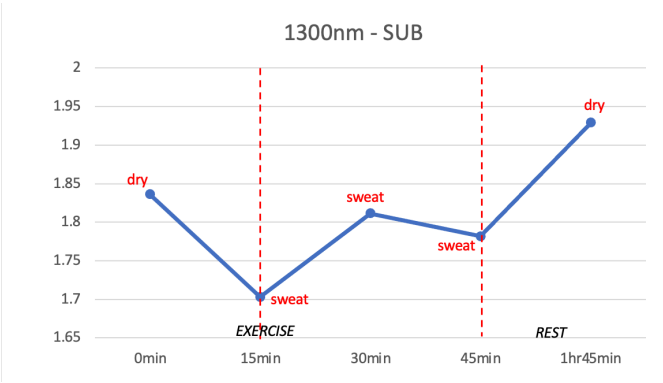

975nm - SUB
2.05
2
1.95
1.9
1.85
1.8
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min


1050nm - SUB
2.35
2.3
2.25
2.2
2.15
2.1
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min

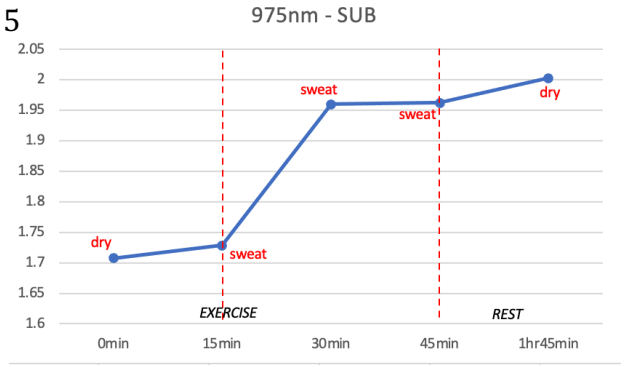

1300nm - SUB
1.9
1.88
1.86
1.84
1.82
1.8
1.78
1.76
1.74
dry
sweat
sweat
sweat
dry
EXERCISE
REST
0min
15min
30min
45min
1hr45min

Pt 4





Pt 5



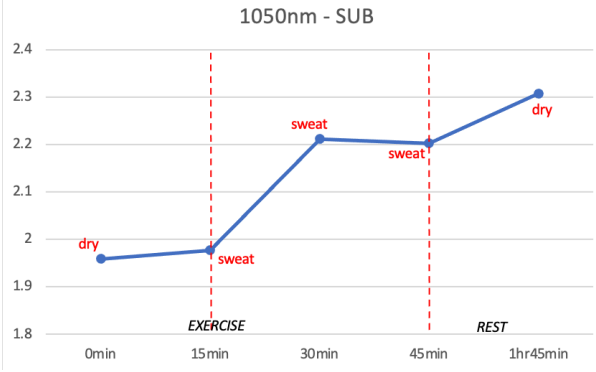


Figure 10.14: Optical sensor voltage output of 3 wavelengths across the 5 participants after subtraction of 1450 nm output to eliminate surface-level skin effects

To validate the optical sensor's capability in monitoring hydration levels, a comprehensive correlation analysis was conducted by comparing the optical sensor measurements, averaged across three wavelengths (975 nm, 1050 nm, and 1300 nm), with body weight measurements taken during an exercise-induced dehydration protocol. The optical sensor captures voltage outputs at these three distinct wavelengths, each sensitive to water absorption at varying skin depths. The 975 nm wavelength targets superficial hydration levels, while the 1050 nm and 1300 nm wavelengths provide deeper penetration, detecting water content further beneath the skin. For the purpose of correlating these outputs with body weight, the three measurements were averaged to produce a unified optical sensor output. This approach mitigates the variability inherent in individual wavelength readings, leading to a more stable and representative metric of overall hydration status. By combining the wavelengths, the influence of surface-level artifacts, such as sweat interference and skin-type variations, is reduced, ensuring that the output more accurately reflects internal hydration levels. This holistic measure balances the superficial and deeper hydration information, thereby enhancing the reliability of the sensor output in correlation analyses.

The analysis employed the use of linear regression, which quantifies the relationship between the sensor outputs and weight, providing key metrics such as the correlation coefficient (R-value) and the slope. The R-value indicates the strength and direction of the correlation, while the slope and intercept describe the sensor's sensitivity and baseline measurement level. Regression analysis is essential for assessing how well the sensor can predict changes in hydration based on weight fluctuations.

Body weight was chosen as the reference measurement because it is a widely accepted and non-invasive method to estimate hydration levels, particularly in short-term dehydration studies. As participants lose fluids through sweat during exercise, the corresponding decrease in body weight serves as a direct indicator of fluid loss and, therefore, dehydration status. Given the controlled nature of the exercise protocol, weight measurements provide a reliable baseline against which the optical sensor's performance can be validated. The results from the Bland-Altman plots and regression analysis across the five participants showed consistent patterns. The mean difference (bias) between weight and the combined optical sensor measurements remained close to zero for most participants, indicating that the sensor output aligns well with the weight reference. The limits of agreement varied across participants, with narrower limits for some, suggesting higher consistency in sensor readings for those individuals. Variations in these limits could be attributed to individual skin differences and external factors like sweat.

Table 10.2: Combined Linear Regression Analysis Results

	Participant	R_value	P_value	Slope	Intercept	Std_err
1	1.0	-0.30361274851126124	0.6194522699526592	-0.15082511627907427	22.52435355814014	0.27327049506690039
2	2.0	-0.3679684453954 99	0.5422866989753885	-0.14102105555555551	21.27782432222216	0.20574068253696434
3	3.0	-0.8147261114584979	0.09302541061192336	-0.07998201754386053	12.842370964912401	0.03286493250026188
4	4.0	-0.5384402884370838	0.3491777814290894	-0.06177922018348 6305	13.645944418960257	0.05582111607418441
5	5.0	-0.8742911302882358	0.0524830767025647	-0.1276547760210812	26.176439017127958	0.04091870247592254

The regression analysis revealed R-values for the participants ranging from -0.30 to -0.87, indicating a moderate to strong negative correlation between weight and optical sensor voltage. Higher absolute R-values, such as -0.87 for Participant 5, suggest a stronger inverse relationship, confirming that as weight decreases (indicating dehydration), the optical sensor voltage increases due to reduced water absorption. The lower R-values observed for some participants, such as -0.30, highlight the need for

further refinement in sensor calibration to account for individual variability. The p-values for participants with higher R-values approached significance, reinforcing the reliability of the sensor under these conditions. The slope values, varying between approximately -0.08 to -0.15, demonstrate the sensor's sensitivity to changes in body weight and hydration, validating the expected inverse relationship where dehydration corresponds to increased reflectance and thus higher sensor voltage outputs.

In summary, the developed optical sensor effectively detects hydration changes in the controlled, exercise-induced dehydration protocol. The sensor's ability to correlate its readings with weight measurements supports its potential application in monitoring hydration levels. However, variability in the R-values and agreement limits suggests that refining the sensor's calibration model, particularly to account for individual skin properties and external factors like sweat, would enhance its accuracy and reliability. Further analysis techniques, such as multivariate regression or machine learning models, could integrate additional factors like skin temperature and environmental conditions to improve the sensor's performance. Additionally, conducting larger-scale studies with diverse skin types and ethnicities would refine the calibration model, ensuring its applicability across a broader population.

10.2.3 Summary

This case study examined the impact of exercise-induced dehydration on body hydration using a custom-developed optical sensor. Five participants engaged in a controlled exercise routine designed to induce dehydration, with hydration levels measured at baseline, every 15 minutes during exercise, and twice post-exercise. The methodology aimed to monitor hydration changes by combining optical sensor outputs with weight loss measurements to infer fluid loss through sweat.

A consistent decrease in participant weight was observed, corresponding to water loss. This was reflected in the optical sensor readings, which showed a notable drop in voltage between the baseline and first 15-minute readings. The presence of sweat on the

skin initially caused lower reflectance values and decreased voltage output, due to strong water absorption at the 1450 nm wavelength.

To address this, signal processing was applied. The 1450 nm signal—most sensitive to superficial hydration—was subtracted from the outputs of other wavelengths (975 nm, 1050 nm, 1300 nm), allowing estimation of deeper tissue hydration. This correction helped smooth the data and reveal a consistent increase in reflectance over time, suggesting reduced hydration and confirming the sensor's responsiveness to dynamic hydration changes.

The sensor's performance was evaluated using correlation analysis between average voltage readings and participant weight. Averaging across the three deeper-penetrating wavelengths minimised variability due to surface interference such as sweat. The analysis through regression plots indicated moderate to strong negative correlations ($R = -0.30$ to -0.87), supporting the sensor's ability to track dehydration. However, variability between participants highlighted the influence of skin type, sweat production, and environmental conditions.

These results suggest that the optical hydration sensor can effectively monitor hydration changes during physical activity. Nonetheless, further optimisation is required to reduce inter-subject variability and account for confounding factors. Enhancing the calibration algorithm—potentially using multivariate regression or machine learning—could improve prediction accuracy. Additionally, integrating complementary metrics such as skin temperature or impedance may enable more comprehensive, individualised hydration profiling suitable for clinical, fitness, or military applications.

CHAPTER 11:

DISCUSSION, CONCLUSIONS & FUTURE WORK

11.1 Discussion

11.1.1 Hydration Measurement Techniques and the Need for Innovation

Hydration measurement is a critical parameter in both clinical and consumer health applications, impacting areas such as hospital monitoring for fluid management, athletic performance optimization, and the efficacy of cosmetic products. Accurate assessment of skin and body water content is essential, yet the current methods available often fall short of delivering reliable, non-invasive measurements that can adapt to a range of environments and individual variations.

Traditionally, electrical techniques, including capacitance and BIA, have been the standard for hydration measurement. However, these indirect methods are limited by the need for direct skin contact, susceptibility to external variables, and a narrow range of accuracy, particularly at extreme hydration levels. For example, BIA is highly sensitive to factors such as skin temperature, recent food or liquid intake, and the electrolyte balance within the body, which can all significantly skew results [22, 104]. This has driven the development of alternative methods, particularly optical techniques like NIRS, which offer a non-invasive approach to directly measure tissue water content [54, 89].

NIRS is particularly advantageous because it can penetrate several millimetres into the skin, enabling the assessment of water content in both the epidermal and dermal layers. The technique relies on the absorption of NIR light by water molecules at specific wavelengths, allowing for a direct and quantitative measurement of hydration. This method is less influenced by surface variables compared to electrical techniques, making it more reliable across a broader range of conditions. However, challenges such as the overlap of absorption spectra and the influence of other skin chromophores must be carefully managed through sophisticated signal processing and calibration techniques [89].

Despite the promise of NIRS, the field continues to face significant challenges that necessitate further innovation. For instance, the accurate calibration of NIRS devices remains a complex task, particularly when accounting for variations in skin pigmentation, age-related changes in skin structure, and the presence of other biochemical components

that may interfere with water absorption signals. The integration of advanced algorithms, including machine learning, could enhance the accuracy and adaptability of NIRS-based hydration measurement systems by continuously learning and adjusting to individual baseline variations and environmental changes.

Furthermore, the development of multimodal sensors that combine NIRS with other techniques, such as bioimpedance, could provide a more comprehensive assessment of hydration status. These hybrid systems would leverage the strengths of each modality, offering a more robust and accurate picture of hydration by cross-validating the data from different measurement approaches. For instance, while NIRS provides precise information on the direct water content in the superficial layers of the skin, bioimpedance could offer complementary data on intracellular and extracellular fluid distribution, addressing some of the limitations inherent in single-modality devices [17, 18, 138].

Another significant consideration is the application of these technologies in diverse populations and settings. As hydration needs and skin properties can vary greatly across different demographic groups, it is crucial to ensure that the developed technologies are both inclusive and universally applicable. This requires extensive testing across various skin types, age groups, and health conditions to validate the device's accuracy and reliability. Moreover, the impact of environmental factors, such as temperature and humidity, on both the measurement techniques and the physiological state of hydration, must be rigorously studied to develop devices that can provide accurate readings in real-world scenarios [13, 118, 120].

As the field progresses, there is also a growing interest in the potential for continuous monitoring of hydration status, particularly in the context of wearable technology. Continuous monitoring could provide real-time feedback, which is particularly valuable in scenarios where hydration levels can fluctuate rapidly, such as during intense physical activity or in patients with certain medical conditions. This would require the development of not only the hardware components but also the software algorithms capable of interpreting dynamic changes in hydration status and providing actionable insights to the user [22, 45].

In conclusion, while significant advances have been made in the field of hydration measurement, there remains a clear need for ongoing innovation. Future advancements will likely focus on improving the accuracy, reliability, and usability of these technologies, particularly in the context of wearable devices. By addressing the current limitations and exploring new approaches, such as multimodal sensing and machine learning integration, it will be possible to develop more effective tools for monitoring hydration status, with broad applications in both clinical and consumer health settings. The ultimate goal is to create a device that is not only accurate and reliable but also accessible and practical for everyday use, which will permit individuals to take control of their hydration and overall health.

11.1.2 Challenges with Traditional Hydration Measurement Techniques

Electrical methods, while widely used, present significant challenges in achieving consistent and accurate hydration measurements across a broad spectrum. Capacitance-based devices, such as the Corneometer®, are effective in measuring skin hydration but struggle with high hydration levels, leading to inaccuracies. This limitation is particularly critical in clinical settings, where precise hydration monitoring is necessary. The reliance on skin contact also introduces variability due to factors such as skin texture, the presence of lotions, and environmental conditions, thus requiring strict protocols for accurate measurements [144, 146].

BIA, although valuable for estimating overall body hydration, is influenced by external factors such as body temperature and recent physical activity. The tetrapolar electrode configuration has been proposed to mitigate some of these issues by separating the current injection and voltage measurement electrodes, thus reducing the impact of contact impedance. However, this approach increases the complexity and cost, potentially limiting its application, particularly in wearable devices where size, weight, and power consumption are critical considerations [134].

Bioelectrical techniques, particularly those involving constant current injection and resultant voltage measurement, also face challenges. Bipolar electrode configurations suffer from inaccuracies due to contact impedance, while tetrapolar

configurations, although more accurate, require more complex circuitry and can be less practical in portable applications. The exploration of materials such as nanomesh electrodes, which offer larger surface areas and therefore smaller impedance, presents a potential solution but adds another layer of complexity to device design [130].

Furthermore, these electrical methods often fail to provide accurate readings across a diverse population. Factors such as skin tone, which affects the dielectric properties of the skin, can significantly influence the measurements obtained through capacitance-based devices. For instance, darker skin tones with higher melanin content can alter the skin's electrical properties, leading to skewed hydration readings. This variation underscores the need for devices that can either compensate for such individual differences.

Another critical issue with traditional methods is their inability to distinguish between different types of water content in the skin. For example, capacitance methods measure the total water content but do not differentiate between free water (which is more indicative of hydration) and bound water (which is associated with structural components of the skin like collagen). This lack of specificity can lead to misleading conclusions about an individual's hydration status [49].

In clinical practice, these limitations are particularly problematic. For example, managing patients with conditions such as heart failure or chronic kidney disease requires precise fluid management. The inability to accurately assess hydration levels can lead to either fluid overload or dehydration, both of which can have severe consequences. In sports science, inaccurate hydration measurements can impair performance assessments and lead to inadequate hydration strategies, increasing the risk of heat-related illnesses in athletes.

Additionally, the invasive nature of some bioelectrical methods, which require direct contact with the skin, is a significant drawback in terms of patient comfort and compliance. Patients with sensitive skin or dermatological conditions may find these methods particularly uncomfortable, leading to reluctance in using these devices regularly, thus limiting the effectiveness of long-term hydration monitoring.

The complexities associated with the manufacturing and maintenance of these devices further these challenges. The need for frequent calibration to ensure accuracy,

the potential for electrode wear and tear, and the necessity for skilled personnel to operate and interpret the results contribute to the overall burden of using traditional hydration measurement techniques.

There is also a growing recognition of the limitations posed by the static nature of these measurements. Most traditional methods provide a snapshot of hydration status at a single point in time, failing to capture dynamic changes that may occur throughout the day or in response to various activities. This lack of continuous monitoring capability limits the ability to detect early signs of dehydration or fluid overload, which could otherwise be mitigated through timely interventions [45].

Given these significant challenges, there is a clear need for innovation in hydration measurement techniques. The development of non-invasive, real-time monitoring technologies that can accurately assess both skin and body hydration levels across diverse populations and varying conditions is imperative. Future devices must address the limitations of current methods, offering greater specificity, improved user comfort, and the ability to provide continuous, reliable data that can inform clinical decisions and optimize individual health management.

Emerging technologies such as multi-wavelength NIRS offer promising solutions. These methods can bypass some of the limitations of electrical techniques by directly measuring the optical properties of water in the skin, providing a more direct and potentially more accurate assessment of hydration levels. Moreover, integrating these technologies into wearable formats could revolutionize hydration monitoring, making it more accessible and practical for everyday use across various settings, from hospitals to athletic training environments [64].

In conclusion, while traditional hydration measurement techniques have laid the groundwork for understanding and managing hydration, their limitations necessitate the development of more advanced, reliable, and user-friendly technologies. Innovations in optical sensing and wearable technology represent the future of hydration monitoring, with the potential to significantly enhance health outcomes through better-informed fluid management strategies.

11.1.3 Challenges of Validating Hydration Measurement Techniques

Validating hydration monitoring techniques presents significant challenges due to the absence of a single, universally accepted gold standard for quantifying skin or tissue hydration non-invasively. Existing reference methods such as gravimetric analysis, TEWL, and capacitance-based measurements each assess different aspects of hydration—surface water loss, dielectric properties, or water content within the stratum corneum—but none provide a direct, absolute measurement of total tissue water content, especially across multiple skin depths. Furthermore, many of these techniques have limitations in precision, sensitivity to environmental factors like humidity and temperature, and dependency on skin condition, leading to inconsistent results across studies.

A further complication arises from the inherent variability in human physiology. Hydration levels can fluctuate due to anatomical site differences, vascularisation, age, skin type, or environmental exposure, and they can change rapidly in response to activities like exercise, diet, or skincare routines. These dynamic factors make it difficult to ensure that validation measurements are directly comparable across time points, participants, or devices, even within the same study. When validating new technologies, particularly optical or electrical methods that probe deeper skin layers, a mismatch in measurement depth or sampling volume between the reference and the test device can further undermine comparisons. For example, capacitance-based devices are sensitive to superficial hydration, whereas optical methods like NIR spectroscopy can probe deeper into the dermis where hydration dynamics are different.

Additionally, the lack of standardised testing protocols across studies makes replication and comparison difficult. Best practices for validation may include using controlled environmental conditions, repeated measurements across diverse populations, and applying dynamic hydration protocols (such as controlled dehydration and rehydration cycles) to test device responsiveness across physiologically meaningful changes. In research, some groups have addressed this challenge through the use of skin-mimicking phantoms with controlled water content to simulate known hydration states, allowing for repeatable and objective calibration and validation procedures. However, even these phantoms often fail to replicate the complex, multi-layered optical and

electrical properties of real skin, particularly its variable scattering and absorption characteristics.

Therefore, a robust validation strategy for hydration monitoring systems should ideally combine multiple reference methods, dynamic testing protocols, and standardised phantom models to build confidence in device accuracy. Moreover, there is a growing need in the field to establish international guidelines or standards for hydration measurement validation to improve consistency across research and commercial development, especially as wearable and non-invasive hydration sensors continue to advance.

11.1.4 The Role of NIRS in Hydration Measurement

NIRS offers a promising alternative to traditional methods by leveraging the specific absorption characteristics of water in the near-infrared spectrum. The O-H (hydroxyl) bonds in water molecules exhibit strong absorption at specific wavelengths, particularly around 970 nm, 1190 nm, 1450 nm, and 1940 nm. These wavelengths correspond to the vibrational transitions of the O-H bonds, making them highly sensitive indicators of water content in tissues. This molecular interaction is crucial for accurately assessing hydration, as the vibrational energy associated with these bonds directly correlates with water concentration, allowing NIRS to provide precise measurements of tissue hydration [14, 46, 47].

One of the significant advantages of NIRS is its ability to penetrate deeper into tissues compared to visible light, allowing for the non-invasive monitoring of hydration levels at various depths. This capability is particularly valuable for understanding the distribution of water within the skin layers, which is crucial for evaluating skin health and the effectiveness of topical treatments. The ability to assess hydration across different layers of the skin provides a more comprehensive understanding of overall hydration status. Additionally, NIRS's deeper penetration into tissues makes it useful in applications beyond dermatology, including monitoring cerebral hydration and muscle tissue water content, which are critical in both clinical settings and sports science [14, 89].

However, some challenges include the overlapping absorption bands in the NIR region, which can complicate the interpretation of spectra, particularly when other chromophores are present in the tissue. Chromophores like haemoglobin, fat, and collagen also absorb NIR light, potentially interfering with the accurate detection of water-specific signals. Advanced signal processing techniques, such as multivariate analysis, are often required to deconvolute these overlapping spectra and isolate the water-specific signals. Additionally, temperature variations can cause shifts in the absorption peaks, potentially leading to inaccuracies in hydration assessment. This sensitivity to temperature highlights the importance of controlled environmental conditions or real-time temperature correction algorithms in NIRS-based devices [55, 89].

Despite these challenges, the non-contact nature of NIRS and its ability to provide continuous, real-time data make it an attractive option for integration into wearable hydration monitoring devices. Wearable NIRS devices are being developed that combine the sensitivity of NIR light to water content with the ease and non-invasiveness required for continuous monitoring. These devices could revolutionize hydration assessment by providing real-time feedback, which is particularly beneficial in settings such as intensive care units, sports performance monitoring, and chronic disease management [45, 55].

Recent advancements in NIRS technology have focused on improving the specificity and sensitivity of the measurements by selecting wavelengths that correspond to the peaks in the water absorption spectra. For example, wavelengths near 1450 nm and 1950 nm have been suggested for their high sensitivity to alterations in water content on the skin surface, offering increased prediction accuracy compared to standard NIR spectrophotometers. This precision makes NIRS particularly suitable for clinical and cosmetic applications, where accurate and non-invasive hydration measurement is essential. Moreover, the development of compact, portable NIRS devices that can be easily integrated into wearable technologies has expanded its potential applications, making it accessible for everyday use in both professional and consumer settings [193].

As mentioned, further research is focusing on the integration of NIRS with other sensing modalities to enhance its diagnostic capabilities. For instance, combining NIRS with BIA or Raman spectroscopy could provide complementary data that enhances the

accuracy of hydration assessments. Additionally, the incorporation of machine learning algorithms into NIRS data processing could allow for more sophisticated analysis of hydration trends over time, enabling personalized hydration strategies for users based on their specific needs and conditions [82, 83].

The future of NIRS in hydration measurement looks promising, with ongoing developments aimed at overcoming the current challenges and expanding its applications. The potential to combine NIRS with other diagnostic tools and the advances in wearable technology could lead to more comprehensive and accurate hydration monitoring systems. These innovations are not only crucial for advancing clinical care but also for enhancing the quality of life in everyday health management.

11.1.5 Development and Validation of the Optical Hydration Sensor

The development of the optical hydration sensor described in this thesis aimed to address the limitations of existing hydration measurement techniques. By utilizing multiple wavelengths in the NIR spectrum—specifically 975 nm, 1050 nm, 1300 nm, and 1450 nm— and a single photodetector, the sensor provides a comprehensive assessment of both skin and body hydration levels. These wavelengths were carefully selected based on their specific absorption characteristics related to water content, allowing for a detailed analysis of hydration across different tissue layers. This multi-wavelength approach capitalizes on the varying penetration depths of NIR light, enabling the sensor to distinguish between superficial hydration and deeper, more intrinsic hydration levels.

One of the key innovations of this sensor is its ability to differentiate between surface-level hydration, such as sweat, and deeper hydration levels within the skin's dermis layer. This is achieved by subtracting the 1450 nm wavelength, which is highly sensitive to surface water, from the outputs of the other wavelengths. This method effectively isolates the signal related to internal hydration, improving the accuracy of the measurements in dynamic environments where surface factors like sweat could otherwise interfere. The ability to distinguish between these layers is particularly important in scenarios where external factors might influence the measurement, such as during physical activity or in varying humidity levels.

The validation of the sensor involved extensive in vitro and in vivo testing. Initial tests on porcine skin provided a controlled environment to refine the sensor's accuracy, while subsequent in vivo studies on human subjects assessed its performance in more variable conditions. These studies demonstrated the sensor's ability to detect changes in hydration levels accurately, even in the presence of confounding factors such as varying skin tones and environmental conditions. The results were compared with traditional hydration measurement techniques like BIA and TEWL, as well as a standard spectrophotometer, confirming the sensor's accuracy and reliability. This comprehensive validation process underscores the robustness of the developed sensor in both laboratory settings and real-world applications.

To enhance the sensor's ability to provide accurate and individualized hydration assessments, machine learning algorithms were developed and integrated. These algorithms were trained using a regression model that incorporated data from the in vivo study, which included a diverse range of participants. The model utilized input variables such as age, gender, ethnicity, skin type, and moisturization habits, along with the voltage outputs from the sensor's four wavelengths. By using these inputs, the model was able to predict the skin hydration index, providing a more personalized assessment.

The development of this machine learning model is a significant advancement in making the sensor's output more specific to individual users. The model's ability to account for individual differences, such as skin tone and texture, ensures that the sensor's hydration measurements are not only accurate but also tailored to the unique characteristics of each user. This level of customization is particularly important in achieving reliable hydration monitoring across diverse populations, addressing the variability introduced by factors like ethnicity and skin type. The use of machine learning also opens the door to continuous improvement of the sensor's accuracy, as the model can be periodically retrained with new data to refine its predictions further. This approach represents a step forward in the development of personalized health monitoring technologies, where sensor readings are adjusted in real-time to reflect the specific needs and characteristics of the user.

To further enhance the credibility of the optical hydration sensor, its design will also aim to integrate more advanced signal processing algorithms that account for

potential artifacts introduced by movement or environmental changes. This is crucial for ensuring the device's practicality in everyday use, particularly in wearable forms where user movement is inevitable. The incorporation of these machine learning techniques to calibrate and adjust the sensor's readings in real-time has been explored, offering the potential for personalized hydration monitoring that adapts to individual physiological differences.

The potential applications of this sensor extend beyond the initial scope of hydration monitoring. For instance, its ability to non-invasively assess tissue hydration could be invaluable in managing conditions like lymphedema, where fluid accumulation needs to be closely monitored, and eczema, where additional hydration and moisturization is required. Additionally, in sports science, this sensor could provide athletes with real-time feedback on their hydration status, helping to optimize performance and prevent dehydration-related issues. The medical field could also benefit, particularly in monitoring patients in critical care where maintaining optimal hydration is crucial, as well as monitoring for the elderly and neonates.

In terms of commercial viability, the development of this sensor also considers the integration of the technology into consumer-friendly devices. The trend towards wearable health technology is rapidly growing, with consumers increasingly interested in monitoring their physiological states, such as hydration, on-the-go. This sensor, with its compact design and non-invasive measurement technique, is well-positioned to meet this demand. Furthermore, it could be made affordable for widespread consumer use, potentially revolutionizing how hydration is monitored.

Looking forward, future iterations of the sensor could incorporate additional wavelengths or different types of optical sensing techniques to expand its diagnostic capabilities. For example, combining NIR spectroscopy with Raman spectroscopy could provide complementary data on other tissue properties, such as collagen content or blood oxygenation levels. Such enhancements could further solidify the sensor's role as a versatile tool in both medical diagnostics and personal health monitoring.

The successful development and validation of this multi-wavelength NIR optical hydration sensor represents a significant advancement in the field of hydration monitoring. By addressing the limitations of traditional techniques and offering a non-

invasive, accurate, and reliable method for assessing skin and body water content, this sensor holds the potential to greatly impact both clinical practice and consumer health technology. The research and development process detailed in this thesis lays a strong foundation for the continued evolution of this technology.

11.1.6 Discussion of Experimental Results

The experimental results offer a detailed evaluation of the sensor's performance across various testing environments. Chapter 8 focused on validating the sensor design through in silico experiments. This was crucial in establishing the fundamental accuracy and reliability of the sensor when measuring hydration in a controlled setting.

The Monte Carlo simulations conducted in this study provided valuable insight into photon behaviour within skin, particularly regarding photon penetration depth and pathlength at each of the selected wavelengths. These simulations revealed how different wavelengths interact with skin layers and allowed estimation of the average optical pathlengths through tissue, which are essential parameters for applying a modified form of the Beer-Lambert Law to biological media. Unlike the traditional Beer-Lambert Law, which assumes a direct path through a homogenous medium, the modified version accounts for scattering by incorporating an effective pathlength, which can be derived from simulation data based on the optical properties of skin and the source-detector geometry used in the device.

With the pathlength approximations obtained from the Monte Carlo models, it becomes feasible to use multi-wavelength measurements to estimate absolute water concentration in tissue. By selecting wavelengths that include strong water absorption peaks (such as 975 nm and 1450 nm) alongside those with minimal water absorption (such as 1050 nm and 1300 nm), the system can differentiate water-specific absorption from background tissue effects. In this approach, absorbance at each wavelength is calculated from the detected signal, and using the modified Beer-Lambert Law $A=\varepsilon \cdot c \cdot LA=\varepsilon \cdot c \cdot L$, where LL is the simulated pathlength and εε is the known extinction coefficient of water, the absolute water concentration cc can then be determined.

During desorption tests using porcine skin, which closely resembles human skin in optical properties, the sensor's output demonstrated a clear correlation with the gravimetric measurements. As the water content of the skin samples decreased due to evaporation, the sensor recorded an increase in reflectance, indicating its ability to detect hydration changes accurately. This was further corroborated by the sensor's performance when compared to a standard spectrophotometer, showing complementary trends in water loss and high sensitivity to the 1450 nm wavelength, a known water absorption band.

These findings emphasize the sensor's potential for use in non-invasive hydration monitoring, particularly where precise and continuous data are required, such as in clinical or athletic settings. The sensor's ability to accurately reflect changes in hydration levels, validated through in vitro experimentation, is particularly promising for its application in wearable technology, where real-time monitoring is essential.

Chapter 9 expanded on these findings by further validating the sensor through ex vivo experimentation using porcine skin. This chapter further confirmed the sensor's accuracy, with results showing a consistent inverse relationship between the reflectance measurements and skin hydration levels. The sensor's outputs were highly correlated with the gravimetric reference measurements, and the regression analysis demonstrated high R^2 values, indicating strong predictive accuracy. These results reinforce the sensor's capability to provide reliable hydration assessments, even in environments that more closely mimic real-world conditions.

In addition, the development of a multimodal sensor that combines optical and bioimpedance measurements was explored. This approach could enhance the accuracy of hydration monitoring by providing complementary data on both surface and deep hydration levels. The ex vivo results demonstrated that while bioimpedance measurements are more sensitive to electrode–skin contact, the tetrapolar configuration minimized these errors, making it a suitable addition to the optical sensor in a multimodal setup.

Chapter 10 involved the first in vivo application of the developed sensor on human participants. The study assessed the effectiveness of a moisturization regimen over seven days, comparing the sensor's readings against those from a Corneometer, the current

gold standard in skin hydration measurement. The results showed that the sensor's measurements closely matched those of the Corneometer, particularly at the 1450 nm wavelength, which is highly sensitive to water content. This indicates that the sensor is not only capable of measuring hydration accurately but also sensitive enough to detect changes induced by moisturizing products.

Another in vivo study examined the sensor's performance in detecting body hydration changes during exercise-induced dehydration. Here, the sensor successfully tracked the decrease in hydration levels as participants lost water through sweat. However, the presence of sweat initially caused some fluctuations in the readings, which were mitigated by applying advanced signal processing techniques. By isolating the 1450 nm wavelength output, which is more affected by surface hydration, from the other wavelengths, the sensor was able to provide a more accurate representation of internal hydration levels.

These results underscore the potential of the developed sensor for use in dynamic environments where hydration levels can fluctuate rapidly, such as during physical exercise. The ability to provide real-time, accurate hydration data in such settings is a significant advancement, particularly for applications in sports science and health monitoring.

In conclusion, the experimental results demonstrate the effectiveness of the developed optical hydration sensor across a variety of settings, from controlled laboratory conditions to real-world applications. The sensor's accuracy, reliability, and potential for integration into wearable devices highlight its promise for widespread application and adoption. Future research will focus on refining its capabilities, particularly in enhancing its robustness against environmental factors and further improving its accuracy. These advancements will ensure that the sensor can meet the diverse needs of its users, providing a reliable tool for continuous hydration monitoring.

In addition to these findings, the in vivo data also supported the sensor’s intended depth discrimination capabilities. The 1450 nm wavelength, characterised by its strong water absorption and relatively shallow penetration depth, demonstrated the greatest sensitivity to changes in superficial hydration levels. Across both moisturisation and dehydration studies, the 1450 nm wavelength consistently showed the most pronounced

changes in reflectance, particularly in response to sweat and moisturiser application. These changes are attributable to the high sensitivity of this wavelength to water present in the stratum corneum and upper dermal layers.

By comparison, the other wavelengths (975 nm, 1050 nm, and 1300 nm), which penetrate deeper into tissue, exhibited more gradual changes, reflecting their sensitivity to deeper hydration states. This contrast confirms the sensor's ability to differentiate between superficial and deeper hydration levels, based on the selective absorption properties of each wavelength. Furthermore, the ratio of reflectance changes at 1450 nm to those at the other wavelengths was consistent with theoretical expectations based on water absorption coefficients, thereby validating the system's depth-specific discrimination performance. This wavelength-dependent behaviour provides strong evidence that the developed sensor can distinguish hydration profiles across different skin depths, further reinforcing its value for layered tissue hydration analysis in real-world applications.

11.1.7 Discussion of the Developed Sensor's Accuracy and Reliability

The accuracy and reliability of the developed optical hydration sensor are fundamental to its potential for widespread adoption. The multi-wavelength approach, combined with advanced signal processing techniques, has proven effective in overcoming many challenges associated with traditional hydration measurement methods. The ability to differentiate between surface and deep hydration levels is noteworthy, addressing one of the most significant limitations of existing devices.

The sensor's performance was validated across different skin types, ethnicities and gender, demonstrating its adaptability and accuracy. This is particularly important for ensuring that the device can be used effectively in diverse populations and settings. Different skin types can present varying challenges for optical sensors due to differences in melanin content, skin thickness, and other physiological factors. For instance, darker skin tones, which have higher melanin levels, tend to absorb more light, which can affect the accuracy of optical measurements. The developed sensor's ability to maintain accuracy across these variables suggests its suitability for a wide range of users. This

adaptability is crucial for broadening the device's applicability in global healthcare settings, where patient demographics can vary widely.

The non-invasive nature of the sensor, coupled with its ability to provide real-time data, makes it an attractive option for continuous monitoring in both clinical and consumer health applications. Non-invasive monitoring is particularly advantageous in clinical settings, where patient comfort and compliance are important. For example, in the management of chronic conditions like kidney disease, where fluid balance needs to be carefully monitored, a non-invasive, reliable hydration sensor could significantly improve patient outcomes by allowing for more frequent and less intrusive assessments.

However, there are areas where further improvements could enhance the sensor's performance. For example, while the sensor's accuracy remained within acceptable limits in varying environmental conditions, further refinement could improve its sensitivity to extreme temperature variations. Temperature can affect the absorption spectra of water, leading to potential inaccuracies in hydration measurements. Therefore, incorporating temperature compensation algorithms could further refine the sensor's accuracy in diverse conditions.

Although machine learning has been incorporated with the sensor, augmenting these algorithms could enhance the sensor's adaptability, allowing it to learn from individual user data and adjust its readings accordingly. Machine learning could enable the sensor to develop more advanced personalized hydration profiles for each user, taking into account individual variations in skin properties, activity levels, and environmental exposures. This would be particularly valuable in developing personalized hydration monitoring systems, where the device could provide tailored recommendations based on the user's specific needs. For instance, athletes could benefit from personalized hydration strategies that optimize performance and recovery, while patients with chronic conditions could receive alerts when their hydration levels fall outside of a safe range.

Another area for potential improvement is the sensor's sensitivity and noise reduction. While the current sensor design effectively differentiates between surface and deep hydration, further enhancement of its signal-to-noise ratio and motion artefact effects could provide even more precise measurements. This could be achieved through

advancements in sensor materials or by refining the algorithms used to process the sensor data. High sensitivity is particularly important in detecting subtle changes in hydration status, which could be critical in early diagnosis and intervention.

Furthermore, the flexibility of the multi-wavelength approach opens the possibility of expanding the sensor's applications beyond hydration monitoring. For example, by targeting different wavelengths, the sensor could potentially be adapted to monitor other physiological parameters, such as oxygenation or glucose levels, making it a versatile tool in personalized health monitoring. This could significantly increase the sensor's value in both consumer and clinical markets, offering a multi-functional device that meets a wide range of health monitoring needs.

Finally, the long-term reliability and durability of the sensor in real-world conditions is a critical consideration that requires testing for its commercial viability. Continuous use in varying environmental conditions, including exposure to moisture, temperature fluctuations, and physical wear, could impact the sensor's performance over time. Therefore, future studies should focus on assessing the sensor's durability and identifying any potential degradation in accuracy or reliability with prolonged use. This could involve stress testing under simulated real-world conditions, as well as long-term user studies to evaluate the sensor's performance over extended periods.

In conclusion, the developed optical hydration sensor demonstrates significant potential as a reliable and accurate tool for hydration monitoring across diverse settings. Its multi-wavelength approach and advanced signal processing capabilities allow for precise differentiation between surface and deep hydration levels, making it a valuable addition to both clinical and consumer health monitoring technologies. However, continued refinement and validation are essential to address the challenges associated with environmental variability, personalization, and long-term reliability. By addressing these challenges, the sensor will be efficient for personalized health monitoring, offering users a comprehensive, non-invasive solution for maintaining optimal hydration.

11.1.8 Implications for Clinical and Consumer Applications

In clinical settings, the developed optical sensor could be used to monitor patients' hydration status continuously, providing early warning signs of fluid imbalances and allowing for timely intervention. This is particularly valuable for managing conditions like heart failure, where fluid balance is critical to patient outcomes.

In consumer health, the sensor could be integrated into wearable devices, providing continuous hydration monitoring for athletes, individuals in extreme environments, or those managing health conditions. The ability to monitor hydration in real-time could help prevent dehydration-related complications, optimize performance, and improve overall well-being.

The potential for this sensor to be integrated into wearable technology represents a significant advancement in the field of hydration monitoring. As technology continues to evolve, the sensor's capabilities could be further enhanced, making it an even more powerful tool for monitoring hydration and other health metrics.

The ability to continuously monitor hydration has broader implications beyond immediate health management. In preventive healthcare, such technology can play a crucial role in detecting early signs of dehydration before they escalate into more serious conditions. Dehydration can often go unnoticed until it leads to significant health issues such as kidney stones, urinary tract infections, or heatstroke. The real-time data provided by this sensor could allow for earlier interventions, reducing hospital admissions and improving patient outcomes.

Moreover, for individuals with chronic conditions like diabetes or kidney disease, where hydration levels can directly impact disease management, having a reliable and continuous monitoring system can significantly enhance quality of life. For instance, maintaining optimal hydration can help regulate blood glucose levels in diabetic patients, potentially reducing the risk of complications.

Another important implication is the integration of the sensor within broader smart health systems. With the rise of Internet of Things (IoT) in healthcare, the ability to collect, analyse, and act on hydration data in conjunction with other health metrics, such as heart rate, blood pressure, and activity levels, could lead to more holistic health management solutions. This integration could enable personalized health interventions,

where users receive tailored recommendations based on their unique health data, enhancing the effectiveness of health management strategies.

Despite the potential benefits, the integration of such sensors into wearable devices also brings challenges that need to be addressed for widespread adoption. User comfort, device durability, and data privacy are significant concerns that could impact user acceptance. Ensuring that the wearable device is comfortable for long-term use, especially during physical activity or in extreme environments, is crucial. Additionally, the device must be robust enough to withstand various environmental conditions without compromising accuracy.

Data privacy is another critical issue, particularly as health data is increasingly collected and stored digitally. Ensuring that the hydration data, alongside other sensitive health information, is securely stored and transmitted is essential to gaining user trust and complying with regulatory requirements. Future iterations of the sensor and associated wearable devices will need to include advanced encryption and data protection measures to address these concerns.

Looking forward, there is significant potential for further innovation in the field of hydration monitoring. Enhancements in sensor miniaturization, battery efficiency, and signal processing algorithms could lead to even more accurate and user-friendly devices. Moreover, advancements in material science could enable the development of sensors that are more sensitive, flexible, and less obtrusive, making them suitable for a wider range of applications.

In conclusion, the developed optical hydration sensor not only represents a significant technological advancement but also opens up new possibilities for health monitoring and management. Its potential applications in both clinical and consumer health are wide, offering opportunities for improved health outcomes through continuous, non-invasive hydration monitoring. As technology continues to evolve, the integration of this sensor into everyday health management practices could revolutionize the way hydration and health are monitored, leading to a future where personalized, real-time health insights are readily accessible to all.

11.2 Conclusions & Future Work

The research presented in this thesis marks a significant advancement in the field of hydration measurement through the development of a multi-wavelength NIR optical sensor. This sensor has been meticulously designed and validated to address the limitations of traditional hydration measurement techniques, offering a non-invasive, reliable, and accurate method for assessing both skin and body hydration levels. The integration of multiple wavelengths, specifically 975 nm, 1050 nm, 1300 nm, and 1450 nm, allows for a comprehensive analysis of hydration across different tissue layers, distinguishing between surface and deeper hydration levels. This capability is crucial for applications ranging from clinical diagnostics to consumer health monitoring, particularly in settings where real-time hydration monitoring is essential.

1. Calibration and Personalization

Future work will focus on developing more advanced calibration algorithms to further enhance the sensor's accuracy and reliability. The current developed approach involves a self-calibration technique, where the device establishes a baseline for each individual user, allowing it to account for personal variations in skin properties. However, a universal calibration model could be developed, enabling the sensor to provide accurate hydration readings without the need for individual calibration. Both approaches have their merits, and further research will be needed to determine the most effective method for widespread application.

Moreover, the development of personalized hydration monitoring systems, integrating machine learning algorithms, could enhance the sensor's adaptability. These systems can learn from user-specific data over time, adjusting the sensor's readings to provide tailored hydration insights. This would be particularly valuable in clinical settings, where individualized care is essential, and in consumer health, where personalized wellness strategies are increasingly sought after.

2. Extensive In Vivo Testing

Although preliminary in vivo experiments have demonstrated the sensor's potential, further extensive testing is necessary to validate its performance across more diverse populations and conditions. Future studies will involve testing the sensor on larger and more varied groups of participants, including individuals with different skin types, ages, and health conditions. These studies will not only assess the sensor's accuracy but also its reliability in detecting hydration changes under various environmental and physiological conditions.

In particular, studies could focus on specific patient groups, such as those with chronic conditions like kidney failure, diabetes, or eczema, where hydration monitoring plays a critical role in disease management. Understanding how the sensor performs in these contexts will be essential for its integration into clinical practice. Additionally, the sensor's application in sports science will be explored further, with studies designed to assess its effectiveness in monitoring hydration during physical activity and its impact on athletic performance.

3. Multi-modal Sensing and Combined Approaches

A promising avenue for future research involves exploring the combination of optical sensing with other measurement techniques, such as BIA. A multi-modal approach could leverage the strengths of each technique, providing a more robust and comprehensive assessment of hydration status. For instance, while NIRS offers precise measurements of water content in the superficial layers of the skin, BIA could provide complementary data on intracellular and extracellular fluid distribution. Combining these approaches could significantly enhance the accuracy and reliability of hydration monitoring in complex clinical scenarios.

4. Commercial Viability and Wearable Integration

The ultimate goal of this research is to develop a wearable hydration monitoring device that is both user-friendly and commercially viable. The current sensor design will undergo further refinement to meet the full needs of a wearable device. This includes enhancing battery life, ensuring durability in various environmental conditions, and

integrating wireless communication capabilities, such as Bluetooth, for seamless data transmission to external devices or applications.

The potential integration of this sensor into smart health systems is another exciting direction for future work. By combining hydration data with other health metrics, such as heart rate, blood pressure, and physical activity levels, the sensor could contribute to a more holistic approach to health monitoring. This integration could enable personalized health interventions, improving overall wellness and potentially reducing the risk of hydration-related health issues.

5. Broader Applications and Long-term Implications

Beyond hydration monitoring, the sensor's technology could be adapted for other applications, such as monitoring tissue oxygenation or glucose levels. The flexibility of the multi-wavelength approach makes it a versatile tool for various health monitoring needs. Additionally, the sensor's non-invasive nature and real-time data capabilities make it well-suited for continuous monitoring in a wide range of settings, from hospitals to athletic training facilities and everyday consumer use.

The broader implications of this technology extend to preventive healthcare, where continuous hydration monitoring could play a crucial role in early detection and intervention for conditions like dehydration, heatstroke, or fluid imbalances in chronic disease patients. As the technology evolves, its integration into wearable devices could revolutionize the way hydration and other health metrics are monitored, leading to better health and lifestyle outcomes.

6. Conclusion

In conclusion, this thesis has established the foundational development and early-stage validation of a novel multi-wavelength near-infrared (NIR) optical sensor for non-invasive skin and body hydration monitoring. The sensor design incorporates four NIR wavelengths—975 nm, 1050 nm, 1300 nm, and 1450 nm—chosen for their differential water absorption properties and varying tissue penetration depths. This spectral approach enables depth-specific hydration analysis, addressing a key limitation of many existing hydration measurement tools.

Across a series of in silico, ex vivo, and in vivo experiments, the sensor demonstrated consistent responsiveness to changes in hydration levels. While direct quantitative comparison to gold-standard methods such as the Corneometer confirmed a strong level of agreement in controlled trials, further testing is needed to evaluate its reproducibility and sensitivity across wider physiological and environmental conditions. These early results suggest that the sensor is capable of capturing hydration trends and detecting shifts in dermal water content, particularly in superficial layers.

However, it must be emphasised that the in vivo testing was limited in sample size, with some protocols based on data from a single participant. As such, caution should be exercised in generalising these findings to broader populations. For the device to be confidently used by end-users or considered for clinical or commercial deployment, further validation through large-scale, longitudinal studies will be essential. These studies should involve diverse demographic groups, repeated measures to assess reliability, and comparisons under different skin conditions, use environments, and hydration interventions.

The sensor's wearable form factor and integration into a real-time monitoring system underscore its potential in personalised health applications. Notably, the device performed consistently during moisturisation and dehydration protocols, indicating utility in dermatological, cosmetic, athletic, and potentially clinical contexts. Still, its efficacy must be explored further across more diverse skin types, pathologies, and settings to ensure robustness.

Importantly, this work highlights the need for continued improvement in signal processing, calibration, and algorithmic interpretation. Future directions include refining regression and classification models using machine learning, incorporating multi-modal sensing (e.g., impedance, temperature), and transitioning to a robust, low-power wearable platform suitable for mass deployment.

Ultimately, while this research provides a promising foundation, it represents a first step toward a more comprehensive solution for hydration monitoring. With continued development, validation, and refinement, the technology has the potential to enhance health and wellness outcomes globally through improved hydration assessment, early intervention, and real-time personalised feedback.

7. Future Work

To ensure the technology could be fully validated and prepared for regulatory approval, future work must focus on establishing a rigorous, standardised validation framework that demonstrates accuracy, repeatability, and clinical relevance. This would involve conducting large-scale clinical studies across diverse populations to assess performance under real-world conditions, including varying skin types, hydration states, and environmental exposures. Validation should incorporate further comparisons against multiple established reference techniques alongside the development and use of standardised skin phantoms with known, controllable water content to simulate different hydration levels. Additionally, protocols should include dynamic hydration challenges (such as controlled dehydration and rehydration phases) to evaluate device responsiveness over physiologically relevant changes. Pathlength-corrected absolute hydration quantification methods, such as those incorporating modified Beer-Lambert Law calculations supported by Monte Carlo modelling, could also be integrated to enhance the accuracy of results.

Beyond technical validation, gaining regulatory approval would also require a risk and safety assessment, including biocompatibility testing of materials, electrical safety certification, and verification of mechanical robustness for prolonged skin contact. Usability studies would be needed to assess human factors and ensure that the device can operate safely and effectively, with validated instructions for use. Comprehensive software validation, particularly for the calibration model and data handling processes, would also be essential. Privacy, cybersecurity, and data protection protocols would need to comply with regulations, particularly if data transmission or storage is involved.

Finally, the regulatory submission would require detailed documentation covering the device design, risk management, quality control processes, and clinical evidence supporting the device's safety and effectiveness in its intended use. These steps are crucial to demonstrate substantial equivalence to existing devices or justify the approval of a novel technology. Post-market surveillance strategies would also need to be in place to monitor long-term performance, user feedback, and ongoing safety. These efforts will ensure the developed hydration monitoring device meets the standards required for clinical deployment and regulatory approval.

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Appendices

Ethics ETH2223-2173: Miss Iman Gidado (Low risk)

Date Created	25 May 2023
Date Submitted	27 May 2023
Date forwarded to committee	28 May 2023
Academic Staff	Miss Iman Gidado
[redacted]	[redacted]
Category	Doctoral Researcher
Supervisor	Prof Panicos Kyriacou
Project	Multi-wavelength Optical Sensing and Monitoring of Dermal water content and Barrier Function
School	School of Science & Technology
Current status	Awaiting Mathematics & Engineering Research Ethics Committee meeting

Ethics application

Risks

R1) Does the project have funding?

Yes

R2) Does the project involve human participants?

Yes

R3) Will the researcher be located outside of the UK during the conduct of the research?

No

R4) Will any part of the project be carried out under the auspices of an external organisation, involve collaboration between institutions, or involve data collection at an external organisation? No

R5) Does your project involve access to, or use of, terrorist or extremist material that could be classified as security sensitive?

No

R6) Does the project involve the use of live animals?

No

R7) Does the project involve the use of animal tissue?

No

R8) Does the project involve accessing obscene materials?
No

R9) Does the project involve access to confidential business data (e.g. commercially sensitive data, trade secrets, minutes of internal meetings)?
No

R10) Does the project involve access to personal data (e.g. personnel or student records) not in the public domain? No

R11) Does the project involve deviation from standard or routine clinical practice, outside of current guidelines? No

R12) Will the project involve the potential for adverse impact on employment, social or financial standing?
No

R13) Will the project involve the potential for psychological distress, anxiety, humiliation or pain greater than that of normal life for the participant?
No

R15) Will the project involve research into illegal or criminal activity where there is a risk that the researcher will be placed in physical danger or in legal jeopardy?
No

R16) Will the project specifically recruit individuals who may be involved in illegal or criminal activity?
No

R17) Will the project involve engaging individuals who may be involved in terrorism, radicalisation, extremism or violent activity and other activity that falls within the CounterTerrorism and Security Act (2015)? No

Applicant & research team

T1) Principal Applicant

Name
Miss Iman Gidado

T2) Co-Applicant(s) at City

[redacted]

[redacted]

T5) Do any of the investigators have direct personal involvement in the organisations sponsoring or funding the research that may give rise to a possible conflict of interest?
No

T6) Will any of the investigators receive any personal benefits or incentives, including payment above normal salary, from undertaking the research or from the results of the research above those normally associated with scholarly activity?
No

T7) List anyone else involved in the project.

Project details

P1) Project title
Multi-wavelength Optical Sensing and Monitoring of Dermal water content and Barrier Function

P1.1) Short project title

P2) Provide a lay summary of the background and aims of the research, including the research questions (max 400 words).
Research conducted previously by the Research Centre for Biomedical Engineering at City have been successful at proving the potential of Near Infrared (NIR) Spectroscopy in providing more accurate and comprehensive analysis of skin health. Thus, this research aims to design, develop and validate a wearable multi-wavelength optical sensor for monitoring dermal water content and skin barrier function.

The presence of water in the skin is crucial for maintaining the properties and functions of the skin, in particular its outermost layer, known as the stratum corneum, which consists of a lipid barrier. External exposures can affect the skin's hydration levels and in turn, alter its mechanical and physical properties. Monitoring these alterations in the skin's water content can be applicable in clinical, cosmetic, athletic and personal settings. Many techniques measuring this parameter have been investigated, with electrical-based methods currently being widely used in commercial devices. Furthermore, the exploration of optical techniques to measure hydration is growing due to the outcomes observed through the penetration of light at differing levels.

The main objective of this study is to conduct an in vivo experiment on human participants for the analysis of the skin of the forearm using a fibre optic probe to investigate the properties of the Stratum Corneum in the Near-infrared region, and determine its water content and levels of hydration.

For the in vivo study, a fibre optic module connected to the spectrophotometer with an external probe will be used to shine light on the forearm of human participants, and generate similar spectra. Also, a commercial device widely employed to measure skin moisturisation through electrical impedance techniques may also be applied on our volunteers, as a means of analyzing our results and comparing them to current standards. Finally, there will also be measurements taken from a developed multi-wavelength optical sensor wearable that has the ability to provide skin hydration readings.

P4) Provide a summary and brief explanation of the research design, method, and data analysis.

An in vivo examination will be conducted using a fibre optic probe connected to a spectrophotometer as well as a developed multi-wavelength wearable sensor. For this part, the probe and wearable will be placed on the forearms of human volunteers, all with healthy skin conditions with 20 readings taken in total for each volunteer for averaging purposes. As light from the fibre optic probe strikes the skin, a portion of this light will be completely absorbed by the tissue chromophores, but a portion will become reflected and detected by a detector inserted into the probe. Consent will be obtained prior to any examination from adult individuals, along with a form where volunteers are asked about the condition of their skin.

P4.1) If relevant, please upload your research protocol.

P5) What do you consider are the ethical issues associated with conducting this research and how do you propose to address them?

An ethical issue involves confidentiality, as participants will be asked to complete a form asking for their age, ethnic background, and contact details, and give information regarding any skin condition they may suffer from. Age and Ethnic background will be requested because variations may occur in the results of the experiment which may be linked to these factors. This issue will be addressed by replacing individuals' names with numbers so that they cannot be identified at a later stage, and only the researcher will see the original forms filled, which would be kept safe by the principal investigator. Also, all data achieved from the experiments will only be available on a computer that is password protected.

P6) Project start date

The start date will be the date of approval.

P7) Anticipated project end date

30 Sept 2024

P8) Where will the research take place?

Research Centre for Biomedical Engineering at City, University of London

P10) Is this application or any part of this research project being submitted to another ethics committee, or has it previously been submitted to an ethics committee? No

Funding

F1) Funder

City, University of London

F2) Does the funder require external membership on the approving REC?

No

F3) Has the funding been approved?

Yes

F4) Value of grant

£ 0

Human participants: information and participation

The options for the following question are one or more of:
'Under 18'; 'Adults at risk'; 'Individuals aged 16 and over potentially without the capacity to consent'; 'None of the above'.

H1) Will persons from any of the following groups be participating in the project?
None of the above

H2) How many participants will be recruited?
10

H3) Explain how the sample size has been determined.
No formal power study was carried out. The selected number would allow for sufficient optical measurements to be made against reference measurements. The participants will be of varying ages and ethnic backgrounds but will have to have healthy skin that is free from any skin condition in the examination area of the body. This information will be obtained through a form that will be filled out by all volunteers asking them to state the required details prior to any testing. Information regarding the ethnic background and age is necessary because results may later show that there are variations in the results and the optical properties of the skin due to these factors. However, at the moment this is just a precaution, and only after the test is carried out, that it may be confirmed.

H4) What is the age group of the participants?
LowerUpper
18
H5) Please specify inclusion and exclusion criteria.
Participants will be suitable for the study if they do not suffer from any skin pathologies, especially their forearms. We will not include participants with cardiovascular, respiratory, neurological, or metabolic diseases, those taking any related medications, or are pregnant.

H6) What are the potential risks and burdens for research participants and how will you minimise them?
A risk assessment was successfully conducted and discussed with the university safety department and manager. The issue of electrical safety of the spectrophotometer has already been addressed by the manufacturer as those tests are performed before supplying the unit, and the fiber optic probe that would be connected is well insulated and only releases light, which poses no danger or risk to

the operator or any volunteers. In terms of health issues, the porcine meat will be in very small quantities, and once examined, it will be passed to the catering department for disposal or even put in disposable plastic bags and then in the bin. All surfaces and apparatus in contact with the meet will be cleaned and disinfected with methanol as well as soap and water.

H7) Will you specifically recruit pregnant women, women in labour, or women who have had a recent stillbirth or miscarriage (within the last 12 months)?
No

H8) Will you directly recruit any staff and/or students at City?

Staff
Students

H8.1) If you intend to contact staff/students directly for recruitment purpose, please upload a letter of approval from the respective School(s)/Department(s).

H8.2) Will you recruit students by virtue of their attendance on specific programmes or modules? No

H9) How are participants to be identified, approached and recruited, and by whom? The majority of the participants will come from SST research community (mainly PhD and postdoc students), but also UG students from the Biomedical Engineering programs as these students have a direct interest in the field. After gaining permission from the Dean, an email will be sent to students and staff in the department, highlighting the key points of the experiment and inviting individuals to take part. Those interested can then reply by email or a contact number included in the message.

H10) Please upload your participant information sheets and consent form, or if they are online (e.g. on Qualtrics) paste the link below.
H11) If appropriate, please upload a copy of the advertisement, including recruitment emails, flyers or letter.
H12) Describe the procedure that will be used when seeking and obtaining consent, including when consent will be obtained.
An email will be sent to both staff and students in the department asking them to participate. The participant information sheet will be attached to this email for individuals to view before signing up. Later, If they express a wish to participate then we will arrange for a fixed slot(depending on the participant's availability)in order to conduct the study. During this slot, the participants will be fully informed of the details of the study. Also, the explanatory statement will be given to the participants again to read and keep as a copy. Once they feel comfortable with all the information given to them and they are still interested in participating then the written consent form will be given to them to sign. The consent form will be obtained from either investigator. The participant will be informed that despite giving consent he/she will still be able able to withdraw from the study at any time if they wish.

H13) Are there any pressures that may make it difficult for participants to refuse to take part in the project?
No

H14) Is any part of the research being conducted with participants outside the UK?
No

Human participants: method

The options for the following question are one or more of:
'Invasive procedures (for example medical or surgical)'; 'Intrusive procedures (for example psychological or social)'; 'Potentially harmful procedures of any kind'; 'Drugs, placebos, or other substances administered to participants'; 'None of the above'.

M1) Will any of the following methods be involved in the project:
None of the above

M2) Does the project involve any deceptive research practices?
No

M3) Is there a possibility for over-research of participants?
No

M4) Please upload copies of any questionnaires, topic guides for interviews or focus groups, or equivalent research materials.
M5) Will participants be provided with the findings or outcomes of the project?
No

M6) If the research is intended to benefit the participants, third parties or the local community, please give details.

M7) Are you offering any incentives for participating?
No

M8) Does the research involve clinical trial or clinical intervention testing that does not require Health Research Authority or MHRA approval? No

M9) Will the project involve the collection of human tissue or other biological samples that does not fall under the Human Tissue Act (2004) that does not require Health Research Authority Research Ethics Service approval? No

M10) Will the project involve potentially sensitive topics, such as participants' sexual behaviour, their legal or political behaviour, their experience of violence?
No

M11) Will the project involve activities that may lead to 'labelling' either by the researcher (e.g. categorisation) or by the participant (e.g. 'I'm stupid', 'I'm not normal')?
No

Data

D1) Indicate which of the following you will be using to collect your data.
Physiological measurements

D2) How will the the privacy of the participants be protected?
De-identified samples or data

D3) Will the research involve use of direct quotes?
No

D5) Where/how do you intend to store your data?
Storage on encrypted device (e.g. laptop, hard drive, USB

Storage at City

D6) Will personal data collected be shared with other organisations?
No

D7) Will the data be accessed by people other than the named researcher, supervisors or examiners? No

D8) Is the data intended or required (e.g. by funding body) to be published for reuse or to be shared as part of longitudinal research or a different/wider research project now or in the future? No

D10) How long are you intending to keep the research data generated by the study? All electronic research data will be stored on my City OneDrive or SharePoint account as well as in the shared filed of RCBE provided by the university until September 2025.

D11) How long will personal data be stored or accessed after the study has ended? All electronic personal data will be stored on my City OneDrive or SharePoint account as well as in the shared filed of RCBE provided by the university until September 2025.

D12) How are you intending to destroy the personal data after this period?
In line with City's guidance, this data will be stored for 10 years and then deleted.

Health & safety

HS1) Are there any health and safety risks to the researchers over and above that of their normal working life?

No

HS3) Are there hazards associated with undertaking this project where a formal risk assessment would be required?
No

Attached files

Final protocol_in vivo.docx

InformationSheet.docx

ConsentForm.docx

MedicalHistoryQuestionaire.doc

RecruitmentFlyer.docx

MedicalHistoryQuestionaire.doc

Investigation into the effects of daily moisturization on skin hydration and barrier function and assessment of short-term daily application of a high lipid content cream

STUDY PROTOCOL

1. Study would be performed on volunteers with normal skin (no previous history of asthma, eczema or allergies). Volunteers with dry or oily skin may also be included at a later stage.
2. Panellists would be divided into two groups identified as follows:
3. Group 1 – consisting of individuals who moisturize their arms daily.
4. Group 2 – Those who do not at all moisturize their arms.
5. On day 0, all volunteers would be asked to wash as normal in the morning but refrain from using any products on their arms before entering the study to ensure minimal contamination of skin measurements from moisturizer residuals.
6. Then, all volunteers would be asked to complete a skin health questionnaire before baseline measurements would be taken from the volar aspect of both forearms using the Lambda 1050 equipped with a probe and the Corneometer, as well as with the developed multi-wavelength optical wearable sensor.
7. Then, a wet patch previously immersed in water would be applied on the right volar forearm of each volunteer. After 30mins, the wet patch would be removed and the skin gently patted dry with paper towels to remove excess water, and measurements with both instruments would be taken again. Once the skin is completely dry, a high lipid-content cream would be applied on the same test site for another 30mins before spectra and corneometer readings are taken again.
8. The same tests performed in step 7 would be repeated for the left volar forearm.
9. Then, the volunteers would be randomized to have either left or right upper volar forearm (same as previous test site) treated with the same high lipid content cream, 2/3 times daily for the following 7 days (Days 1-7).
10. The other upper forearm would serve as a symmetrical control.
11. The randomization code would be blinded to the investigator.
12. On day 8, participants will again be asked to wash in the morning but refrain from using any products on the test site before baseline measurements are obtained once more
13. Next, tests outlined previously in step 7 would be performed again on both arms of each participant.
14. Results from all instruments would be analysed, and the hydration and barrier function of individuals with normal skin who moisturize their arms daily would be compared to those who do not apply moisturizer. Results would also allow one to examine the effects on skin barrier function and hydration after a period of daily use of a high lipid content cream.

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Information Sheet

Title of study: Multi-Wavelength Optical Sensing and Monitoring of Dermal Water Content and Skin Hydration

We would like to invite you to take part in a research study. Before you decide whether you would like to take part it is important that you understand why the research is being done and what it would involve for you. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is not clear or if you would like more information.

What is the purpose of the study?

Research conducted previously by the Research Centre for Biomedical Engineering at City have been successful at proving the potential of Near Infrared (NIR) Spectroscopy in providing more accurate and comprehensive analysis of skin health. Thus, this research aims to design, develop and validate a wearable multi-wavelength optical sensor for monitoring dermal water content and skin barrier function.

The presence of water in the skin is crucial for maintaining the properties and functions of the skin, in particular its outermost layer, known as the stratum corneum, which consists of a lipid barrier. External exposures can affect the skin's hydration levels and in turn, alter its mechanical and physical properties. Monitoring these alterations in the skin's water content can be applicable in clinical, cosmetic, athletic and personal settings. Many techniques measuring this parameter have been investigated, with electrical-based methods currently being widely used in commercial devices. Furthermore, the exploration of optical techniques to measure hydration is growing due to the outcomes observed through the penetration of light at differing levels.

This research project aims to design an optical device using a multi-wavelength sensing approach to measure skin water content and skin hydration. This device will be developed and calibrated for in vitro and in vivo experiments to take place. The device will focus on the use of near-infrared spectroscopy (NIRS) for the detection of water bands within the skin and give an accurate measure of the absorption properties . Results from this project will help me further understand the optical properties of skin, particularly in the NIR region and assist in developing a novel technique in the future to detect moisture and possible early signs of skin disease.

I am seeking students and staff from City University with no history of skin disease (forearm and leg area) who would be willing to participate in this experiment.

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Why have I been invited?

You have been invited to participate in this study as a healthy volunteer at least 18 years old. We have decided to exclude people who are currently on any medication, and those with physical and mental health problems, and cardiac, liver or kidney abnormalities. Persons who consume more than 14 units of alcohol per week are also advised not to take part in the study. Please inform us before the commencement of this study if any of these apply to you. You can also refuse to share with us information you think are too personal and withdraw from the study at any time without giving a reason.

Do I have to take part?

Participation in this study is voluntary. You can withdraw at any stage. You can avoid answering to questions which you may feel intrusive or too personal.

It is up to you to decide whether or not to take part. If you decide to take part you are still free to withdraw at any time.

For students only: Your participation or withdrawal will not affect your grade during your studies at City University.

What will happen if I take part?

Initially, participants will be asked to complete a short form including their name, age and some contact details, ethnic background and questions regarding their skin history and use of certain products.

For confidentiality purposes, names on the completed forms will be replaced with codes for the remainder of the project. Also, all paper forms will be kept in a locked cupboard and all digital data will be saved on a password protected computer.

What do I have to do?

The actual experiment will involve placing a fibre optic probe, a Corneometer device and a developed multi-wavelength optical device on the lower arm area for about 15 minutes to obtain average spectra and hydration readings. This procedure is then repeated twice, once after applying a wet patch on the chosen skin site for 10 minutes, and once more 30 minutes after applying cream on the same area. The technology here is based on light, and poses no danger or risk to the operator or the participant, and a risk assessment has been taken into account and completed previously. Overall, examination will roughly last an hour to an hour and a half and require individuals to attend once only.

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What are the possible disadvantages and risks of taking part?

There are no likely risks in this experiment.

Expenses and Payments

Your participation has to be completely voluntary. There will not be monetary reward or any other kind of reward. Travel expenses will not be refunded.

What are the possible benefits of taking part?

This study will not have a direct benefit to you. However, we believe that it can contribute to the scientific community.

Will my taking part in the study be kept confidential?

You will be asked to sign a consent form. Your personal information will not be saved except the consent form which will be secured and locked in the biomedical research laboratory inside the university. Your measurement data will be saved anonymously as a number (e.g. “Participant 1”) into a computer. Only your age and sex will be saved. Your measurement data will only be accessed by the investigators and some selected researchers from the biomedical research laboratory at City University.

What will happen if I don’t want to carry on with the study?

In case of withdrawal before commencing the study, no data will be taken and saved. If you withdraw during the study, data may be considered to be saved for future analysis.

What will happen to results of the research study?

Results will be utilized for scientific publications and further research. Anonymity of the participants will be kept when publishing the results.

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The study is NOT considered a medical examination of your health status. Participation in this study should not be taken as substitution to regular medical examinations and your GP will not be notified.

What if there is a problem?

If you would like to complain about any aspect of the study, City University London has established a complaints procedure via the Secretary to the University's Senate Research Ethics Committee. To complain about the study, you need to phone 020 7040 3040. You can then ask to speak to the Secretary to Senate Research Ethics Committee and inform them that the name of the project is: Multi-Wavelength Optical Sensing and Monitoring of Dermal Water Content and Body Hydration

You could also write to the Secretary at:

[redacted]

This study has been approved by City University London Senate Research Ethics Committee.

Thank you for taking time to read this information sheet

[redacted]

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CONSENT FORM

Name of principal investigator/researcher: Iman Gidado / Meha Qassem

REC reference number: [redacted]

Title of study: Multi-Wavelength Optical Sensing and Monitoring of Dermal Water Content and Skin Hydration

		Please tick or initial box
1	I confirm that I have read and understood the participant information sheet for the above study. I have had the opportunity to consider the information and ask questions which have been answered satisfactorily.	
2	I understand that my participation is voluntary and that I am free to withdraw without giving a reason without being penalised or disadvantaged.	
3	I understand that I will be able to withdraw my data up to the time that all analysis is complete. If I decide to withdraw then no new data will be collected but data already collected will be retained.	
4	I understand that my data will be pseudo anonymised and agree to information being held and processed for the purpose of scientific publications	
5	I agree to City recording and processing this information about me. I understand that this information will be used only for the purpose(s) explained in the participant information and my consent is conditional on City complying with its duties and obligations under the General Data Protection Regulation (GDPR).	
6	I agree to take part in the above study.	

_____	_____	_____
Name of Participant	Signature	Date
_____	_____	_____
Name of Researcher	Signature	Date

When completed, 1 copy for participant; 1 copy for researcher file.

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MEDICAL AND SKIN HISTORY QUESTIONNAIRE

Name_____ Date _____

Contact Number _____

Email_____

Age_____ Gender _____

PERSONAL DATA:
Please circle yes or no for the following questions (**PLEASE NOTE:** all questions regarding your skin are focused on the forearms area)

Do you suntan: Yes or No

Do you use sunscreen: Yes or No

Cosmetic Surgery: Yes or No If Yes, when _____

Describe Procedure:
_____

Do you have any skin allergies: Yes or No
If yes, please explain: _____
_____ _____

How often do you use the following products on your arms (Please tick the appropriate box):

	Daily	Few times a week	Once a week	Few times a month	Once a month	Hardly ever	Never
Cleanser/So ap							
Moisturizer							

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Mask							
Toner							
Scrub							
Exfoliator							
Other							

Please circle no or yes to the following questions:

Do you use Retin-A gel or cream? Yes or No

If yes, what strength? _____

Have you ever used Hydroquinone (skin lightener)? Yes or No

Have you ever been on Acutance? Yes or No If yes, when? _____

Do you or have you previously suffered from Contact Dermatitis or Eczema, Asthma or food allergies? Yes or No

Are you aware of any other skin condition on your arms area? Yes or No

If yes, please specify _____
_____
_____

How would you describe your skin?

Normal Rough Dry Oily/Acne-prone

If any of the following affect the skin of your arms, or is applied to them, please circle yes and explain.

_____

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Skin Cancer	Yes	No
Photosensitive Disorder	Yes	No
Waxing	Yes	No
Electrolysis	Yes	No
Hypersensitivity to Skin Products	Yes	No
Skin Infections	Yes	No
Tanning within the last 6 weeks	Yes	No
Use of Acne Products/Drugs	Yes	No
Laser skin resurfacing	Yes	No
Chemical Peels	Yes	No
Photo Sensitizing substances	Yes	No
Laser work of any type	Yes	No

What is the complexion of your skin? (Non-exposed areas)

Reddish/Pale Fair Beige

Olive/Beige with a brown tint Brown Dark

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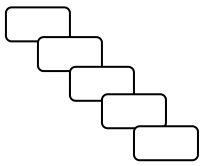
Brown/Black

Ethnic Background

Which group do you most identify with? (Please tick)

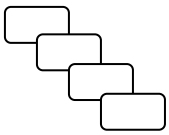
White

English
Irish
Scottish
Welsh
Any other White background
(specify if you wish) _____



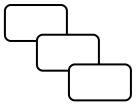
ASIAN

Bangladeshi
Indian
Pakistani
Any other Asian background
(specify if you wish) _____



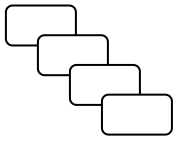
BLACK

African
Caribbean
Any other Black background
(specify if you wish) _____



Oriental

Chinese
Japanese
Korean
Any other oriental background
(specify if you wish) _____



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City, University of London
Northampton Square
London EC1V 0HB

MIXED ETHNIC BACKGROUND

Asian and White ☐
Black African and White ☐
Black Caribbean and White ☐
Any other Mixed ethnic background ☐
(specify if you wish) _____

ANY OTHER ETHNIC BACKGROUND

Any other ethnic background ☐

(Please specify) _____

School of Science and Technology

School of Science and Technology
City, University of London
Northampton Square
London EC1V 0HB

Research Centre for Biomedical Engineering City, University of London

VOLUNTEERS NEEDED FOR RESEARCH IN ASSESSMENT OF SKIN HYDRATION

The Research Centre of Biomedical Engineering (RCBE) is developing new sensors and technology in order to non-invasively analyse skin hydration using wearable devices. In order to optimise such techniques, we are conducting experiments on standard bodily and optical measurements within the laboratory before and after undertaking a moisturization routine. This study will investigate the effects of daily use of moisturizers on normal skin.

We are looking for volunteers to participate in a study that will involve measuring the skin hydration of volunteers and relating these measures to standard references using a spectrophotometer and standard Corneometer device. These measurements will be taken before and after a 1-week regime.

For this study, a pre-assessment questionnaire will be given to determine the suitability of participants. This includes providing information on medical and skin history. Participants should be healthy and a minimum age of 18 years. The study will target individuals with healthy skin and will not include participants with cardiovascular, respiratory, neurological, or metabolic diseases, those taking any related medications or are pregnant.

The first stage of the study would involve collecting baseline measurements. The experiment will involve an initial measurement using a safe and non-invasive technology used by dermatologists. This will last roughly an hour to an hour and half. After that you will be asked to apply a commonly known cream on the forearm for a period of 7 days, before coming in for a second and final measurement.

For more information about this study, or to volunteer for this study, please contact:

[redacted]

This study has been reviewed by, and received ethics clearance through the Senate Research Ethics Committee, City University London.

If you would like to complain about any aspect of the study, please contact the Secretary to the

[redacted]
School of Science and Technology
City, University of London
Northampton Square
London EC1V 0HB

Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]

Ethics ETH2223-2115: Miss Iman Gidado (Low risk)

Date Created	16 May 2023
Date Submitted	25 May 2023
Date forwarded to committee	26 May 2023
Academic Staff	Miss Iman Gidado
Student ID	[redacted]
Category	Doctoral Researcher
Supervisor	Prof Panicos Kyriacou
Project	Multi-wavelength Optical Sensing and Monitoring of Dermal water content and Barrier Function
School	School of Science & Technology
Current status	Awaiting Mathematics & Engineering Research Ethics Committee meeting

Ethics application

Risks

R1) Does the project have funding?

Yes

R2) Does the project involve human participants?

Yes

R3) Will the researcher be located outside of the UK during the conduct of the research?

No

R4) Will any part of the project be carried out under the auspices of an external organisation, involve collaboration between institutions, or involve data collection at an external organisation? No

R5) Does your project involve access to, or use of, terrorist or extremist material that could be classified as security sensitive?

No

R6) Does the project involve the use of live animals?

No

R7) Does the project involve the use of animal tissue?

No

R8) Does the project involve accessing obscene materials?

No

R9) Does the project involve access to confidential business data (e.g. commercially sensitive data, trade secrets, minutes of internal meetings)?
No

R10) Does the project involve access to personal data (e.g. personnel or student records) not in the public domain? No

R11) Does the project involve deviation from standard or routine clinical practice, outside of current guidelines? No

R12) Will the project involve the potential for adverse impact on employment, social or financial standing?
No

R13) Will the project involve the potential for psychological distress, anxiety, humiliation or pain greater than that of normal life for the participant?
No

R15) Will the project involve research into illegal or criminal activity where there is a risk that the researcher will be placed in physical danger or in legal jeopardy?
No

R16) Will the project specifically recruit individuals who may be involved in illegal or criminal activity?
No

R17) Will the project involve engaging individuals who may be involved in terrorism, radicalisation, extremism or violent activity and other activity that falls within the CounterTerrorism and Security Act (2015)? No

Applicant & research team

T1) Principal Applicant

Name
Miss Iman Gidado

T2) Co-Applicant(s) at City

T3) External Co-Applicant(s)
T4) Supervisor(s)
[redacted]

[redacted]

T5) Do any of the investigators have direct personal involvement in the organisations sponsoring or funding the research that may give rise to a possible conflict of interest?
No

T6) Will any of the investigators receive any personal benefits or incentives, including payment above normal salary, from undertaking the research or from the results of the research above those normally associated with scholarly activity?
No

T7) List anyone else involved in the project.

Project details

P1) Project title
Multi-wavelength Optical Sensing and Monitoring of Dermal water content and Barrier Function

P1.1) Short project title

P2) Provide a lay summary of the background and aims of the research, including the research questions (max 400 words).
Research conducted previously by the Research Centre for Biomedical Engineering at City have been successful at proving the potential of Near Infrared (NIR) Spectroscopy in providing more accurate and comprehensive analysis of skin health. Thus, this research aims to design, develop and validate a wearable multi-wavelength optical sensor for monitoring dermal water content and skin barrier function.

The presence of water in the skin is crucial for maintaining the properties and functions of the skin, in particular its outermost layer, known as the stratum corneum, which consists of a lipid barrier. External exposures can affect the skin's hydration levels and in turn, alter its mechanical and physical properties. Monitoring these alterations in the skin's water content can be applicable in clinical, cosmetic, athletic and personal settings. Many techniques measuring this parameter have been investigated, with electrical-based methods currently being widely used in commercial devices. Furthermore, the exploration of optical techniques to measure hydration is growing due to the outcomes observed through the penetration of light at differing levels.

The study aims to achieve the following:

- Conduct a small-scale study on human volunteers using an approved protocol of exercise-induced dehydration and rehydration.
- Obtain data on hydration measurements using a developed optical sensor, as well as spectroscopic data using a fiber optic probe connected to a spectrometer.
- Measurements will also be taken using a standard Corneometer device as a reference measure.
- This study will provide measurements of both skin and body hydration.

P4) Provide a summary and brief explanation of the research design, method, and data analysis.

The first stage of the study would involve collecting baseline measurements, where subjects would be placed in a well-ventilated area. Baseline data collection would include core and skin temperature, body mass, sweat analysis, and blood pressure. Baseline measurements of optical sensors would also be collected at this stage. This would include heart rate, ppg/pulse oximetry, developed optical hydration sensor, standard spectroscopic sensing systems, and hydration index with a Corneometer device.

A standard swab would be used per sampling for the collection of sweat on the skin surface, which will be placed within a solution to analyze the electrolyte concentration differences caused by dehydration inside a blood gas analyzer (ABL800 FLEX).

The exercise regime would begin once subjects are entered into a temperature-controlled room, regulated at a temperature of 25-28°C. There would be a period of 10 min for acclimatization after which subjects would begin to exercise on a bicycle ergometer with the speed and resistance adjusted to elicit metabolic heat production (60-90mins). Measurements of body mass would be acquired at intervals of 15 mins upon subjects momentarily stepping off the bicycle.

Subjects would be divided into two groups; in the first group, the fluid loss would be replaced by having subjects drink water administered in aliquots based on the body mass measurements taken every 15 min. In the second group, subjects would not consume any fluid and exercise continuously until an increase of core temperature of around ~1°C is observed (NOTE: this could be replaced with a time-match dehydrated condition).

At the end of the exercise regime, subjects would remain in the regulated room for 10mins and a second sweat sample would be collected together with body mass, heart rate, pgg and blood pressure to complement the data acquired from the optical sensors and Corneometer device. This will be followed by an extended resting period outside the temperature-controlled room and in a wellventilated area where subjects would be given measured amounts of fluid and left to rehydrate for a period of 60mins or until core temperature return to normal. Wearable sensors will continue to be worn at this stage, and a final round of intermittent measurements would be collected.

P4.1) If relevant, please upload your research protocol.

P5) What do you consider are the ethical issues associated with conducting this research and how do you propose to address them?

There are no major ethical issues associated with this protocol. The exercise on the bicycle ergometer will cause dehydration. This will be taking place in a well-ventilated room and water is given in aliquots as set within the study. The ingestion of water may increase urinary output. Washrooms will be accessible at all times should you need to use them.

P6) Project start date

The start date will be the date of approval.

P7) Anticipated project end date

30 Sept 2024

P8) Where will the research take place?

City, University of London

P10) Is this application or any part of this research project being submitted to another ethics committee, or has it previously been submitted to an ethics committee? No

Funding

F1) Funder

The PhD student leading this project is funded by SST. All resources needed for the project are funded by the Research Centre for Biomedical Engineering.

F2) Does the funder require external membership on the approving REC?

No

F3) Has the funding been approved?

Yes

F4) Value of grant

£ 0

Human participants: information and participation

The options for the following question are one or more of:
'Under 18'; 'Adults at risk'; 'Individuals aged 16 and over potentially without the capacity to consent'; 'None of the above'.

H1) Will persons from any of the following groups be participating in the project?

None of the above

H2) How many participants will be recruited?

10

H3) Explain how the sample size has been determined.

No formal power study was carried out. The number of participants has been chosen based on similar studies that have been published. The selected number would allow for sufficient optical measurements to be made against reference measurements.

The investigators will recruit a total of ten (10) healthy volunteers. The recruitment will initially start with 6 volunteers and, following analysis, then additional volunteers will be recruited if necessary.

H4) What is the age group of the participants?

LowerUpper

18

H5) Please specify inclusion and exclusion criteria.

A pre-assessment stage will be used to determine the suitability of participants. This includes age and weight criteria, as well as smoking and health status. The study will target individuals with fitness levels that are considered above average i.e. perform rigorous exercise more than 2 days a week) and will not include participants with cardiovascular, respiratory, neurological, or metabolic diseases, those taking any related medications, or are pregnant. Enrolled participants would be asked to refrain

from strenuous physical activity, and alcohol and caffeinate consumption for a period of 24hrs prior to the day of study.

The participant must understand this will involve:

- Refraining from strenuous physical activity, alcohol, and caffeine consumption for a period of 24 hrs prior to the day of the study.
- Taking part in strenuous physical activity during the study.
- Providing samples of blood and urine, along with other bodily measurements. • Dedicating my time for participation in the study.

Vigorous/strenuous exercise is defined as follows: the exercise regime would begin once subjects are entered into a temperature-controlled room, regulated at a temperature of 25-28°C. There would be a period of 10 min for acclimatization after which subjects would begin to exercise on a bicycle ergometer with the speed and resistance adjusted to elicit metabolic heat production (60-90mins).

H6) What are the potential risks and burdens for research participants and how will you minimise them?

The exercise on the bicycle ergometer will cause dehydration. This will be taking place in a wellventilated room and water is given in aliquots as set within the study. The ingestion of water may increase urinary output. Washrooms will be accessible at all times should you need to use them.

H7) Will you specifically recruit pregnant women, women in labour, or women who have had a recent stillbirth or miscarriage (within the last 12 months)?

No

H8) Will you directly recruit any staff and/or students at City?

Staff
Students

H8.1) If you intend to contact staff/students directly for recruitment purpose, please upload a letter of approval from the respective School(s)/Department(s).

H8.2) Will you recruit students by virtue of their attendance on specific programmes or modules? No

H9) How are participants to be identified, approached and recruited, and by whom?

Advertisements in the form of a recruitment flyer/poster will be affixed around the University premises, and emails, with the advertisement attached, will be sent to groups inviting interested persons to contact the investigator(s) about participating in the studies. The recruitment flyer has been attached below.

H10) Please upload your participant information sheets and consent form, or if they are online (e.g. on Qualtrics) paste the link below.

H11) If appropriate, please upload a copy of the advertisement, including recruitment emails, flyers or letter.

H12) Describe the procedure that will be used when seeking and obtaining consent, including when consent will be obtained.

Participants will be individuals who volunteer to take part in the study. They will be provided with a consent form to read and sign, along with a study information sheet outlining the entirety of the study and its potential risks. These will be provided the day before the study takes place to allow for time to restrain from activities mentioned within the consent form. The study can then take place 24 hours after this has been signed and confirmed, which will be completed within the same day.

Volunteers will be clearly informed that participation in the study is voluntary and that refusal to participate will not affect their role, study, etc within the University. No participants will be recruited if they lack capacity and if it is deemed that capacity is lost during the trial period then the participant will be removed from the trial and all information relating to them will be destroyed.

H13) Are there any pressures that may make it difficult for participants to refuse to take part in the project?

No

H14) Is any part of the research being conducted with participants outside the UK?

No

Human participants: method

The options for the following question are one or more of:
'Invasive procedures (for example medical or surgical)'; 'Intrusive procedures (for example psychological or social)'; 'Potentially harmful procedures of any kind'; 'Drugs, placebos, or other substances administered to participants'; 'None of the above'.

M1) Will any of the following methods be involved in the project:

None of the above

M2) Does the project involve any deceptive research practices?

No

M3) Is there a possibility for over-research of participants?

No

M4) Please upload copies of any questionnaires, topic guides for interviews or focus groups, or equivalent research materials.

M5) Will participants be provided with the findings or outcomes of the project?

No

M6) If the research is intended to benefit the participants, third parties or the local community, please give details.

M7) Are you offering any incentives for participating?

No

M8) Does the research involve clinical trial or clinical intervention testing that does not require Health Research Authority or MHRA approval? No

M9) Will the project involve the collection of human tissue or other biological samples that does not fall under the Human Tissue Act (2004) that does not require Health Research Authority Research Ethics Service approval? No

M10) Will the project involve potentially sensitive topics, such as participants' sexual behaviour, their legal or political behaviour, their experience of violence?
No

M11) Will the project involve activities that may lead to 'labelling' either by the researcher (e.g. categorisation) or by the participant (e.g. 'I'm stupid', 'I'm not normal')?
No

Data

D1) Indicate which of the following you will be using to collect your data.
Physiological measurements

D2) How will the the privacy of the participants be protected?
De-identified samples or data

D3) Will the research involve use of direct quotes?
No

D5) Where/how do you intend to store your data?
Storage on encrypted device (e.g. laptop, hard drive, USB
Storage at City

D6) Will personal data collected be shared with other organisations?
No

D7) Will the data be accessed by people other than the named researcher, supervisors or examiners?
No

D8) Is the data intended or required (e.g. by funding body) to be published for reuse or to be shared as part of longitudinal research or a different/wider research project now or in the future? No

D10) How long are you intending to keep the research data generated by the study? All electronic research data will be stored on my City OneDrive or SharePoint account as well as in the shared filed of RCBE provided by the university until September 2025.

D11) How long will personal data be stored or accessed after the study has ended? All electronic personal data will be stored on my City OneDrive or SharePoint account as well as in the shared filed of RCBE provided by the university until September 2025.

D12) How are you intending to destroy the personal data after this period?
In line with City's guidance, this data will be stored for 10 years and then deleted.

Health & safety

HS1) Are there any health and safety risks to the researchers over and above that of their normal working life? No

HS3) Are there hazards associated with undertaking this project where a formal risk assessment would be required?
No

Attached files

Information Sheet_Hydration_16May_smallscale.docx

ConsentForm_Hydration.docx

RecruitmentFlyer_Hydration_16May.docx

CITY UNIVERSITY LONDON

School of Science and Technology
Northampton Square
London EC1V 0HB
Tel. +44(0)20 7040 8131
Email: p.kyriacou@city.ac.uk

Information Sheet

Title of study: Multi-Wavelength Optical Sensing and Monitoring of Dermal Water Content and Body Hydration

We would like to invite you to take part in a research study. Before you decide whether you would like to take part it is important that you understand why the research is being done and what it would involve for you. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is not clear or if you would like more information.

What is the purpose of the study?

Research conducted previously by the Research Centre for Biomedical Engineering at City have been successful at proving the potential of Near Infrared (NIR) Spectroscopy in providing more accurate and comprehensive analysis of skin health. Thus, this research aims to design, develop and validate a wearable multi-wavelength optical sensor for monitoring dermal water content and skin barrier function.

The presence of water in the skin is crucial for maintaining the properties and functions of the skin, in particular its outermost layer, known as the stratum corneum, which consists of a lipid barrier. External exposures can affect the skin's hydration levels and in turn, alter its mechanical and physical properties. Monitoring these alterations in the skin's water content can be applicable in clinical, cosmetic, athletic and personal settings. Many techniques measuring this parameter have been investigated, with electrical-based methods currently being widely used in commercial devices. Furthermore, the exploration of optical techniques to measure hydration is growing due to the outcomes observed through the penetration of light at differing levels.
The thermoregulatory system adjusts a variety of physiological mechanisms to attain a balance between the heat produced within the body and the heat lost to the environment, through a combination of dry heat exchange and evaporative heat loss. Heat balance is easily disturbed during exposure to a warmer environment and with changes in metabolic heat production (due to physical activity).

Heat dissipation during exercise in hot environments is primarily accomplished via two major pathways: 1) convection via increased skin blood flow to the periphery and 2) evaporation through sweating. Evaporation of sweat acts as a mechanism for minimizing exercise-induced increase in core temperature, which in turn, causes changes in body water and electrolytes. During exercise in hot environments, fluid loss via sweating can exceed intake via drinking, which decreases total body water and results in dehydration. Hence, measurement of dehydration levels using core

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temperature and/or percentage change in bodyweight from pre to post exercise is considered a reliable method to estimating induced dehydration. For mild cases of dehydration, core temperature changes may be difficult to perceive and hence percentage bodyweight becomes the dominant variable.

This research project aims to design an optical device using a multi-wavelength sensing approach to measure skin water content and body hydration. This device will be developed and calibrated for in vitro and in vivo experiments to take place. The device will focus on the use of near-infrared spectroscopy (NIRS) for the detection of water bands within the skin and give an accurate measure of the absorption properties in effect to differing levels of hydration.

In this particular study, we are interested in measuring the skin and body hydration of volunteers and related these measures to both each other and to a standard reference of sweat electrolyte concentrations. Sweat samples and standard bodily measurements will be taken before and after a rigorous exercise regime, as well as the use of developed optical wearable sensors and a standard Corneometer device.

Why have I been invited?

You have been invited to participate in this study as a healthy volunteer at least 18 years old. We have decided to exclude people who are currently on any medication, and those with physical and mental health problems, and cardiac, liver or kidney abnormalities. Persons who consume more than 14 units of alcohol per week are also advised not to take part in the study. Please inform us before the commencement of this study if any of these apply to you. You can also refuse to share with us information you think are too personal and withdraw from the study at any time without giving a reason.

Do I have to take part?

Participation in this study is voluntary. You can withdraw at any stage. You can avoid answering to questions which you may feel intrusive or too personal.

It is up to you to decide whether or not to take part. If you decide to take part you are still free to withdraw at any time.

For students only: Your participation or withdrawal will not affect your grade during your studies at City University.

What will happen if I take part?

The pre-assessment stage would be used to determine the suitability of participants. This includes age and weight criteria, as well as smoking and health status. The study will target individuals with fitness levels that are considered above average i.e.

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perform exercise more than 2 days a week) and will not include participants with cardiovascular, respiratory, neurological, or metabolic diseases, those taking any related medications or are pregnant. Enrolled participants would be asked to refrain from strenuous physical activity, and alcohol and caffeinate consumption for a period of 24hrs prior to the day of study. You will then be prepared to take part in the study.

What do I have to do?

The first stage of the study would involve collecting baseline measurements, where subjects would be placed in a well-ventilated area. Baseline data collection would include core and skin temperature, body mass, sweat analysis, and blood pressure. Baseline measurements of optical sensors would be collected at this stage. This would include heart rate, ppg/pulse oximetry, and City's spectroscopic and hydration systems. Measurements from a standard Corneometer device will also be taken.

The exercise regime would begin once subjects are entered into a temperaturecontrolled room, regulated at a temperature of 25-28°C. There would be a period of 10 min for acclimatisation after which subjects would begin to exercise on a bicycle ergometer with the speed and resistance adjusted to elicit metabolic heat production (60-90mins). Measurements of body mass would be acquired at intervals of 15 mins upon subjects momentarily stepping off the bicycle.

Subjects would be divided into two groups; in the first group, fluid loss would be replaced by having subjects drink water administered in aliquots based upon the body mass measurements taken every 15 min. In the second group, subjects would not consume any fluid and exercise continuously until an increase of core temperature of around ~1°C is observed (**NOTE:** this could be replaced with a timematch dehydrated condition).

At the end of the exercise regime, subjects would remain in the regulated room for 10mins and a second sweat sample would be collected together with body mass, optical measurements and blood pressure to complement the data acquired from the hydration sensors. This will be followed by an extended resting period outside the temperature-controlled room and in a well-ventilated area where subjects would be given measured amounts of fluid and left to rehydrate for a period of 60mins or until core temperature return to normal. Wearable sensors will continue to be worn at this stage, and a final round of intermittent measurements (sweat, blood pressure, etc.) would be collected.

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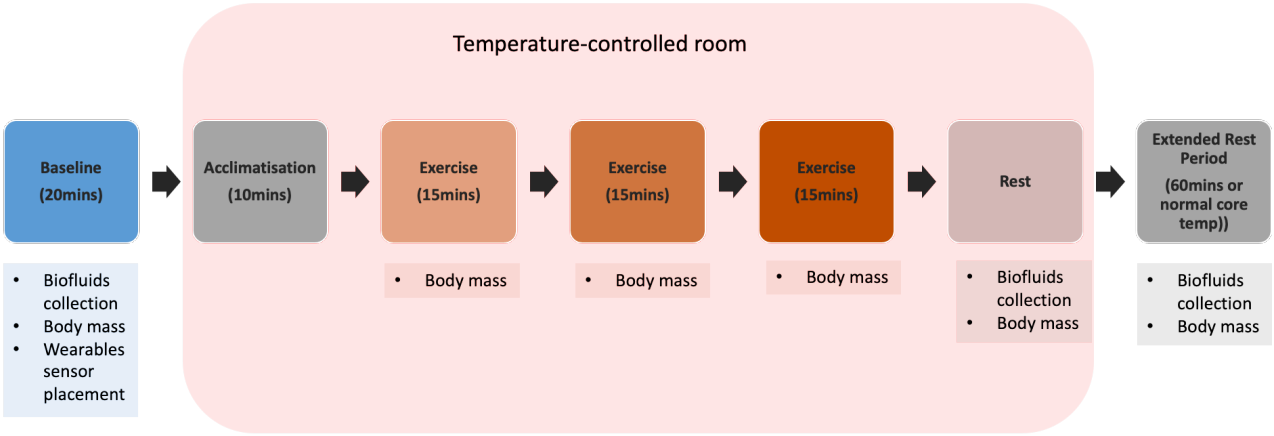

What are the possible disadvantages and risks of taking part?

The exercise on the bicycle ergometer will cause dehydration. This will be taking place in a well-ventilated room and water is given in aliquots as set within the study. The ingestion of water may increase urinary output. Washrooms will be accessible at all times should you need to use them.

Expenses and Payments

Your participation has to be completely voluntary. There will not be monetary reward or any other kind of reward. Travel expenses will not be refunded.

What are the possible benefits of taking part?

This study will not have a direct benefit to you. However, we believe that it can contribute to the scientific community.

Will my taking part in the study be kept confidential?

You will be asked to sign a consent form. Your personal information will not be saved except the consent form which will be secured and locked in the biomedical research laboratory inside the university. Your measurement data will be saved anonymously as a number (e.g. “Participant 1”) into a computer. Only your age and sex will be saved.

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Your measurement data will only be accessed by the investigators and some selected researchers from the biomedical research laboratory at City University.

What will happen if I don't want to carry on with the study?

In case of withdrawal before commencing the study, no data will be taken and saved. If you withdraw during the study, data may be considered to be saved for future analysis.

What will happen to results of the research study?

Results will be utilized for scientific publications and further research. Anonymity of the participants will be kept when publishing the results.

The study is NOT considered a medical examination of your health status. Participation in this study should not be taken as substitution to regular medical examinations and your GP will not be notified.

What if there is a problem?

If you would like to complain about any aspect of the study, City University London has established a complaints procedure via the Secretary to the University's Senate Research Ethics Committee. To complain about the study, you need to phone 020 7040 3040. You can then ask to speak to the Secretary to Senate Research Ethics Committee and inform them that the name of the project is: Multi-Wavelength Optical Sensing and Monitoring of Dermal Water Content and Body Hydration

You could also write to the Secretary at:

Annah Whyton
Research & Enterprise Office
City, University of London
Northampton Square

School of Science and Technology
Northampton Square
London EC1V 0HB
Tel. +44(0)20 7040 8131
Email: p.kyriacou@city.ac.uk

London, EC1V 0HB

Email: [redacted]

This study has been approved by City University London Senate Research Ethics Committee.

Thank you for taking time to read this information sheet

Further information and contact details:

- [redacted]
- [redacted]
- [redacted]

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CONSENT FORM

Name of principal investigator/researcher: Iman Gidado / Meha Qassem

REC reference number: ETH2223

Title of study: Multi-Wavelength Optical Sensing and Monitoring of Dermal Water Content and Body Hydration

		Please tick or initial box
1	I confirm that I have read and understood the participant information sheet for the above study. I have had the opportunity to consider the information and ask questions which have been answered satisfactorily.	
2	I understand that my participation is voluntary and that I am free to withdraw without giving a reason without being penalised or disadvantaged.	
3	I understand that I will be able to withdraw my data up to the time that all analysis is complete. If I decide to withdraw then no new data will be collected but data already collected will be retained.	
4	I understand that my data will be pseudo anonymised and agree to information being held and processed for the purpose of scientific publications	
5	I agree to City recording and processing this information about me. I understand that this information will be used only for the purpose(s) explained in the participant information and my consent is conditional on City complying with its duties and obligations under the General Data Protection Regulation (GDPR).	
6	I agree to take part in the above study.	

_____	_____	_____
Name of Participant	Signature	Date
_____	_____	_____
Name of Researcher	Signature	Date

When completed, 1 copy for participant; 1 copy for researcher file.

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Research Centre for Biomedical Engineering City University London

VOLUNTEERS NEEDED FOR RESEARCH IN ASSESSMENT OF BODY HYDRATION

The Research Centre of Biomedical Engineering (RCBE) is developing new sensors and technology in order to non-invasively analyse body hydration using wearable devices. In order to optimise such techniques, we are conducting experiments on sweat and standard bodily and optical measurements within the laboratory before and after undertaking a rigorous exercise routine.

Measurement of dehydration levels using core temperature and/or percentage change in bodyweight from pre to post exercise is considered a reliable method to estimating induced dehydration.

We are looking for volunteers to participate in a study that will involve measuring the body hydration of volunteers and relating these measures to both each other and to a standard reference of sweat electrolyte concentrations. Sweat samples and standard bodily measurements will be taken before and after a rigorous exercise regime, as well as the use of developed optical wearable sensors.

For this study, a pre-assessment stage will be used to determine the suitability of participants. This includes age and weight criteria, as well as smoking and health status. Participants should be healthy and a minimum age of 18 years. The study will target individuals with fitness levels that are considered above average i.e. perform exercise more than 2 days a week) and will not include participants with cardiovascular, respiratory, neurological, or metabolic diseases, those taking any related medications or are pregnant. Enrolled participants would be asked to refrain from strenuous physical activity, and alcohol and caffeinate consumption for a period of 24hrs prior to the day of study.

The first stage of the study would involve collecting baseline measurements.

For more information about this study, or to volunteer for this study, please contact:

[redacted]

This study has been reviewed by, and received ethics clearance through the Senate Research Ethics Committee, City University London.

If you would like to complain about any aspect of the study, please contact the Secretary to the

[redacted]
School of Science and Technology
City, University of London
Northampton Square
London EC1V 0HB

Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]	Volunteerstudy: [redacted]

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# HYDRATION INDEX PYTHON CODE

import matplotlib.pyplot as plt
import numpy as np
import pandas as pd
import pickle as pkl
import tempfile
import os
from flask import Flask, request, render_template

app = Flask(__name__)

# Load the trained models
def load_models():
    with open('label_encoder.pkl', 'rb') as file:
        label_encoder = pkl.load(file)
    with open('REAL_CALIBRATION_MODEL.pkl', 'rb') as file:
        calibration_model_trained = pkl.load(file)
    return label_encoder, calibration_model_trained

label_encoder, calibration_model_trained = load_models()

# Define wavelength weightings based on water absorption coefficients
WAVELENGTH_WEIGHTS = {
    '975nm': 0.025,
    '1050nm': 0.035,
    '1300nm': 0.20,
    '1450nm': 1.00  # Strongest absorber, used for correction
}

# Function to compute hydration index using weighted wavelengths
def calculate_hydration_index(V_975, V_1050, V_1300, V_1450):
    H_t = (WAVELENGTH_WEIGHTS['975nm'] * V_975 +
           WAVELENGTH_WEIGHTS['1050nm'] * V_1050 +
           WAVELENGTH_WEIGHTS['1300nm'] * V_1300 -
           WAVELENGTH_WEIGHTS['1450nm'] * V_1450)  # Subtract surface
hydration effects
    return H_t

@app.route('/', methods=['GET', 'POST'])
def index():
    if request.method == 'POST':
        ethnicity = request.form['ethnicity']
        age = int(request.form['age'])
        arm = request.form['arm']
        moisturised = 1 if request.form['moisturised'] == "Yes" else 0
        sex = request.form['sex']
        fitzpatrick_scale = int(request.form['fitzpatrick_scale'])
```

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        user_inputs_df = pd.DataFrame({
            'Fitzpatrick': [fitzpatrick_scale],
            'Ethnicity': [ethnicity],
            'Age': [age],
            'Sex': [sex],
            'Arm': [arm],
            'Moisture': [moisturised],
        })

        # Handle file uploads for baseline and sensor data
        baseline_file = request.files['baseline_data']
        sensor_file = request.files['sensor_data']

        if not baseline_file or not sensor_file:
            return "Error: Please upload both baseline and sensor data
files."

        temp_dir = tempfile.mkdtemp()
        baseline_file_path = os.path.join(temp_dir, "baseline_data.lvm")
        sensor_file_path = os.path.join(temp_dir, "sensor_data.lvm")

        baseline_file.save(baseline_file_path)
        sensor_file.save(sensor_file_path)

        # Function to process .lvm files
        def process_lvm_file(file_path):
            with open(file_path, "r") as file:
                lines = file.readlines()

            cleaned_data = [line.strip().split(",") for line in lines if
line.strip()]
            cleaned_data = [row[1:5] for row in cleaned_data if len(row)
>= 5]

            df = pd.DataFrame(cleaned_data, columns=["V_975", "V_1050",
"V_1300", "V_1450"])
            df = df.apply(pd.to_numeric, errors='coerce')  # Convert to
numeric values
            return df

        # Process uploaded baseline and sensor data
        baseline_df = process_lvm_file(baseline_file_path)
        sensor_df = process_lvm_file(sensor_file_path)

        if baseline_df.empty or sensor_df.empty:
            return "Error: One or both files are empty or incorrectly
formatted."
```

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        # Compute mean voltage values
        V_baseline = baseline_df.mean().to_dict()
        V_new = sensor_df.mean().to_dict()

        # Calculate hydration index using weighted wavelength values
        hydration_index = calculate_hydration_index(
            V_new["V_975"], V_new["V_1050"], V_new["V_1300"],
V_new["V_1450"]
        )

        # Define hydration thresholds
        threshold_upper = hydration_index * 1.1
        threshold_lower = hydration_index * 0.9

        # Determine hydration status
        if hydration_index > threshold_upper:
            message = "Alert: You are above your ideal hydration level."
        elif hydration_index < threshold_lower:
            message = "Alert: You are below your ideal hydration level."
        else:
            message = "Your hydration level is within the ideal range."

        return render_template('result.html',
hydration_level=hydration_index, message=message)

    return render_template('index.html')

if __name__ == '__main__':
    app.run(debug=True)
```