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"Development and Evaluation of a Novel Method for In-Vivo Determination of Human Skin Elasticity Based on Capacitive Measurement"

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Abstract

Since the medicine and research has great interest on the human skin and its physiological properties, parameter like thickness, moisture, toughness and the elasticity received more attention. Therefore, many types of measurement methods were implemented and are nowadays useable for specialists within a laboratory environment. The aim of this work was to find a handy method, usable for every person without paying attention to environmental influences. The application of a flexible capacitor on human skin, is the idea for the developed elasticity measurement method in this thesis. This principle works with the measurement of a certain capacity value regarding the change of length. Furthermore, the capacity output stands for a specific grade of elasticity of the skin area.

Some key figures of the measurement outputs have been investigated. The reproducibility of the values of one Hydrogel-Sensor, the flexible capacitor, has been verified as well as the sensor to sensor correlation. Influencing parameters on the capacity outputs like skin contact and geometry of the capacitor have been examined. Finally, the measurement outputs done with the Hydrogel-Sensor on test groups have been directly compared to the outputs of the "gold-standard", the *Cutometer*. The results of these investigations are promising but require further improvements. The production execution, used materials and the measurement procedure of the developed novel method need to be further developed and optimized.

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1. Introduction

1.1. Thesis conditions and co-worker

This work is inspired by a project of the department “Center for Health & Bioresources” of the AIT “Austrian Institute of Technology”, and realized in cooperation with the AIT, the LBI “Ludwig Boltzmann Institute for Experimental and Clinical Traumatology” and the TU Wien “Technical University of Vienna”. The content and idea of this thesis is to develop a novel simple method to measure the human skin elasticity and therefore make it usable for a wide range of users.

The department of “Center for Health & Bioresources” from the AIT brought up the topic of this thesis by pass on the question, from the open market to the research and development center. Inventions and solutions to simplify the detection of diseases and to help older people to be secure and medically supervised is the focus of this department. That is why the technical development part of this work was supported by the AIT. Further, solutions regarding the chemistry and biology were worked out with the help of the laboratory of the LBI. The laboratory has great experience with chemical experimental work within different topics, as for example tissue regeneration with the aim to create and improve diagnostic and therapeutic treatment methods. The combination of these different influences helped to create this thesis.

1.2. Motivation

Skin represents the largest organ of the human body and fulfills several different tasks such as protection, thermal regulation and physical, mechanical and chemical barrier functions. [1] Another very important task of the human skin is the visual effect, that is responsible for our personal “picture” towards other people and ourselves. Especially in the face region, the 26 muscles, responsible for the mimic, placed on the boundary of the epidermis moving the facial skin. Communication between two people would be very difficult because, as well known, most of the conversations takes place through non-verbal communication. The mimics of each individual are characteristic and unique and helps to evaluate the mood of the opposite. All this is possible because of the collaboration between the facial skin and the underlying muscles. [2] Furthermore, it is a significant indicator for the age of every human being. The stage of wrinkles, the tautness and the

pigmentation of the skin gives us important signals that help to evaluate the age of the person in front of us and adapt our behavior as demanded by our social systems. Wrinkles occur through all three layers of the skin. Those three layers are beginning at the top, the epidermis, dermis or stratum corneum and the subcutis. [3] The interaction between those layers is essential for fulfilling the diverse functions of the skin. Especially regarding the mechanical requirements, a consistent and stable collaboration is needed. One layer by itself wouldn't stand the different mechanical loads like shear forces, pressure or tension. To make this cooperation measurable and predictable, the concept of the skin elasticity was introduced to the medicine and dermatology. It provides values to assign different skins to different levels of elasticity, in meaning of the ability of the skin layers to withstand mechanical loads. This allows to measure a possible impact on the skin elasticity by medical treatments, in the field of dermatology. Some typical and common measurement methods including the specific values are given in Table 1-1.

Type of modulus	Measurement method and position	max. value
tangent modulus	extension 1%/s, forearm	3,0 MPa
	suction	0,8 MPa
	torsion	1,1 MPa
dynamic modulus		600 MPa
shear modulus	volar forearm	1-2 kPa

Table 1-1: Standard values of the human skin with different measurement methods [4]

The current measurement methods are all made for the use within a controlled laboratory environment. In addition, the handling of the measurements need to be done by a qualified person with a specific measurement process. Besides the impact of the measuring technique on the outputs for the modules, parameters like age, sex and hydration, also have an impact on the results. Many research papers showed, that ageing increases the young's modulus. [5] This can be explained by the fact that ageing reduces hydration which lowers extensibility. [1] Elasticity is an important parameter for plastic surgery as well as skin replacement within emergency medicine. Since skin elasticity plays an important role in other medical sciences, too, it is of high interest to have the possibility to measure the skin elasticity quick, easy and without special skills. It would also open

the possibility to use the transportable and simple method at home, in order to control the success of their medical treatment. The elasticity of human skin is a big issue not only for the society but also for example the medical treatment, plastic surgery and the research field for tissue biomechanics.

1.3. Structure of this thesis

At the beginning, the hypothesis considering the requirements on the developed measurement method are defined. Furthermore, the scheme and idea of the novel method are described and the extent of this work is elaborated. For a better understanding, the background knowledge regarding the physiology of the human skin and physical definition of the concept of skin elasticity is described in the next chapter. Additionally, the most common measurement methods to measure the elasticity of human skin are listed. In order to get an impression of the usability of these methods, the output quantities and the applications are specified. The physical connections and principles necessary for the comprehensibility of the measurement procedure are outlined in the following section. Furthermore, the components and their production process of the measurement instruments are described. Following, two measurement setups are pictured. The first description is for the measurements of the validation, whereas the second setup is done for the measurements on test persons. The results of these setups are all listed within the next chapter. Starting with the description of the production process and the finding process of the appropriate geometry of the measurement sensor. The results of the validating measurements and the comparable outputs of the test series with the novel method and the "gold-standard" the Cutometer, are listed at the end of this section. To close this report, the discussion and conclusion are described in detail at last.

2. Hypothesis

Some measurement methods are already on the market and used in the research field. However, only a few methods emerged as a practical method. Therefore, the method examined in this thesis requires several important basic functions. Because the novel method is still in the first development phase, the following hypotheses are encountered. These hypotheses refer, on the one hand to basic physical functions and connections and on the other hand to the correlation of the new developed measurement method, using flexible capacitors, and the "gold-standard" within this research field, the *Cutometer* of Courage&Khazaka.

Hypothesis **H0**:	It is possible to examine a method to measure the skin elasticity with determinable reproducibility, linearity and logic physical parameters.

H0: Each Hydrogel-Sensor needs to be consistent to a different sensor, regardless the production batch of the individual components and the environmental conditions. To ensure the further studies about the usability of the novel method, the hypothesis of a good linearity and reproducibility of the measurement results is provided and must be investigated. Furthermore, the consistence of the material parameters with the literature must be verified.

Hypothesis **H1**:	$C \sim \Delta l \sim E_{modulus}$ of the skin

H1: The physical parameter, called the Young´s modulus or elastic modulus describes the stiffness of a solid material and is therefore a common reference for the elastic properties within the material science. In the case of an applied force to stretch or contract a material, the Young´s Modulus is directly proportional to the initial length and indirectly proportional to the amount by which the length of the object, consisting of the examined

material, changes. Usually, the elastic parameter for human skin to categorize is the Young's Modulus, since the comparability with other materials is ensured. For this reason, the second hypothesis refers to the correlation between the measurement values of this novel method, which is a capacity quantity, and the resulting length and its change after applying a displacement to the skin.

Hypothesis **H2**:	There is a correlation between measurement outputs done with the Hydrogel Sensor and the outputs of the *Cutometer* (Courage&Khazaka), during examinations of the exact same test persons with constant environmental conditions.

H2: The Hydrogel-Sensor and its measurement concept needs to be comparable to the actual "gold-standard" among the measurement techniques. Therefore, the measurement results of the examined new method and the results from the *Cutometer* (Courage&Khazaka), done with the exact same environmental conditions, are compared in four different subject groups. Thus, the correlation between the Hydrogel-Sensor and the *Cutometer* is the third hypothesis.

3. Materials and Methods

Finding a comfortable method to measure the skin elasticity for the patient as well as for the examiner, depends on many different influencing factors. The perfect combination for an appropriate method to measure structure and location specific skin elasticity, is therefore of prime importance for a reproducible and interpretable outcome. Within the area of research regarding this issue, some measurement methods are prominent and well examined. But only a few are actually suitable in applied medicine and used in the research of tissue regarding problems, which are outlined in this chapter. Especially the "gold-standard" of the current methods, the *Cutometer* is presented in detail, as the outcomes of the novel measurement method, described in this thesis are compared interpreted with the *Cutometer*. In the first subsection, the physical properties of human skin influencing the measurement outcomes, plus some of the most important comparable measurement methods are described. Furthermore, the novel measurement method, developed in this thesis, is outlined. A brief description and the set requirements are shown. The measurement principle, the scheme and therefore, the focus of this work is differentiated. Not only the structure and the related components, but also the whole production process is explained in the following. Concluding, the measuring procedures are described separately. One setup was prepared for the validating measurements, while the other one was constructed for the measurements with selected test persons. Both setups are pictured and the procedure is described in detail.

3.1. Biological and Technical Backgrounds of Skin Elasticity

3.1.1. Skin Layer and Components

Epidermis

The Epidermis, representing the protective layer of the skin, is formed out of stratified keratinizing squamous epithelium. By replacing old epithelial cells on the top of the layer with new keratinocytes, produced at the bottom, five to ten layers of keratinocytes occur. The layer with older dead skin cells is called the stratum corneum, which is a very solid but pliable and wrinkled layer. It plays an important role in the measurement of the mechanical properties of skin, because it presents the interface between skin and

environment. Besides the keratinocytes, which are the main component of the epidermis, also melanocytes, merkel-cells and t-lymphocytes are found in this layer. [6, 4]

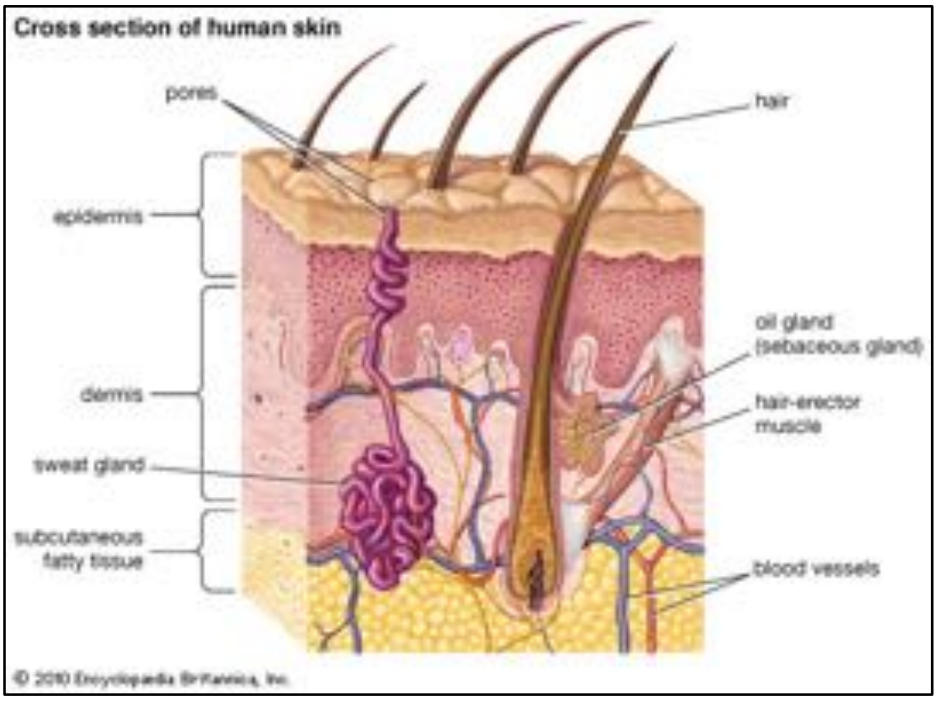

Figure 3-1: Transversal section of human skin [7]

Dermis

The second skin layer, which represents the major mechanical component of the human skin, is also the thickest layer with about 1 mm. From the mechanical view, it is elastic and naturally contracted and anchors the epidermis. Physiologically seen, it supports the epidermis by containing the blood and lymphatic vessels. With most of the sensory cells in it, the dermis functions as the sensory part of the skin. Also, the hair roots are located within the dermis. All these components are embedded in a mixture of collagen and connected elastic fibers. The collagen fibers build a woven scissor-like mesh, in which the fibers are interconnected. This allows a high distention, and, at the same time a huge mechanical strength. After the skin experiences an elongation, the elastic fibers ensure the restoring. All those components are surrounded by the so-called ground substance which is a mixture of proteoglycans and proteins like non-fibrillar collagen. [4, 6]

Subcutis

The subcutis is the linkage layer between the dermis, the muscles and the organs of the body. On the top, it is anchored to the dermis and on the bottom, it is attached to the fascia. With a thickness of 1 mm to 5 cm, depending on sex and body region, the subcutis consists mostly of adipose tissue and behaves like cushion. Furthermore, it is the layer, which is responsible for most of the thermal regulation as it works as a heat insulator and energy storage system. Some big vessels and nerves are also located within it. [4, 6]

3.1.2. Elasticity Responsible Skin Components

The mechanical behavior of the skin depends on many parameters and influences and is a complex interaction between these factors. Two of the above mentioned three layers, dermis and epidermis, are responsible for the ability of the skin to withstand shear loads, tension, pressure and friction. While the dermis responds to the applied elastic deformations, the epidermis provides the skin with the required strength to protect the underlying layers and tissues. As described above, the dermis mostly consists of collagen fibers. These interconnected fibers are working as a network, holding all components, the vessels, nerve cells, hair roots and other fibers together. Via transmitting the outer loads and stresses over the whole network by stretching and conduction to further fibers, the collagen and elastic fiber network works as a spring, like a load-bearing component. The skin is a so-called non-linear elastic material with a viscoelastic behavior, after undergoing a deformation. The typical behavior that characterizes the skin is the two-phase behavior, consisting of the elastic and the viscous part. The stretching process itself undergoes different steps. After applying a force, the collagen fibers will unfold and follow the direction of the applied load. Further, the fibers will stretch until their maximum stretched length. At the end, the collagen and elastic fibers are striated to their maximum, added by little changes of position of the skin components in summary. To give a simple impression of the whole process, the stretching can be divided in three parts. The first part after applying a force is too quick for a viscous deformation and therefore it is purely elastic [8]. The extension occurs immediately after the application. This first elongation is followed by the viscous stretching part of the process. It is divided into two steps, first with variable creep and second with constant creep. All these phenomena are

shown with the so-called creep test, because it describes the stretching way best. A mechanical model to describe the skin components and the different extension parts is shown in Figure 3-2. The first spring represents phase 1, while a dashpot plus a spring represents phase 2, and in phase 3 only a dashpot works. Also, this model is limited by many different parameters originally influencing the mechanical skin behavior, it is a good and simple model to understand the general mechanical behavior. [4]

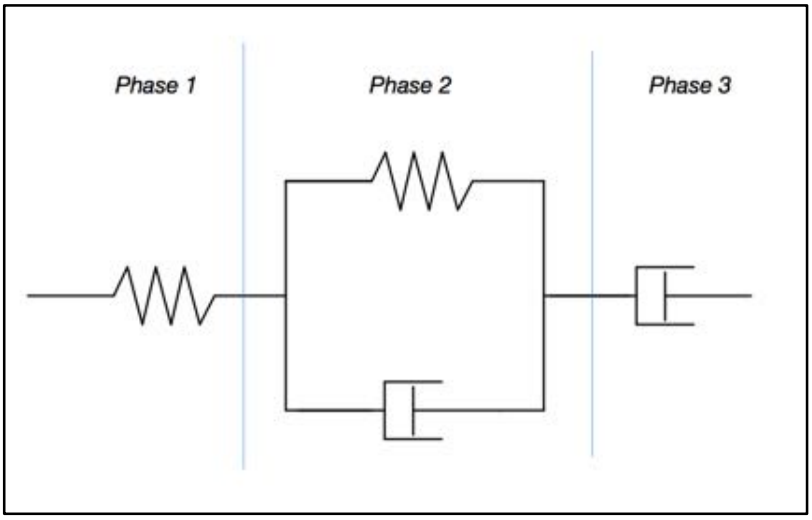

Figure 3-2: Mechanical Model of the Skin. Phase 1: elastic elongation. ,Phase 2: variable creep. Phase 3: constant creep

3.1.3. Engineering Parameters

From the physical view, the elasticity is defined as the returning to the original shape or length of the skin or any elastic material, after termination of the deforming force. In the common mechanic elasticity, tension and elongation are usually measured and further calculated with the Young's modulus or modulus of elasticity. Depending on the stress σ and the resulting strain ε, the modulus is defined as the slope of the stress-strain curve of the loaded material with

$$E_{modulus} = \frac{d\sigma}{d\varepsilon} \qquad (1)$$

. [9]

Because the process of back formation of the skin is not purely elastic, but also viscoelastic, one simple and common formula is used: [10, 4]

$$E_{modulus} = E_{modulus,0} + k\sigma \qquad (2)$$

With $E_{modulus,0}$ as modulus for zero stress and the multiplying parameter k, these two are parameters for the skin under low stress during the elastic phase of the deformation process. [4] Even though this simple physical relation can be applied, many other influences have great impact on the actual deformation process of skin. That means, the modulus of elasticity changes over time and under different experimental conditions. For example, to make results of two different measurement methods comparable, the way of applying a displacement as well as the period, in which the values are taken, must be at least similar.

3.1.4. Influences on the Mechanical Properties and Measurement Results

As already mentioned, the outputs of the measurements are influenced by many different predictable and unpredictable factors, so that it is hard to get a reproducible and comparable value. Even if the measurement set-up is done under the exact same environmental conditions, like constant temperature, pressure and humidity, the time intervals between single measurements make a great different because of the inhomogeneous viscoelastic behavior of the skin, just to mention one difficulty. Further, elasticity of the skin depends strongly on age, body region of the measuring point and gender, as many studies showed. For example, with age, the elasticity is reduced, whereas more significant by women than men. On the other side, the results are better for younger women compared to younger men. [1, 9] Logically, the origin and ancestry, such as Asian, European or Afro-American roots, has a great influence on the results. All these factors must be noted during the measurement of the mechanical conditions of the skin. [11]

Looking at the measurement methods, more various factors influencing the dimension of the results, regarding the measurement set-up arise. Starting from the environment, which should be constant or at least stay under the same conditions, affects the elasticity and

moisture of the skin. Simple parameters such as temperature and humidity apparent change the skin elasticity and therefore need to be considered, to allow furthermore calculations or discussions. The thickness of the skin varies from point to point. Furthermore, underlying tissues like muscle, adipose tissue and bones are changing, which results in the different values in different body regions and further different results by changing the measuring point. Also, important to consider is the time interval between single measurements and the measurement cycles. Because of the viscoelastic behavior, the skin provides different elastic results, when there is not enough time for the viscous parts of the skin to convert back to its initial state. The discovery and classification of the so-called Langer's lines 1861, which are the topological lines on the skin surface all over the body, parallel to the underling tissue, is a determining factor for the measurement, too. It was discovered, that the skin in direction of the lines has a significant higher elasticity, compared to the elasticity perpendicular to the lines. [4] These lines are simple reference points for measurements, to show the direction of action of the elastic fibers within the skin layers. Furthermore, the lines are the site of the skin where the resistance against traction is at best. Hence, the direction of the applied force, to deflect or change the skin from its initial state, is responsible for the biggest difference in the measurement results from method to method. Some common forces, used in known measurement methods are pressure, applied stress or torsion. Every method deflects the skin in a different direction.

3.2. Common Measurement Methods

There are many experimental methods to define or examine the skin elasticity in the literature. But only a few are commonly used and available on the market. Those who are well investigated and in great use, are listed and described here. To see the differences and to light up the wide variations, which are sometimes in a range of multiple kPa, the way of skin adjustment and the measurement range are mentioned. As the method examined in this thesis is supposed to be applied in the regularly usage of every person, the occurrence where the described methods are usually used, are alluded as well.

3.2.1. Torsion

The principle of this *in vivo* method is a rotating disc around the guiding ring, both being attached to the skin with a double-sided adhesive tape. The rotating torque of the disk is constant, leading to a twisted displacement of the skin segment between the ring and the disc. Obviously, the torque displacement takes place in the tangential plane of the skin. According to the elastic resistance of the skin, the angle of the disc stands for a specific elasticity value. With the ability to choose different diameters of the peripheral ring, different tissue depths can be examined. The second smallest ring with a diameter of 3 mm can be used to measure the behaviour of the stratum corneum, while the ring with 5 mm is thought to investigate the behaviour of the whole skin thickness. By sketching the torsion angle to the applied torque, the measurement results are represented. Hence, the size of the torque also has a great influence on the results. [4, 11, 12]

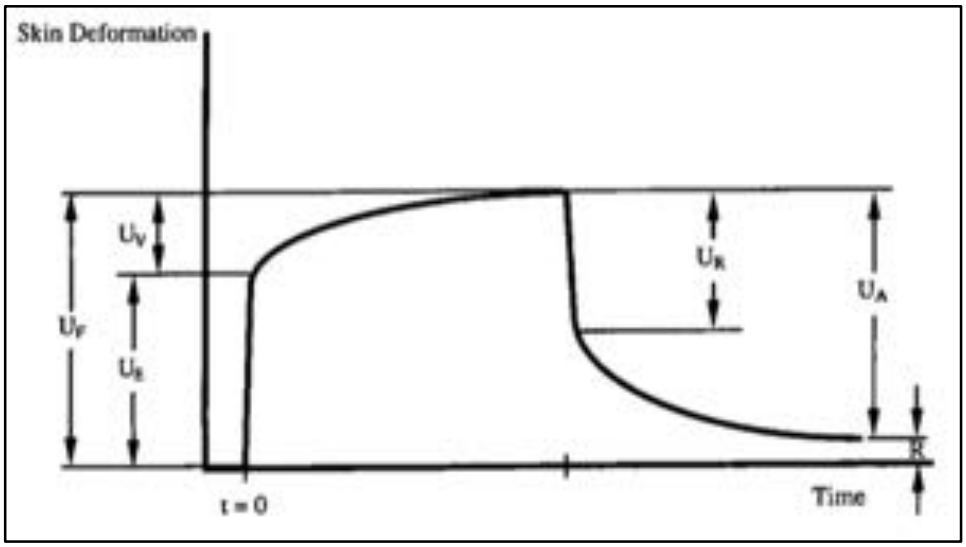

Figure 3-3: Output example of a Dermal Torque Meter [11]

Considering the direction of the Langer's lines, some force vectors applied by the rotating disc are parallel, and some orthogonal to the lines, which are representative for the course of the elastic fibres. One measuring device is the *Dermal Torque Meter*, which has been commercialized from the company 'Dia-Stron Limited'. Another instrument is the *Twistometer* from 'L'Oréal', which was never put on the market. [13] Both are relatively convenient probes. To analyze the raw data, created in the testing probe, it is sent to a personal computer. Subsequently, the data is processed with the associated program.

However, to get frequent measurement of treated skin, the needed equipment and the amount of effort is huge. An example of a resulting graph is given in Figure 3-3. It is close to the evaluation outputs from the *Cutometer*, described in capital 3.2.5. The elasticity in the figure is given by $\frac{U_r}{U_e}$. The value of U_r presents the immediate recovery of the skin after releasing the torsion, while U_e stands for the direct deformation during the stress. Achieved values for the E-Modulus for the methods with the principle of torsion, in the region of the volar forearm are E = 1,12 MPa. [4, 14]

3.2.2. Traction/Extension

An *in vitro* method for measuring the elastic behaviour of skin, is the common principle of the extension method. A skin flap is clamped between two pincers and stretched until its rupture. This method is common and well examined in the field of material science and therefore will no longer be discussed in this thesis. E-modulus values resulting from such measurements are E = 3 MPa. Considering that this method is practicable only for *in vitro* skin, it cannot be applied outside the laboratory. [4, 15]

3.2.3. Indentation

Another widely used method is the ballistometer, where the force is applied by dropping an object onto the skin. The height of the starting point and the resulting recoil, by bouncing of the skin, plus the weight of the falling object corresponds to a specific energy. By dividing the start with the end value, a percentage quantity is given. The greater the resulting height, the more elastic is the skin. The displacement therefore is done in the vertical plane of the skin. Usually, the so-called restitution coefficient is 0,5 on the forearm. [16, 4] One known instrument using this technology is the *Ballistometer* from 'Dia-Stron Limited'. A big difference to the normal ballistometer, is the torsion wire on which the rigid low mass is mounted. By using this principle, the impact of the gravitation is excluded and only the mass and the skin properties influence the measurement results. Just like the other measurement methods, a computer is needed to analyse the results, providing a clear and understandable outcome. [17]

3.2.4. Wave Propagation

The shear wave propagation is another technique to measure the elastic properties of human skin. To do this, two sensors, with a certain distance of a few millimetres are placed on the surface of the skin. One sensor serves as a transmitter, while the other sensor receives the signal. The time the waves need to travel the distance between the sensors,

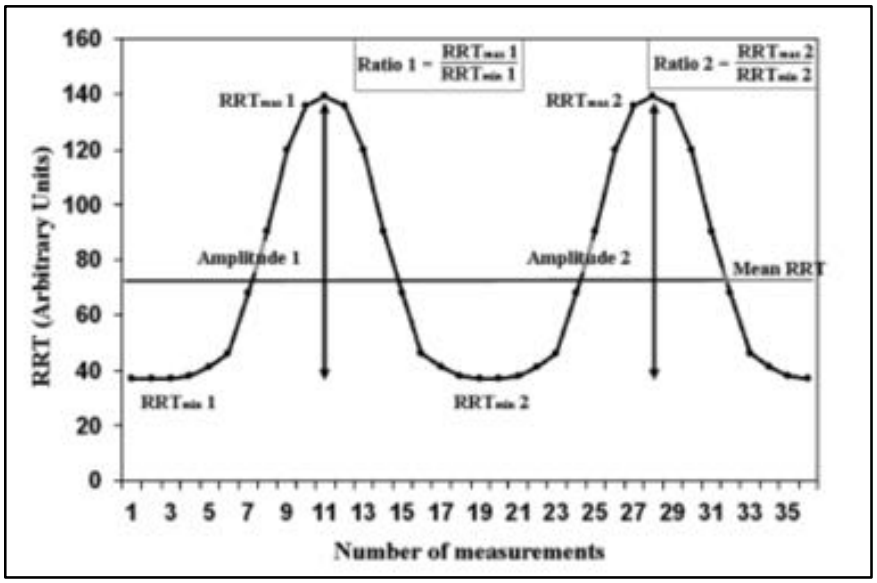

Figure 3-4: Example of an output of measurements done by the Reviscometer (Courage&Khazaka) [19]

is the resonance running time which is expressed in arbitrary units, specifically for an elasticity degree. [18, 19, 13] The *Reviscometer* from 'Courage&Khazaka' is a device based on the described method of measuring. The probes are 2 mm apart and the pressure used on the test material is controlled. Figure 3-4 shows a typical graph-outcome of such measurements. The *Reviscometer* is usually used in research projects and therefore a valuable method of comparison, which is also used as a reference source. For the general user, the *Reviscometer* is not available on the free market. Another instrument using the technique of wave propagation is the so called viscoelastic skin analyser (VESA) constructed in the Radiobiology Laboratory (Sharett Institute of Oncology, Hadassah University Hospital, Jerusalem, Israel). There are experimental as well as clinical prototypes of the VESA system investigated, but both were never put on the market or were developed to the level needed for using this device outside of laboratory conditions. [20, 21]

3.2.5. Suction

The method of applying negative pressure to deform the skin is the most prominent way of measuring the elasticity of human skin. The probe is pressed on the material surface, while the instrument creates a vacuum. By sucking the skin into the probe, the vertical displacement is registered by using an optical system (*Cutometer*, Courage&Khazaka, Cologne, Germany) or by variation of capacity (Dermaflex A, Cortex Technology, Copenhagen, Denmark). A data curve of deformation versus time is created. Figure 3-5 shows a typical outcome of such suction methods. Comparable to the methods of torsion, the parameter $\frac{U_r}{U_e}$ represents, like in the methods of torsion, the net-elasticity without the viscous part. $\frac{U_r}{U_f}$ stands for the elastic recovery to total deformation, while the $\frac{U_a}{U_f}$ shows the gross-elasticity of the skin including the viscous deformation.

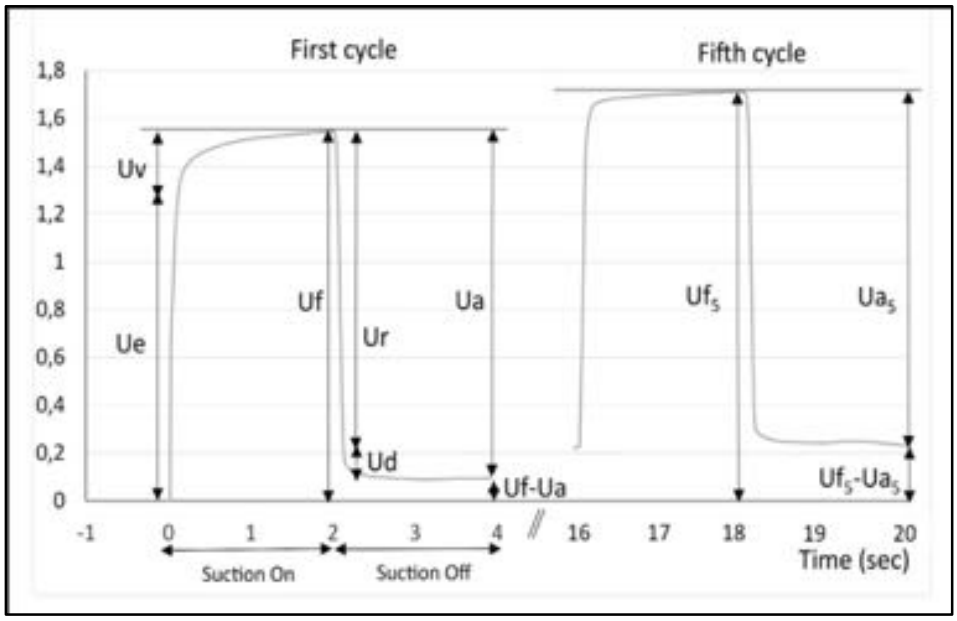

Figure 3-5: Typical outcome scheme of suction method [21]

The final outputs after a measurement using the suction method, are built out of the average of minimum three measurement circles. The *Cutometer* uses defined parameters for these ratios. While $\frac{U_r}{U_e} = R5$, is $\frac{U_r}{U_f}$ defined as $R7$ and $\frac{U_a}{U_f} = R2$. [4, 13, 14, 12, 22]. Both measurement instruments, mentioned above, have relatively large dimensions, and are complicated to transport and to handle. To run the machines and to evaluate the

results, a computer is needed. It is necessary to get familiar with the software to get the wanted results, to interpret them and the arbitrary numbers for a qualitative conclusion about the state of the skin elasticity. Because the *Cutometer* from 'Courage&Khazaka' is the most commonly used system within the cosmetic treatment as well as in the research field, this instrument is defined as the so called 'gold-standard', as mentioned in this thesis.

3.3. Novel Measurement Method

All methods mentioned above are well examined and often used in the research field. However, every single instrument is complicated in the setup. Further, the handling can only be done by an experienced person. The correct classification of the outcomes of each method is difficult to manage and to categorize, because most of the values have random scales. This new examined new method has the aim to be easy in the handling, in contrast to the other methods. The measurement results should be presented in a way so that everybody is able to read and classify the outcomes. Every person, professional or amateur, can use the developed instrument to get a good overview of the current state and the improvement progress of the elasticity of their skin.

3.3.1. Principle and Theoretical Background

The novel method described in this thesis, is thought to be a partly disposable measurement instrument. Some other important requirements are, as already mentioned, the easy handling, the immediate reading and simple interpreting of the dimensionless outputs and especially the comparability of the values with previous results. To fulfil all these given guiding points, the following scheme describing of the measurement method is shown in Figure 3-6. The focus of this thesis is on the measuring principle, done by the capacitive sensor, the so-called Hydrogel-Sensor. For the future, the display of the instrument is supposed to be on the cell phone, by a special designed application tool (app). Almost all newest and common versions of cell phones are provided with the technology of the so-called NFC (Near-Field Communication). NFC devices are able of a contactless exchange, not only of a certain amount of power, but also of data, within a short distance of a few centimetres. This interaction is achieved by electromagnetic induction with loosely linked coils. [23] In general, the measurement device is thought to

be handled as a thin patch, as small as possible, to facilitate the handling. The patch can be simply attached to the examined skin area and removed easily after the measurements are done. To get the needed energy and for the transmission of the resulting data via NFC, the cell phone with the related app, simply needs to be hold a few centimetres on top of the measurement patch.

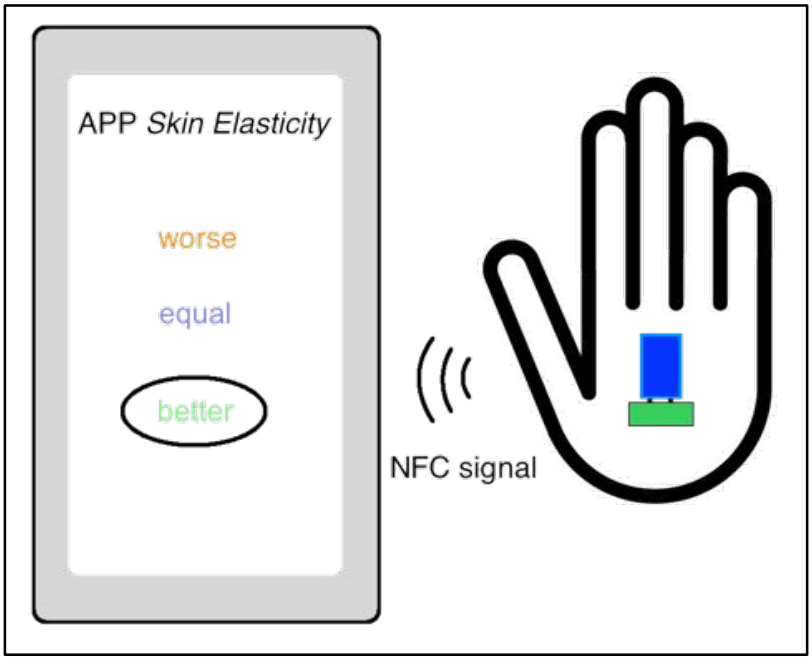

Figure 3-6: Scheme of the novel measurement method

The measurements are comparative studies of grades of elasticity, previous and after a certain treatment. Thus, the results shall be pictured as simple as possible, either through symbols or comparative notification. So, every user is able to read and understand the outcomes. The measuring unit itself is divided in two parts. One is a constant, reused part, containing the NFC transmitter/receiver, and the capacity meter. This section is thought to be the connection between the evaluation unit (cell phone app) and the hydrogel-sensor, the measuring unit. The last unit is the disposable part, shown in Figure 3-7. The Hydrogel-Sensor is built as an elastic capacitor and creates the value, specific for a certain grade of skin elasticity. The connection points are the junction between the reused and disposable part and transfer the necessary energy into the capacitor. In this thesis, the disposable part is the examined unit. The focus of the work is on the usability and the fitting of the capacitor for the above described method.

A common physical principle to define the elasticity of a material is the specific constant value of the so-called Young's modulus or E-modulus (modulus of elasticity). It is a measure of stiffness of an elastic material and is used to define the elastic properties of objects or rather materials.

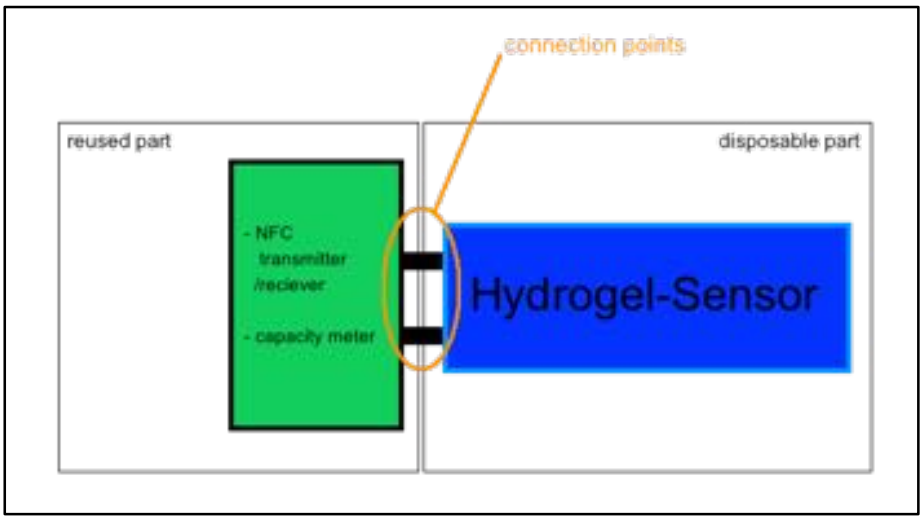

Figure 3-7: Measuring unit. The reused part services as converter (green). Disposable part is the Hydrogel-Sensor, an elastic capacitor (blue).

Usually, it is used to evaluate subjects like wires or columns consisting of metal, polyethylene or rubber. The basic physical description is the ratio of stress along an axis to strain along this axis. When stress is defined as a force per unit area and the strain as ratio of deformation over an initial length, the constant of E-modulus is simply the relation of an applied force to the corresponding change of length. While the definition of E-modulus E [Pa] is

$$E = \frac{\sigma}{\varepsilon} \qquad (3)$$

With $\varepsilon = \frac{\Delta l}{l_0}$ and $\sigma = \frac{F}{A}$ follows the formula for the E-modulus

$$E = \frac{F\, l_0}{A\, \Delta l} \qquad (4)$$

l_0.... initial length [m]

Δl.... change of length caused by force [m]

F.... force per unit area [N]

A.... cross-sectional area [m^2]

The initial length l_0 and the change of length Δl, plus the cross-sectional area A, are the connection to the next physical element used in the method of Hydrogel-Sensor. This novel sensor is constructed as a two-plate-capacitor. The capacity, the dimension of capacitor, is dependent on the distance, the cross-sectional area and the dielectric material between the plates.

$$C = \varepsilon_0 \varepsilon_r \frac{A}{d} \quad (5)$$

C…. capacity [F]

ε_0…. dielectric constant, $= 8.85 \cdot 10^{-12}$ $[\frac{F}{m}]$

ε_r…. permittivity [-]

d…. distance between the plates [mm]

[24]

The dielectric material as well as the electro conductive material, presenting the two plates, has elastic properties. Therefore, the whole capacitor can be stretched and deformed by a force. After releasing the applied load, the capacitive construction returns to the initial form.

The way of building an elastic capacitor is inspired by the work of Christoph Keplinger and Jeong-Yun Sun [25]. They developed an elastic, transparent and ionic material, capable of conducting electricity. In combination with a dielectric material, the ionic conductors build a stretchable capacitor. By mounting the capacitor on a moving underground, e.g. the skin, the sensor is stretched, therefore the area A changes, and a change in capacity is the result. The motion can be obtained by putting on a stretching or squeezing translational force. [25] This correlation between the capacity, length and the force, can be associated with the calculation regarding the E-modulus, to achieve a specific value for a specific state of elasticity.

3.3.2. Components

The elastic capacitor based on the work of Sun and Keplinger [26], is constructed as shown in Figure 3-8. The procedure to stack one layer of 3M VHB Tape and two layers of the hydrogel, ensure the capacitive properties of the stack. To prevent the hydrogel from draining, the first and the third layer of VHB Tape are added to the combination. Additionally, the whole construction can be mounted on an underground and be deformed with it, ensured by the adhesive surface of the VHB Tape. This tape is also used as the dielectric material. It is an elastic transparent double-sided adhesive tape, with an acrylate core. The dynamic shear strength is 48 N/cm^2, while the peeling force is 260 N/100mm [27]. The geometric dimensions are 19 mm wide, 30 mm length and the thickness is 1,1 mm.

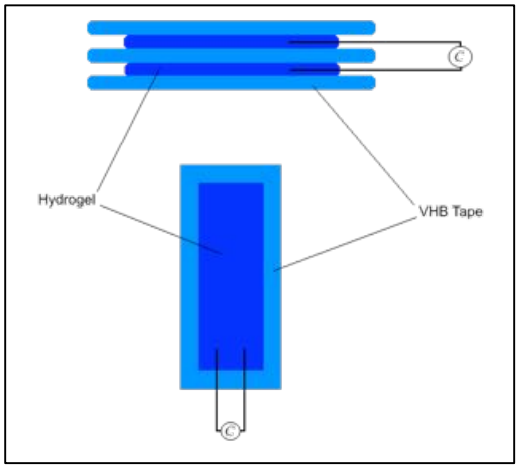

Figure 3-8: Structure of the Hydrogel-Sensor

Sun and Keplinger developed a material with elastic, transparent and conductive properties. They established a special gel based on acrylamide, with NaCl as the conductive ingredient concentrated within the gel. The hydrogel, containing the salt, is connected to a capacity meter, with two metallic wires, consisting of aluminium. [26] Because the acrylamide gel contains a large amount of salt, base metal starts to oxidize after a short time, and the conductivity gets lost. Therefore, the wire should be a noble metal to guarantee a good conductivity for the whole period of use, beginning at the production process to the actual use of the product.

3.3.3. Production Process of Hydrogel-Sensors

The production of the hydrogel is based on the paper of Sun and Keplinger [26]. In this paper, they used a certain amount of every ingredient, fitting to the used mould. Additionally, they followed a given procedure to mix and prepare the gel. The first step is to dissolve the AAm (acrylamide; 2.2 M) and NaCl (2.74 M) into deionized water. Then the MBAA (N,N-methylenebisacrylamide; 0.06 wt%) and AP (ammonium

persulfate; 0.17 wt%) are added. After degassing this solution in a vacuum chamber, the TEMED (N,N,N',N'-tetramethylethylenediamine; 0,25 wt%) is added. This mixture is then poured into a glass mould (100.00 mm x 100.00 mm x 0,1 mm) and covered with a transparent glass plate (3 mm thickness). Subsequently, the mould with the acrylamide fluid is cured with UV light cross-linker for 20 min with 254 nm wavelength and 8 W power. After this time, the gel is immersed in aqueous solution with the same concentration of NaCl like the AAm monomer for more than 24 hours. The estimated thickness of the resulting gel is 0.2 mm. The cutting takes place by a laser cutting system. Before the gel is stacked on the VHB tape, the surface is dried with N_2 gas for 1 min, to improve the adhesion between gel and VHB tape.

Because the device and therefore the measurement method need to fulfil some requirements, the recipe and the production process were changed. The adjustments on the procedure depended on some requirements for the measurement method:

i. the size must be as small as possible to fulfil the requirements of a small and handy device
ii. the method must be easy in handling to facilitate the use for the consumer

To fit the composition to the requirements of this work, some changes are done to the basic recipe of Sun and Keplinger, as described in chapter 4.1.1.

3.4. Measuring Procedure

For the measurements done with the Hydrogel-Sensor, some basic components are used. Because the focus of this thesis is on the Hydrogel-Sensor, the other parts mentioned in the scheme of the measuring method, are still a prospective development task. Therefore, the monitoring and supply of the sensor were done by a regular RCL-meter (Keysight E4980A) and other standard instruments. Figure 3-9 shows the whole scheme of the measurement setup as a boxplot, as well as a photography.

The idea of this method is to have an elastic capacitor, which is presented in this thesis by the Hydrogel-Sensor. The sensor is able to measure a capacity to an according length of the capacitor. This length stands for a specific capability of the skin to withstand the stretching force, applied by the elastic sensor. In turn, this force is attached by the

previous stretching of the Hydrogel-Sensor. Because of its elastic properties, the sensor tries to return to its initial geometry.

3.4.1. Validating Measurements

Two glass plates, one fix, the other relocatable, were placed on a cm/mm scale. To ensure a straight translational movement of the loose plate, it was guided along a rail like metal bar. Both ends of the Hydrogel-Sensor were clued to the plates. The central space of the sensor was free in movement, to provide the freedom of movement of the sensor. Under the glass plates, a paper with a millimetre scale was placed. Therefore, the distance

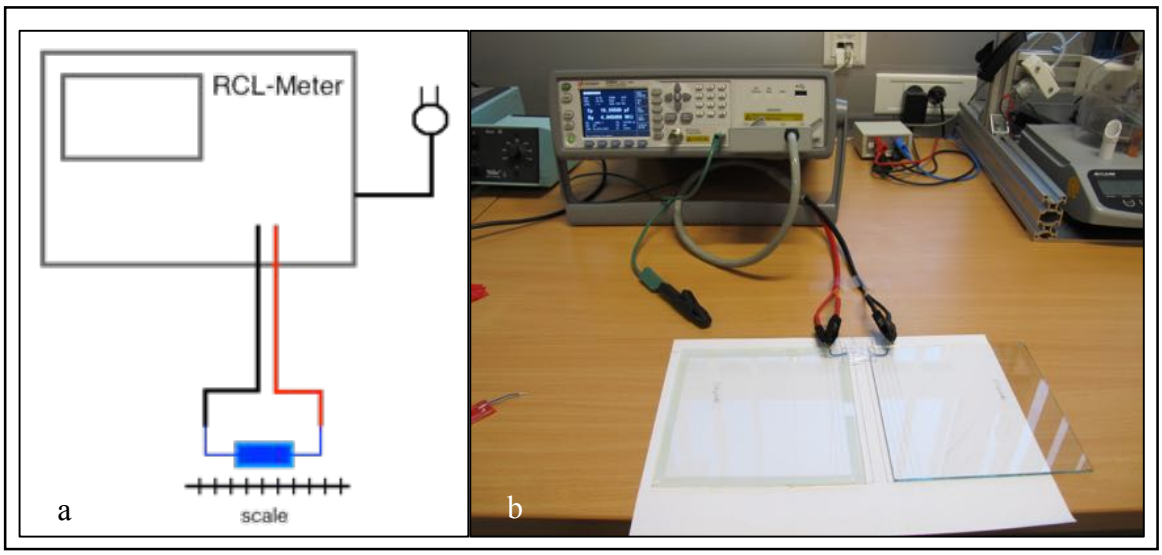

Figure 3-9: Measurement setup scheme for the first studies in the laboratory

between the plates, and therefore the length of the Hydrogel-Sensor could be defined. One advantage of the glass plate is the isolating function. A capacity measurement is very sensitive of surrounding influencing voltages. Because glass is a very good isolator, little shielding of the surrounding occurring capacities was provided. The measurement frequency to run the capacitor was set at 30 kHz. [26] To create a change in capacity, the length of the Hydrogel-Sensor was changed by relocating the right glass plate. Additional, for the precise measurements and the calculations graph paper was used as a scale underneath the glass plates. This measurement setup was used for the results in chapter 4.1.2 and 4.2.1.

3.4.2. Measurements on Test Persons

To see if there is a correlation between the examined measurement method with the Hydrogel-Sensor and the "gold-standard" the *Cutometer*, some measurements were done. For these comparable measurements, 12 persons were investigated, whereby each forearm of each side was tested. Altogether, 24 results were achieved. The test persons were defined within 4 different groups. Because the age and the sex have the greatest impact on the elasticity of the skin, these criterions were used to classify the categories. [5, 4, 1, 18] The set test groups were defined as shown in Table 3-1.

Test Groups		Sex	
		female	male
Age	< 30	*1*	*2*
	> 50	*3*	*4*

Table 3-1: Definition criteria for the test groups

The measurements, with the aim to compare the Hydrogel-Sensor values with the "gold-standard", the *Cutometer* (Courage&Khazaka), were done with the setup shown in Figure 3-10. The left picture (Figure 3-10, a) shows the necessarily parts for the *Cutometer* measurements. The *Cutometer* with its probes was located on the left side, connected to the computer, with the required software to analyse the outputs. On the right picture (Figure 3-10, b), the setup for the Hydrogel-Sensor measurements is shown. The setup to drive the sensors was exact the same as described in chapter 3.4.1. To guarantee comparable values, the measurements must be done at the same environmental conditions. Therefore, the setup was placed in the same room, and the procedure of both measurements was done within 30 min. The temperature and the humidity were controlled by an air-conditioner and monitored by the *Cutometer*. Because of the smaller impact on the skin area of the *Cutometer*, which has influence on the mechanical properties within the treated area, the first measurements were done with the suction method. [28] To have easy access and comfortable conditions for the test persons, the ulnar area of the forearm was chosen. Another reason for the decision to use the forearm

is the great amount of investigations and measurement results of this body region, which are available in literature. This makes it easier to compare the results of this thesis with other measurement methods. The procedure of the measurements with the Hydrogel-Sensors required some preparations. After the measurements with the *Cutometer* were done, the measurement area was again cleaned with alcohol, to eliminate all left residuals of the previous measurement and to ensure the proper adhesion of the sensor on the skin.

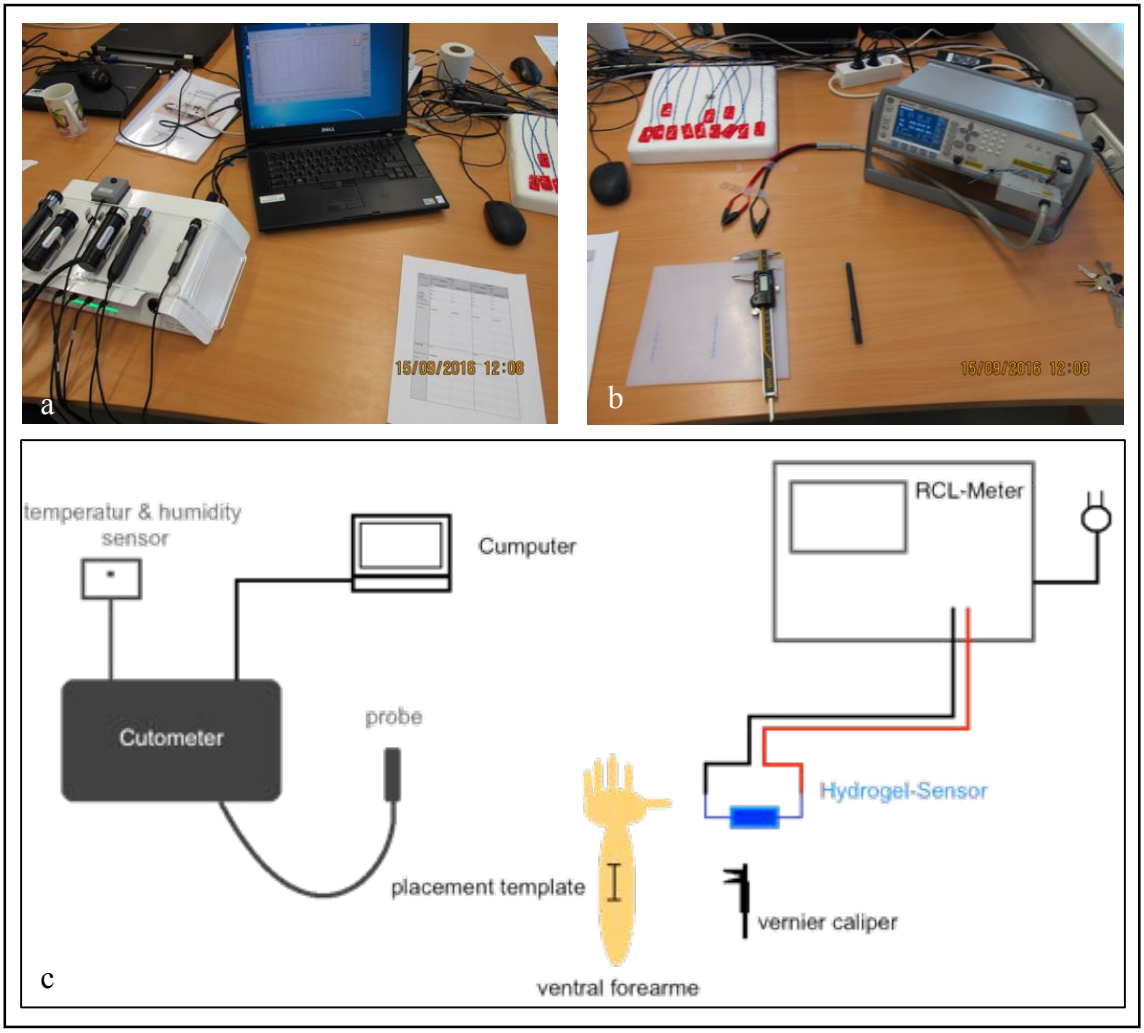

Figure 3-10: Setup for the comparable measurements between Hydrogel-Sensor and Cutometer. a) and b) photograph of the actual setup. c) boxplot of the setup scheme. On the left side, the Cutometer, connected to the computer to run the evaluation program, and the sensor to detect the temperature and the humidity during the measurements. On the right side, the Hydrogel-Sensor with the RCL-Meter to measure the capacity. The Vernier caliper was used to measure the length of the Hydrogel-Sensor.

Before the mounting of the sensor on the skin, the initial length and capacity of the Hydrogel-Sensor was noted. The assessment of the length was done with a regular caliper. The ground capacity was taken, while the sensor had contact with the skin. To transmit

the required force to deform the skin, the Hydrogel-Sensor was pre-stretched to a certain length and subsequently glued on the skin. The previous stretching was done by hand.

In order to ensure the correct starting length of the sensor, a placement template was drawn on the measuring area of the skin. By pressing the sensor onto the skin, the start of the measurement was initiated. Because of the elastic performance of the Hydrogel-Sensor, it tries to get back in its initial geometry. Therefore, after 5-8 sec. the length of the sensor was measured again. However, because of the resistance of the skin to deform, the Hydrogel-Sensor could not get back in the initial length completely. Hence, a small difference between the initial length and the arrived length remained, after mounting the Hydrogel-Sensor on the skin. This difference was then used as the representing measured value for the elasticity of the skin.

4. Results

4.1. Production Process of Hydrogel-Sensor

As already described in chapter 3.3.3, the recipe of the hydrogels was recreated based on Sun and Keplinger [26] with some changes in the composition and production process. Because of the already detailed description, only the changes done in the ground recipe on order to get the right results to fulfil the requirements regarding the elasticity, geometry and conductivity, are outlined in this chapter 3.3.3.

To find a usable thickness and elasticity of the gel, some measurements were done with different sizes of the capacitors.

4.1.1. Hydrogel Recipe

The initial indication to get the required recipe of the hydrogel was the necessary consistence of the finished gel, according to the requirements mentioned in chapter 3.3.3. The first composition of the ingredients resulted in a liquid medium without any shape or elastic properties. Therefore, the next step was to add some more TEMED. This ingredient is responsible to solidify the gel and start the crosslinking of the amides. Finally, the right recipe contained 12.5 ml AAm (2.2 M) + NaCl (2.74 M) as monomer, 472.5 µl MBAA (24 g/150 ml) functioning as crosslinker, 472.5 µl AP (2.28 g/ 50 ml) as radical initiator and 24 µl TEMED as a crosslinking accelerator. After preparing the required amount of every substance, the dissolved NaCl in the AAm was blended with MBAA and AP. The TEMED, was added at last, because it has the function to make the fluid become a gel-like consistence. The mixture was then poured in a glass-mould with the size of 128.25 x 129.45 x 0.80 mm. To prevent the gel to stick on the glass, the mould bottom was coated with a Teflon layer. This gives a volume of 13281.57 mm^3. With the above given indications of quantities, the fluid has approximately the volume of 12.5 ml = 12500 mm^3. After some test runs, this amount proved to be right for the used mould. To sustain the fluid in its process to get solid, the setup was irradiated with UV radiation for 30 minutes (Figure 4-1, a). Afterwards, the mould with the solidified hydrogel was immersed in salt water with the same concentration as the originally used dissolution with the AAm + NaCl (Figure 4-1, b). Almost 40 minutes later, the gel absorbed some of the salt water and swelled to around 20% of its original size. This process caused the gel to

detach from the mould. At the end, the hydrogel obtained the size of approximately 154.0 x 155.5 x 10.0 mm (Figure 4-1, c). To guarantee the precise required dimensions of the hydrogel and the VHB Tape, a Teflon template was built, with a millimetre pattern scale. The gel was cut with a regular razor blade. The salt particles need to be free to move to assure the current flow. Thus, the elasticity as well as the conductivity of the hydrogel

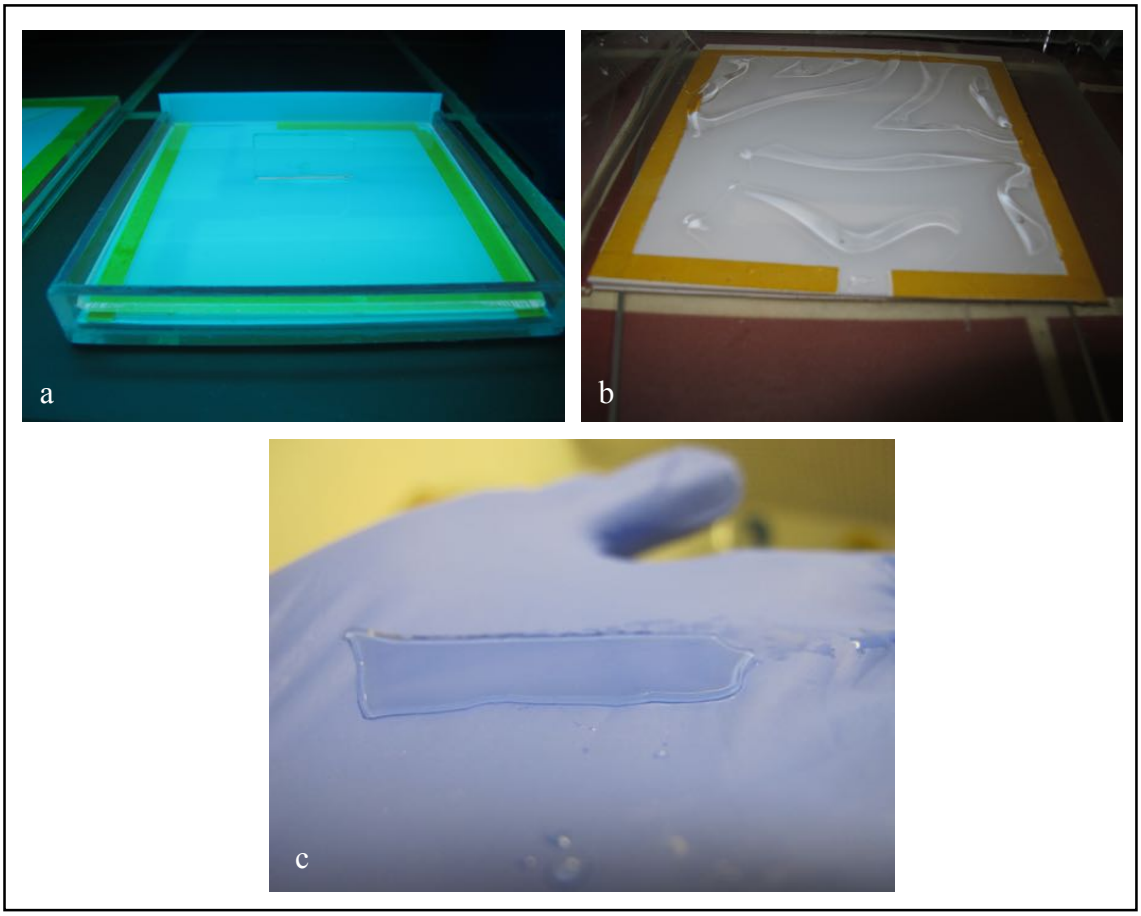

Figure 4-1: Manufacturing process of the hydrogel. a) mould including fluid with UV radiation. b) swelling hydrogel. c) final thickness of the hydrogel

depends, besides the salt content, on the degree of humidity. But that characteristic is a disadvantage to the cohesion of the hydrogel and the VHB Tape. A state, where the surface of the hydrogel is as dry as possible, but not the hydrogel itself, needs to be found. To get a well-prepared surface, approximately 30 min are needed. The next step was to place the hydrogel strip in the middle of the VHB Tapes, to assure the overlay of both strips as exact as possible (Figure 4-2, a). For that reason, a template was build, which facilitates the procedure of correct positioning. Further, to ensure a closed and stabile construction between the VHB tape and the hydrogel, an overlay of the VHB tape over

the hydrogel was necessary. Between the first VHB Tape and hydrogel strip, and the second hydrogel and last VHB Tape, two isolated aluminum wires were placed. The adhesion of the VHB tape was used to fix the wire (Figure 4-2, b). The last step was to stack the components on top of each other, with the middle VHB Tape as the adhesive and dielectric component (Figure 4-2, c).

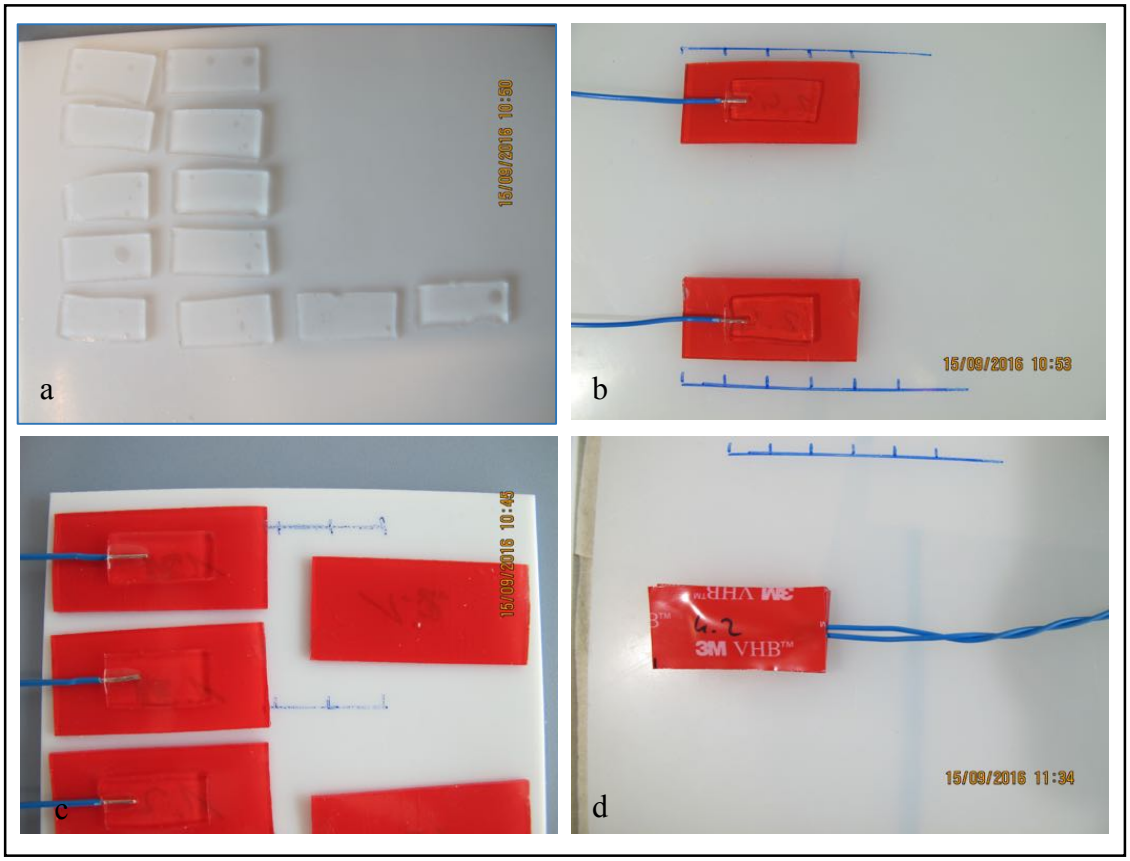

Figure 4-2: Production process Hydrogel-Sensor. a) Cut hydrogel strips, 10x20mm. b) positioning of hydrogel on HVB Tape + aluminum wire. c) three component layers. d) finished Hydrogel-Sensor

4.1.2. Shape and Dimensions of the Sensor

In terms of finding the size of the capacitor, fitting the requirements mentioned in chapter 3.3.3, a preselection of four sensors with different sizes, described below, were set and investigated. The capacitors only differ in the cross-sectional area. The thickness was equal for each sensor. Four of each sensor sizes were used for these studies. The utilized measurement setup is described in chapter 3.4.1. The applied stretching length depended on the size of each sensor, whereby the upper limit was set by the failure of the Hydrogel-Sensors. The test criteria were handling, stability and output values.

Sensor 1 (hydrogel 10 x 10 mm, VHB tape 20 x 30 mm):

This sensor construction was the smallest prototype used in this thesis. Due to the small scope for holding/fixing, handling and therefore the tests were difficult to manage, making it impossible to get measurement values. Either the hydrogel within the VHB tape ripped of, or detached from the aluminum wire. It could be observed that the hydrogel did not follow the initiated stretching of the VHB tape.

Sensor 2 (hydrogel 10 x 20 mm, VHB tape 20 x 40 mm):

Using this geometry, the measurement values could be created and the handling was comfortable and easy. The maximal stretching length was about 30 mm, where the capacitor was destroyed. Afterwards, the hydrogel ripped off either in the area where the wires attached, or in the middle of the hydrogel strip. Another case of failed measurement was when the measurement wire detached. Figure 4-3 shows some of the failure examples.

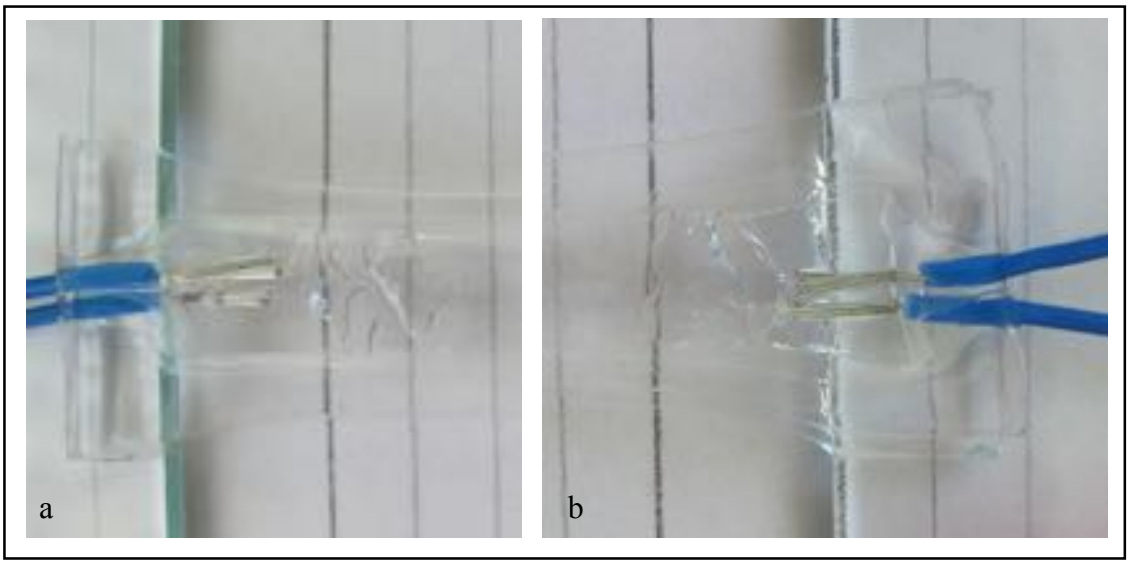

Figure 4-3: Failure cases of measurement, sensor 2. a) rupture of the hydrogel in the middle of the capacitor. b) rupture of the hydrogel on the tips of the measurement wires.

Sensor 3 (hydrogel 10 x 30 mm, VHB tape 20 x 50 mm):

The successful measurements resulted in a stretching length of around 40 to 50 mm. It needed a slight strength application in order to stretch the flexible capacitor, but because of the huge stretching length, the accuracy to stretch in a straight line was challenging.

Sensor 4 (hydrogel 20 x 20 mm, VHB tape 30 x 30 mm):

Due to the geometry, the sensor was difficult in its handling. It required a lot of strength to stretch the capacitor. Furthermore, the constant and homogeneous stretching by hand was almost impossible because of the large width.

4.2. Measurement Results

After the first investigations of the elastic capacitor (chapter 4.1.) the geometry of sensor 2 *(hydrogel 10 x 20 mm, VHB tape 20 x 40 mm)* was used for all further experiments and validations. The operating frequency was chosen after Sun and Keplinger at 30kHz.

4.2.1. Hydrogel-Sensor Key Figures

In order to examine the suitability of the measurements, to investigate the elasticity of human skin with the elastic Hydrogel-Sensors, some validations regarding the key figures were done. Parameters like the reproducibility, the correlation and some material parameters, compared to the literature, were evaluated. The used experimental setup is described in chapter 3.4.1. Additionally, the impact of the environment on the capacity outputs of the sensor was examined.

4.2.1.1. Reproducibility

One important parameter is the reproducibility of the output values. Because of the first impressions und experiences of the preceded measurements (chapter 4.1.2), the generated extension length was 10 mm. Table 4-1 presents the obtained results.

	capacity [pF]		
sensor extension 10 mm	initial state	extension	difference
initial state	8.16		
1. extension	7.97	11.87	3.71
2. extension	8.03	11.64	3.67
3. extension	8.07	11.65	3.62
4. extension	7.89	11.57	3.50
5. extension	7.91	11.64	3.75
6. extension	7.99	11.54	3.63
7. extension	7.92	11.50	3.51
8. extension	7.91	11.84	3.92
9. extension	7.87	11.46	3.55
10. extension	7.93	11.75	3.88
average	7.97	11.65	3.67
standard deviation	± 0.09	± 0.14	± 0.14

Table 4-1: Reproducibility of one sensor over 10 extensions and relief to the initial state. Extension distance = 10 mm. Pause between extensions 10 s.

The values were created by stretching one sensor ten times. In order to guarantee the complete return of the stretchable capacitor to its initial form, a pause time of 10 seconds between each extension was implemented. Within this time, the VHB tape and the hydrogel returned to the original length and the values, measured with the RCL-meter, showed continuous decline to the start value. Figure 4-4 represents the evaluated data of Table 4-1, whereby the values of the column “difference” are shown.

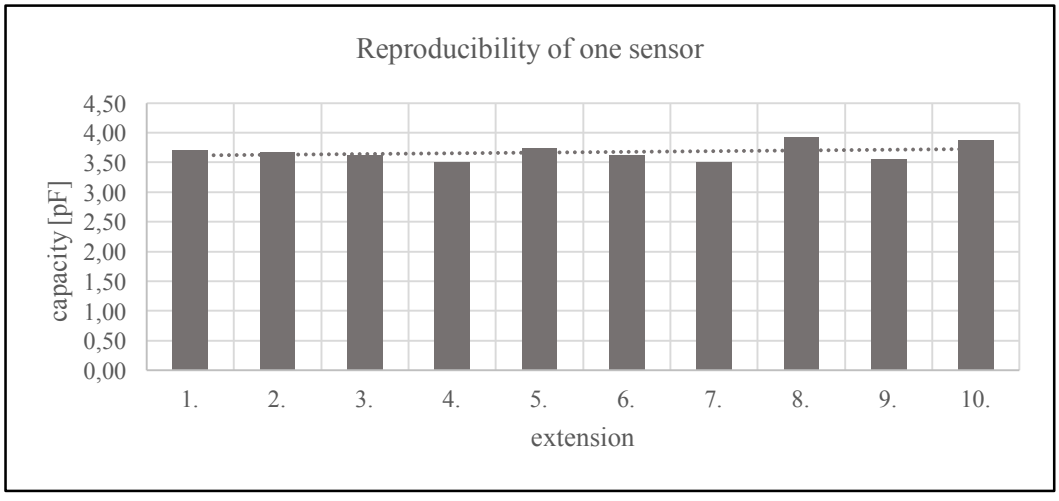

Figure 4-4: Reproducibility of one Sensor

These quantities are important for the analysis, because the difference between the start and the end values are representative for the elasticity of the skin in further investigations. The standard deviations of the measured values are ±0.09 pF and ±0.14 pF, whereas ±0.14 pF for the difference. The dotted line represents the course of the capacity outputs, whereas the slope of 0.0127 is representative for the linearity of the results.

4.2.1.2. Sensor Correlation

Further investigations were done with ten different sensors (Table 4-2). In order to show the correlation between those capacitors, and therefore the independency of the measurement outputs from the applied Hydrogel-Sensor, the experiments were done within 40 minutes and constant environmental conditions. Furthermore, the hydrogel strips came from different production batches. The missing results in Table 4-2 arise out of the failure of sensor 9 and 10 at the try to stretch up to 20 mm.

	extension [mm]					
sensor [pF]	0	10	Δ 0-10	20	Δ 10-20	Δ 0-20
1.	8.11	11.34	3.23	14.44	3.10	6.33
2.	7.90	12.13	4.23	14.15	2.02	6.25
3.	7.75	11.42	3.67	14.74	3.32	6.99
4.	8.08	11.60	3.52	14.45	2.85	6.37
5.	8.13	11.89	3.76	14.57	2.68	6.44
6.	8.08	11.70	3.62	15.12	3.42	7.04
7.	7.92	11.66	3.74	15.12	3.46	7.20
8.	7.63	11.93	4.30	15.33	3.40	7.70
9.	8.57	12.55	3.98	/	/	/
10.	8.77	12.71	3.94	/	/	/
		average	3.80		3.03	6.79
		standard deviation	± 0.32		± 0.50	± 0.52

Table 4-2: Correlation of 10 different Hydrogel-Sensors. Sensor geometry 10x20 mm. Measurement frequency 30 kHz

The hydrogel experienced the same ruptures as shown in Figure 4-3. In the cases of sensor 1 to 8, all ripped off at the tips of the measurement wire during the attempt to bring the Hydrogel-Sensor to a stretching length of 30 mm.

Of interest are again, the differences between the start and the end values. Therefore, the results are classified into two categories. The first category (Figure 4-5, blue box) represents the difference values between the initial state and 10 mm extension. The second category is defined for the extension from 10 mm to 20 mm (Figure 4-5, green box).

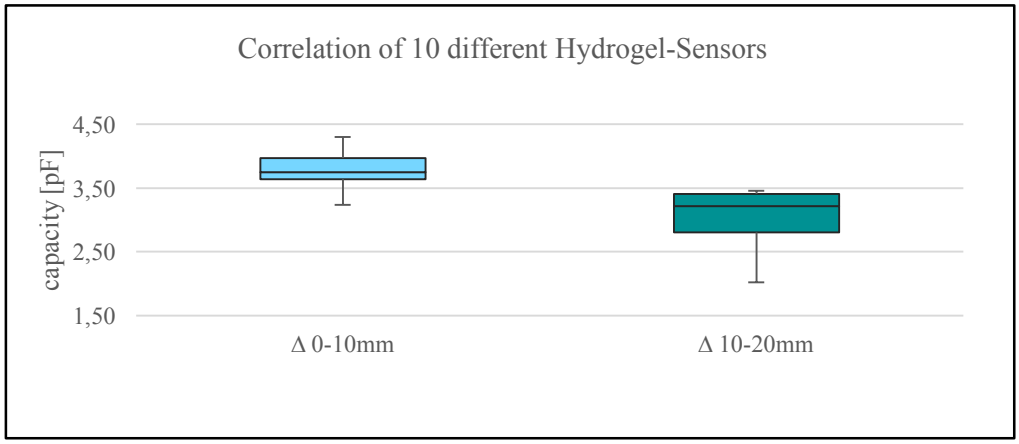

Figure 4-5: Correlation of 10 Hydrogel-Sensors. (blue) Extension difference beginning at the initial state to 10 mm. (green) Extension difference beginning at 10 mm extension to 20 mm extension.

The results are clearly illustrated in a boxplot with the maximum and minimum values for each category plotted as the error bars above and under the colored boxes. Within these boxes, 50% of all obtained numbers are located.

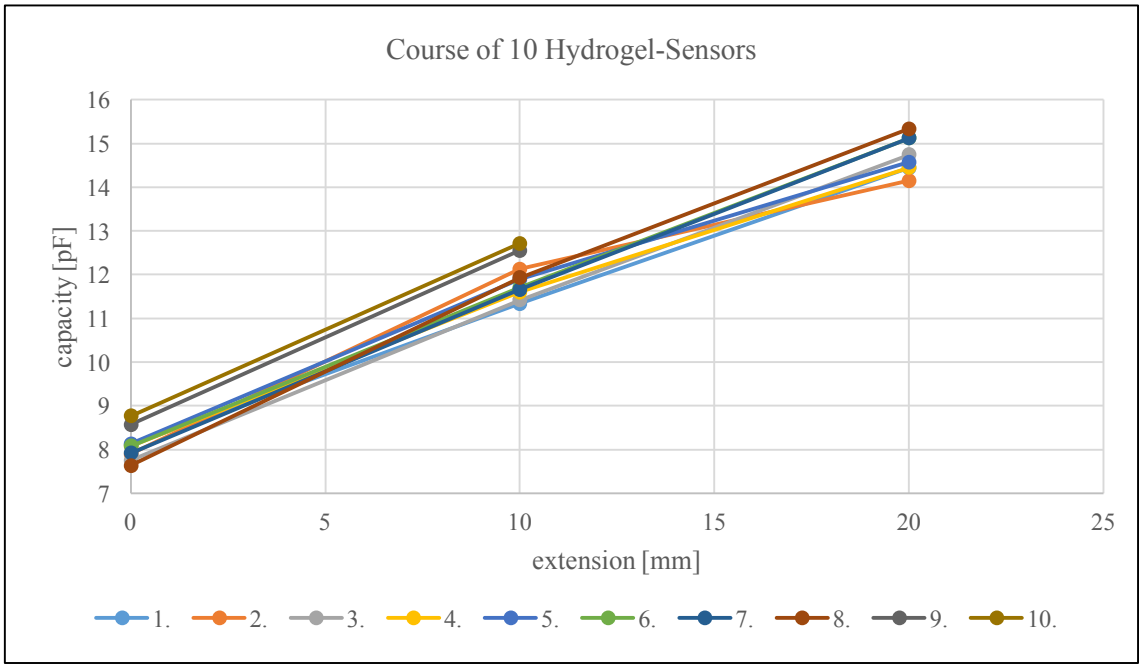

Figure 4-6: Course of different Hydrogel-Sensors.

The line within the middle of the colored boxes, represents the median of all data per category. In order to check the correlation of the investigated capacitors, the courses of the measured outputs for each sensor is monitored in Figure 4-6. Whereby the measuring points, all taken from Table 4-2, are at the initial state 10 mm and 20 mm extension.

4.2.1.3. Relation of extension over 2 mm to capacity value

Another research series was done with the Hydrogel-Sensors. The used sensors in the following examinations are from a different batch and a thicker VHB tape was used, than in the previous measurement. This thickness difference was 1.0 mm. However, the geometry was the same as the one of sensor 2 in chapter 4.1.2. Here, the capacitors were expended to a stretching length of 10 mm, whereby the resulting capacity according to every 2 mm was taken. Therefore, the linearity of the measured output value along a certain length was investigated. In order to confirm the results of one sensor, the procedure was repeated with two other sensors. In Table 4-3 the results are listed.

1. sensor			
extension [mm]	C [pF]	ΔC [pF]	Δ*C*/Δ*l* [pF/mm]
0	18.25	-	-
2	18.56	0.31	0.15
4	19.12	0.56	0.28
6	19.66	0.54	0.27
8	20.17	0.51	0.26
10	20.67	0.50	0.25
	average	0.48	0.24
	standard deviation	± 0.10	± 0.05

2. sensor			
extension [mm]	C [pF]	ΔC [pF]	Δ*C*/Δ*l* [pF/mm]
0	18.10	-	-
2	18.63	0.53	0.26
4	19.29	0.66	0.33
6	19.94	0.65	0.33
8	20.55	0.61	0.31
10	20.87	0.32	0.16
	average	0.55	0.28
	standard deviation	± 0.14	± 0.07

3. sensor			
extension [mm]	C [pF]	ΔC [pF]	$\Delta C/\Delta l$ [pF/mm]
0	17.26	-	-
2	17.70	0.44	0.22
4	18.16	0.46	0.23
6	18.51	0.35	0.18
8	18.98	0.47	0.23
10	19.44	0.46	0.23
	average	0.44	0.22
	standard deviation	± 0.05	± 0.02

Table 4-3: Extension linearity of 3 sensors per 2 mm

The column “difference [pF]” represents the gap between the preceded and the following capacity value, according to the induced change of length. As the table implies, the average of the results in this column is 0.44 pF, with a standard deviation of ±0.05 pF. The third column shows the factor of the capacity in proportion to the extension difference. The average is 1.22 and the standard deviation 0.15. In order to present the results in a better and open way, the ratios of the 3 sensors are collected in Table 4-4.

	1. sensor		2. sensor		3. sensor	
extension [mm]	C/C_0	l/l_0	C/C_0	l/l_0	C/C_0	l/l_0
0	1.00	1.00	1.00	1.00	1.00	1.00
2	1.02	1.11	1.03	1.09	1.03	1.11
4	1.05	1.21	1.07	1.18	1.05	1.23
6	1.08	1.32	1.10	1.27	1.07	1.34
8	1.11	1.42	1.14	1.36	1.10	1.46
10	1.13	1.53	1.15	1.44	1.13	1.57

Table 4-4: Correlation capacity to length change

For illustration, the content of Table 4-4 is plotted in Figure 4-7. The coefficient of determination R^2 is shown as well. For sensor 1, R^2 = 0. 9955, for sensor 2 R^2 = 0. 9922 and for sensor 3 R^2 = 0.9988. It can be seen, that all values are close to 1.0000, which is an indication for a linear relation between *C* and the length change *l*. [29]

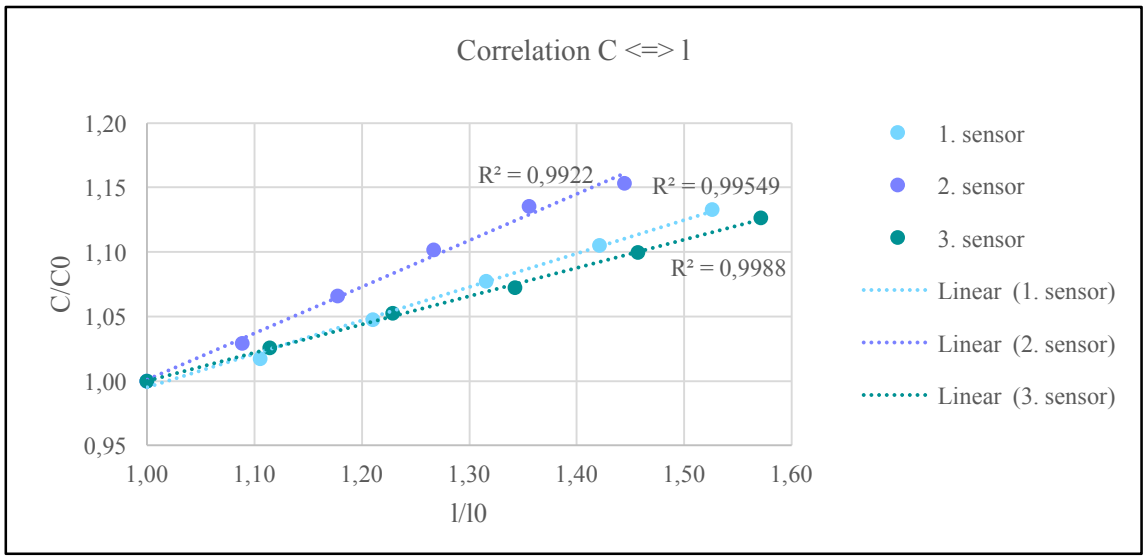

Figure 4-7: Correlation capacity to length

4.2.1.4. Calculation of material parameter

In order to get other important information about the properties of the Hydrogel-Sensor, some calculations and definitions were done.

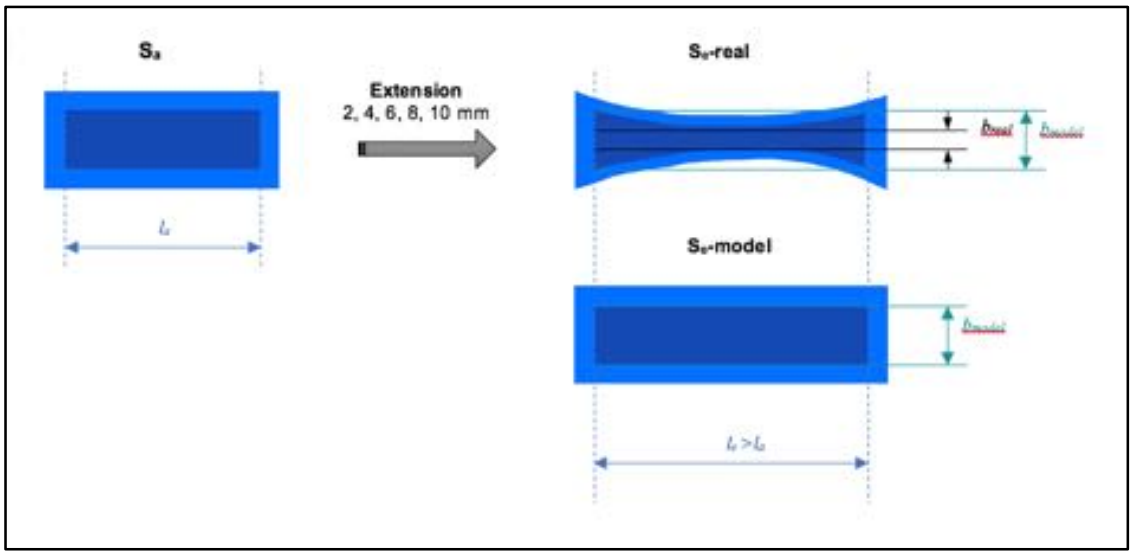

Figure 4-8: Hydrogel-Sensor nomenclature for calculation of material constants

The most important connection of parameters of the measurement with the flexible capacitors is the correlation of the measured capacity (C_{real}) with the real and exactly calculated variables to the capacity (C_{model}), obtained with the set and measured parameters. As explained in chapter 3.3.1, the connection parameter between the elasticity of the skin and the output values of the measurement method, the capacity, is among other things, the cross-sectional area A. On the other hand, this value depends on the length and the width of the capacitor, thus, on the geometry of the hydrogel. However, because the capacity value is in relation to the length change of the sensor, the evidence that the capacity value is independent of the sensors width and only depended on the length of the flexible capacitor, must be rendered. At first, the definitions for the flexible sensor were set (Figure 4-8). The initial length of the un-stretched capacitor (S_a) is defined as l_a. After the extension, the new length is named l_e. In further calculations, l_{real} is equal with l_a or l_e.

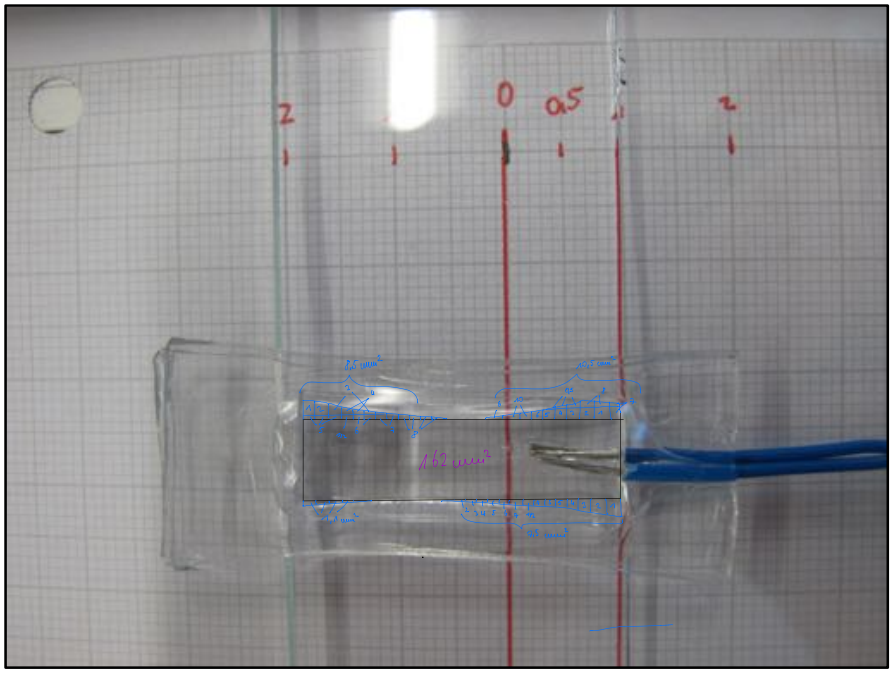

Figure 4-9: Picture analysis of the cross-sectional area of the stretched Hydrogel-Sensor. By subdividing the cross-sectional area of the stretched hydrogel into smaller units, the total area of the hydrogels, which are building the capacitor unit are calculated.

For the calculation of the capacities (C_{real}, C_{model}) S_e-real and S_e-model are defined. In real life, the stretching of the sensor involves a deformation of the hydrogel. The rectangular form of the acrylamide stripe is stretched and gets a necking area on both elongated sides, as demonstrated in S_e-real. Therefore, not only the length change influences the cross-sectional area, but also the irregular width of the hydrogel. Based on

this fact, A_{real}, was calculated with the method of picture analysis (Figure 4-9). The stretching of the Hydrogel-Sensor was done as described in chapter 3.4.1. The graph paper was used as template and for the calculation of A_{real}. With the formula for A_{real}, the average of b_{real} was calculated with

$$\frac{A_{real}}{l_{real}} = b_{real} \qquad (6)$$

Additionally, C_{real} was given by the RCL-meter. For further examinations with formula

$$C = \varepsilon_0 \varepsilon_r \frac{A}{d} \qquad (5)$$

the material constant is achieved out of

$$\varepsilon_r = \frac{C_{real} \cdot d}{A_{real} \cdot \varepsilon_0} \qquad (7)$$

d…. thickness of theVHB tape [mm], assumed to be constant at 0.9 mm

C_{real}…. measured capacity [F]

A_{real}…. calculated cross-sectional area [mm^2]

l_{real}…. measured length of hydrogel strip [mm]

b_{real}…. calculated width of hydrogel strip [mm]

With the obtained ε_r, the capacity for the model S_e-model C_{model}, could be determined. The used formula was

$$C_{model} = \varepsilon_0 \varepsilon_r \frac{b_{model} \cdot l_{real}}{d} \qquad (8)$$

The obtained results of the above-mentioned calculations are summarized in Table 4-5: Capacity of the sensors by picture analysis (C_{real}) to the calculated capacity (C_{model});

measured and calculated cross-sectional areas, extension length = 10 mm. Whereby the first row (C_{real} / C_{model} [pF]) shows the ratio of the calculated and the measured capacity value of the examined sensors.

	1. sensor	2. sensor	3. sensor
C_{real} / C_{model}	0.936	0.945	0.952
C_{real} [pF]	$2{,}067 \cdot 10^{-11}$	$2{,}087 \cdot 10^{-11}$	$1{,}726 \cdot 10^{-11}$
C_{model} [pF]	$2{,}208 \cdot 10^{-11}$	$2{,}209 \cdot 10^{-11}$	$1{,}813 \cdot 10^{-11}$
A_{real} [mm^2]	$18.96 \cdot 10^{-5}$	$24.66 \cdot 10^{-5}$	$15.43 \cdot 10^{-5}$
A_{model} [mm^2]	$20.25 \cdot 10^{-5}$	$26.10 \cdot 10^{-5}$	$16.20 \cdot 10^{-5}$
ε_r	11.09	8.61	11.38

Table 4-5: Capacity of the sensors by picture analysis (C_{real}) to the calculated capacity (C_{model}); measured and calculated cross-sectional areas, extension length = 10 mm

Figure 4-10 clearly illustrates the relation between C_{real} and C_{model}.

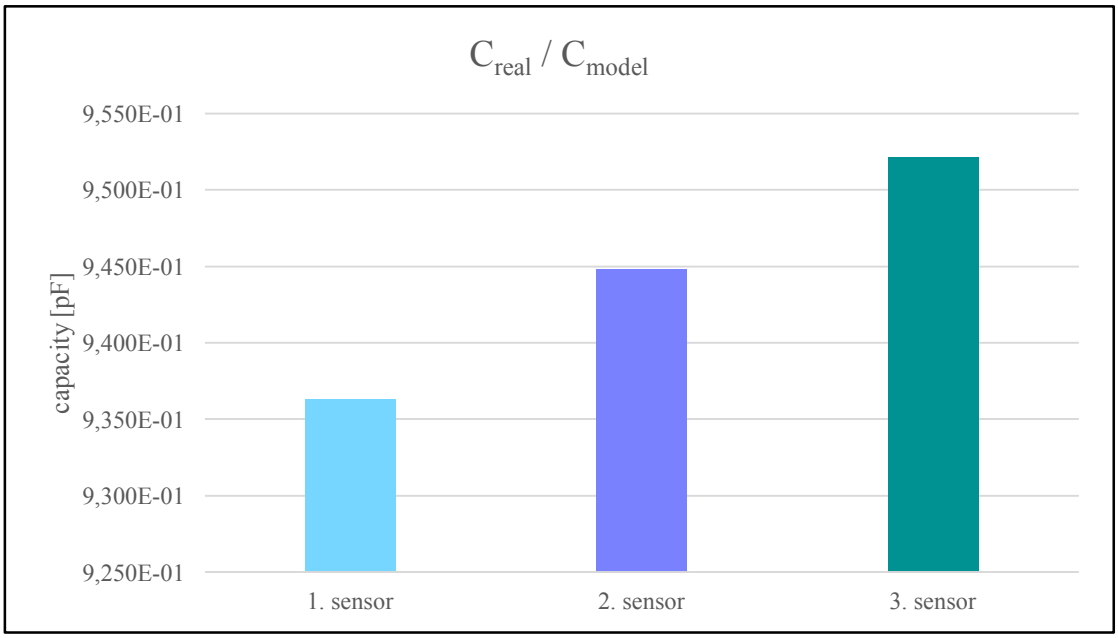

Figure 4-10: Ratio C_{real} to C_{model} of three sensors

4.2.1.5. Skin contact influence

Capacity measurements are very sensitive to all environmental influences, since the capacitor does not just measure the capacity created by itself, but also the capacity of the human body. Therefore, the contact with the skin will change the output.

	capacity [pF]		
sensor	initial geometry	skin contact	difference
1	18.40	17.58	0.82
2	20.05	19.03	1.02
3	18.35	17.46	0.89
4	18.22	17.37	0.85
		average	0.90

Table 4-6: Influence of skin contact

The dimension of this affection was examined and is descript in this chapter. The procedure to note the capacity value of the sensor at rest was simple as no pre-stretching was necessary and the only contact was with the worktable. Afterwards, the sensor was mounted on the skin, using the initial geometry. The results are shown in Table 4-6. As it can be seen, the difference created through the contact of the sensor with the skin, is around 0.90 pF.

4.2.2. Comparability Hydrogel-Sensor to Cutometer

The setup for these measurements is described in chapter 3.4.2. These tests were done to compare the measurements done with the Hydrogel-Sensor with the measurements done with the Cutometer. At first, the similar values of the Cutometer results regarding the outputs of the Hydrogel-Sensors need to be set. Two significant outputs, which are related at best to the principle of the Hydrogel-Sensor measurement are the R7 and R2. As described in chapter 3.2.5, R7 stands for the biological elasticity, whereas R2 is the gross-

elasticity including the viscous deformation. For the measurements done with the Hydrogel-Sensor, the measured ΔC and Δ*l* are given in Table 4-7 as well.

male, <30				
proband	Cutometer		Hydrogel-Sensor	
	R7	R2	ΔC [pF]	Δ*l* [mm]
1, l	0.650	0.892	2.00	3.60
1, r	0.650	0.904	2.14	4.25
2, l	0.543	0.837	1.34	2.85
2, r	0.588	0.873	2.14	2.54
3, l	0.662	0.863	1.08	2.50
3, r	0.673	0.856	2.24	2.40
average:	0.628	0.871	1.82	3.02
standard deviation	±0.051	±0.024	±0.49	±0.74

female, <30				
proband	Cutometer		Hydrogel-Sensor	
	R7	R2	ΔC [pF]	Δ*l* [mm]
1, l	0.564	0.848	1.50	1.80
1, r	0.628	0.885	1.20	2.50
2, l	0.728	0.946	1.65	1.90
2, r	0.695	0.935	2.05	2.50
3, l	0.683	0.893	1.89	2.40
3, r	0.715	0.880	1.90	1.30
average:	0.668	0.898	1.70	2.07
standard deviation	±0.062	±0.036	±0.31	±0.48

male, >50				
proband	Cutometer		Hydrogel-Sensor	
	R7	R2	ΔC [pF]	Δ*l* [mm]
1, l	0.561	0.919	0.98	2.46
1, r	0.562	0.856	1.08	3.99
2, l	0.561	0.938	0.46	0.91
2, r	0.594	0.957	0.79	0.99
3, l	0.521	0.807	1.09	3.50
3, r	0.546	0.824	1.65	5.60
average:	0.557	0.884	1.00	2.91
standard deviation	±0.024	±0.063	±0.39	±1.82

female, >50				
proband	Cutometer		Hydrogel-Sensor	
	R7	R2	ΔC [pF]	Δl [mm]
1, l	0,486	0,805	1,40	2,90
1, r	0,495	0,807	0,93	2,10
2, l	0,504	0,830	1,00	4,20
2, r	0,553	0,913	2,06	5,30
3, l	0,567	0,875	1,43	4,65
3, r	0,613	0,925	2,20	3,20
average:	0,536	0,859	1,50	3,73
standard deviation	±0,050	±0,053	±0,53	±1,20

Table 4-7: Measurement done with Cutometer and Hydrogel-Sensor for comparison. 4 test groups

In Table 1-1, it can be seen, that the results of the *Cutometer* measurements in all test groups differ after the second position after the decimal point. Therefore, the values for the males > 50 have the average of R2 of 0.884 with a standard deviation of ±0.063.

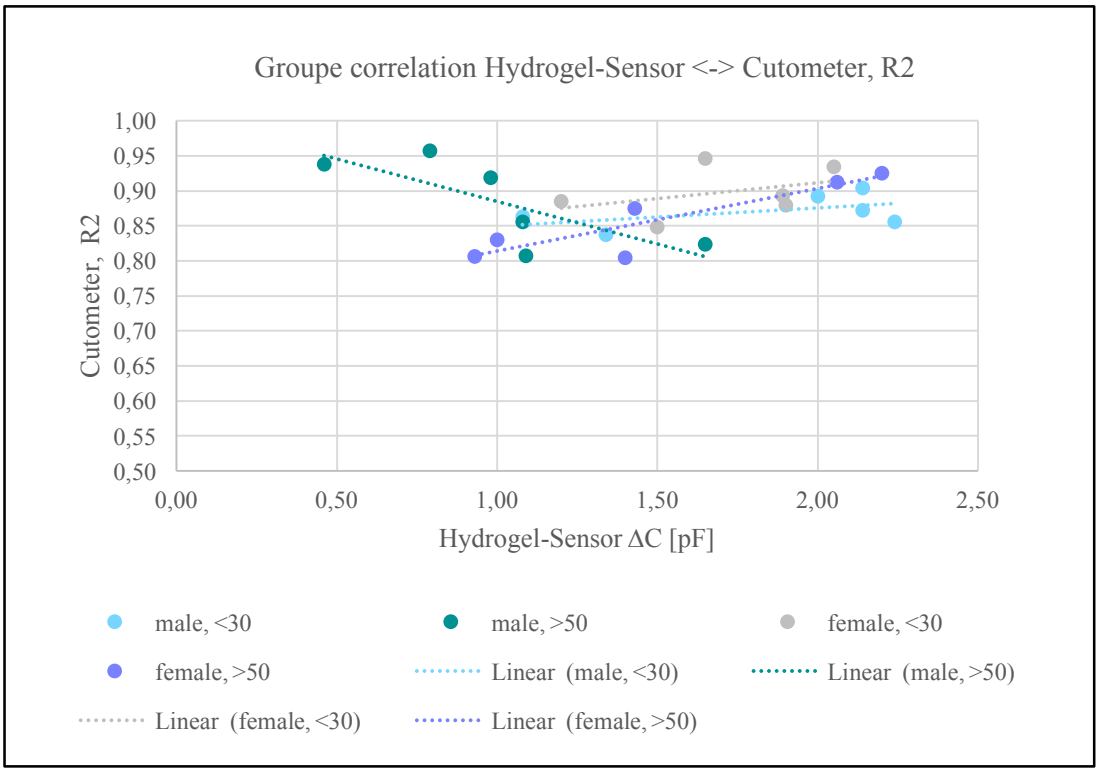

Figure 4-11: Correlation test group results measured with Cutometer and Hydrogel-Sensor. Comparison R2 with ΔC.

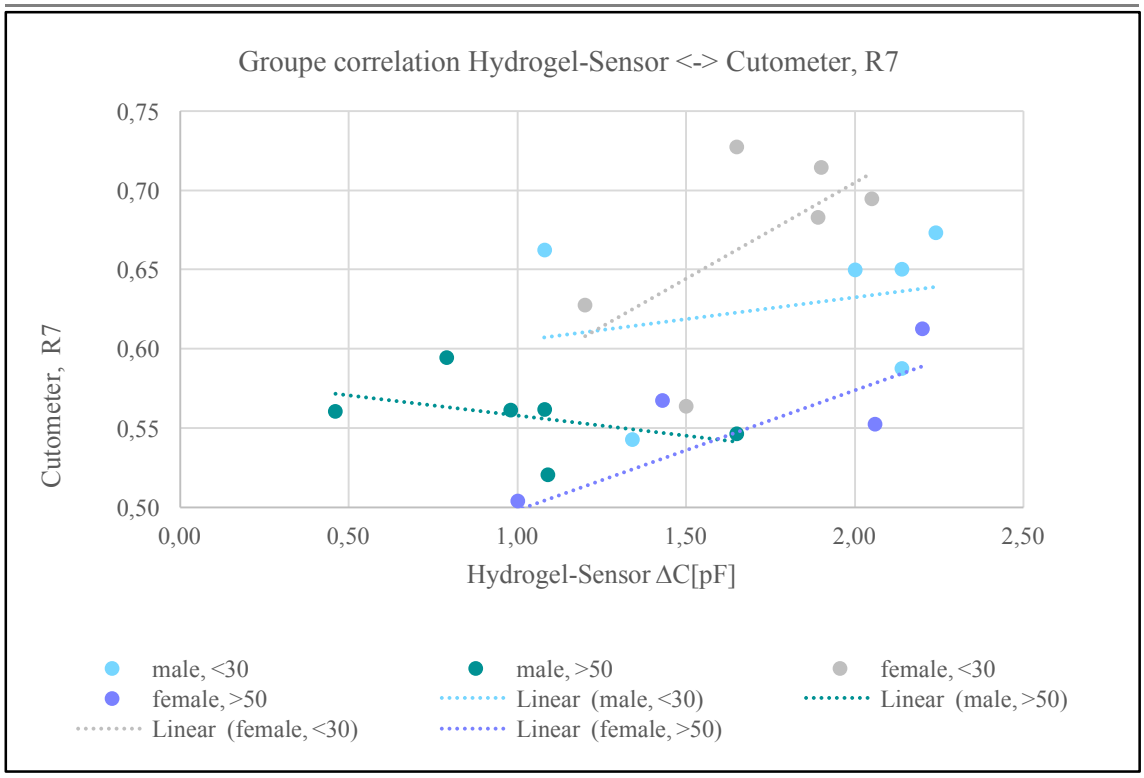

Figure 4-13: Correlation test group results measured with Cutometer and Hydrogel-Sensor. Comparison R7 with ΔC

In comparison, ΔC of the Hydrogel-Sensor on the same test group indicates the average of 1.82 ± 0.49 pF. For group 3, the results average of the *Cutometer* is 0.859 ± 0.053, compared to the examined method with 1,50 ± 0.53 pF.

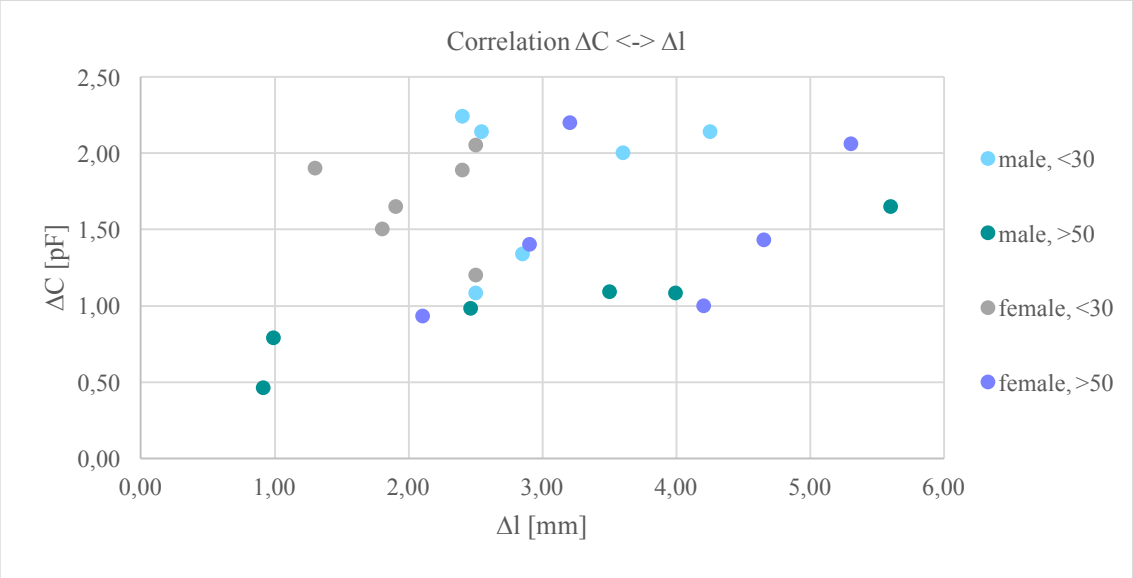

Figure 4-12: Correlation of ΔC to Δl of the test groups

For the first test persons, the values of the suction method resulted in 0.898 ± 0.036 and for the elastic sensors 1,7 ± 0.31 pF. *Test* group 2 demonstrates values for R2 of 0.871 ±

0.024 and ΔC 1,82 ± 0,49 pF. This points toward a result deviation of the Hydrogel-Sensor of the first position after the decimal point. In Figure 4-11 and Figure 4-13, the results of all test groups are presented. Thereby, the *Cutometer* values of R2 and R7 are put against the results of the Hydrogel-Sensors ΔC. In the last figures (Figure 4-12) the resulted ΔC values of all test groups is set against the associated Δ*l* values. This clarifies the relation of the examined measurement method results, the capacity and their length change and extended the examination of the Hypothesis H2 (chapter 2.) In order to oppose the measured outputs of both methods in a direct way, the averages and their deviations are given in Table 4-8.

	<30		**>50**	
	m	**f**	**m**	**f**
R2 [-]	0.871 ±0.024	0.898 ±0.036	0.884 ±0.063	0.859 ±0.053
R7 [-]	0.628 ±0.051	0,668 ±0.062	0.557 ±0.024	0.536 ±0.050
Δ*C* [pF]	1.82 ±0.49	1.70 ±0.31	1.01 ±0.39	1.50 ±0.53

Table 4-8: Direct comparison of the outputs of Hydrogel-Sensor and Cutometer

5. Discussion

The aim of this thesis is to find a simple way to measure the human skin elasticity, practical for the application within the usual use of unspecialized end users. As this work is the beginning of this project, the focus was on the finding process of potential measurement techniques and possible technical upgrades and, if necessary, an invention of new methods and principles. The use of the physical precept of length change suggests it self, because a common and often used relation, to define a certain mechanical strength is the Young's or elastic modulus. Another also very popular principle is the Hooke's law, which associates the length change with the applied force and the regarding elastic modulus of an examined solid material. This theoretical background brought up the idea to apply this simple principle to measure the mechanical properties, which implies the elasticity of the human skin on the wanted measurement method. The invention of the elastic capacitors of Keplinger and Sun allowed this method to transmit to the skin and create respective outputs at the same time. Properties like size, weight, handling and application seems to fit the defined requirements, as well. As the examination of the hydrogel (chapter 4.1.) shows, the recipe and therefore the mechanical properties of the gel itself can be easily changed and adjusted. For example, the elasticity of the gel influences the strength of the whole elastic capacitor and the measurement range of the sensor. The most common failure cases of the measurements done in chapter 4.1.2. were caused by the rupture of the acrylamide gel at certain weak points, as well as the overstretching of the hydrogel. Therefore, the change of the hydrogel strips regarding the firmness and geometry is inevitable.

Another influence factor influencing the elastic properties and the stability of the measurement technique is the utilized VHB tape. This double-sided tape was used after example of the paper of Sun and Keplinger. They already had good experiences with the use of the tape, primarily in the combination with the acrylamide gel. Thus, this tape is the main elastic component within the Hydrogel-Sensor and it generates the implied elongation force on the skin. Additionally, the VHB tape is usually used for construction measures and does not meet any medical legal guidelines of medical technologies. Therefore, not only the dimension of the elasticity, but also the quality of the tape adhesion on the skin is a variable influencing factor on the measurement outputs.

A further factor of uncertainty is the production process. It was all done by hand and with relatively rough tools. The used template for the cutting process of the already solidified gel, indeed is a good direction but because of the great slippage of the gel on the support, a certain divergence is possible. Therefore, an exact size and dimension of the hydrogel strips is not fully guaranteed.

As described in chapter 3.3.2., the measurement wire used for this method was made from aluminum. This metal counts to the base metals, which results in an oxidation with the contained salt in the hydrogel, and further, to a degradation of the conductivity. Because this process starts right after the connection of the wire with the gel strip, an influence on the sensitivity of the measurement method can be expected.

Furthermore, the stacking of the different layers of the flexible capacitor was also done by hand. Important for the range of the capacity output of every sensor was the resulting cross-sectional area, out of the exact size and centering of the hydrogel within the middle of the VHB tapes. The size of the overlapping area, created by the centered acrylamide strips, defines the active area, which is important for the capacitor properties. Since the exact building of the sandwich panel showed discrepancies, another influencing factor on the measurement results will have to be taken into consideration.

Regarding the results of Table 4-1, which are presenting 10 measurements done with one sensor, suggest, that the reproducibility of the results is given. Only 10 sensors were investigated, because this product is thought to be a disposable product, and therefore not made to be extended more than one time. The regarding standard deviation is ± 0.14 pF, presenting a value range from 3.50 pF to 3.92 pF. This confirms the hypothesis H0. Furthermore, it brings up the indication, that a Hydrogel-Sensor, even if there are previous mechanical impacts, still presents reliable results.

The investigation of the correlation of 10 different Sensors which is shown in Table 4-2, brings up, that the reproducibility of the outputs from sensor to sensor is slightly less precise compared to one sensor. Here, the values are at ± 0.32 pF, within a range of 3.23 pF to 4.30 pF, for an extension of 10 mm. The same extension was used in the previous examinations. At length change of 100% of the initial length of the hydrogel strip, there is a standard deviation of ± 0.52 pF. Additionally, some sensors could not even manage this extension and showed ruptures before reaching this length. It may be concluded, that an implementation of sensors with such geometry, cannot be stretched over a length of 50% of its initial length. However, considering the course shown in Figure 4-6, the results

are within a small distance. This means, even if it is a disposable product, the difference of the resulting elasticity measurements should be small enough, to bring significant coherent values.

The results shown in Table 4-3 present a nice linearity of all values, done with three sensors. As it can be seen, the outputs differ from outputs of the previous measurements with a magnitude of almost 10.00 pF. This can be explained with the different batch of hydrogel used and the thicker VHB tape. The distance between the conductive plates of the capacitor has a great influence on the measured capacity. In this case, the capacity got higher, which is apparently attributed to the different batch. The deviations are all within a small range, which guarantees a linear increase of the capacity values, in accordance to the corresponding length. The presentation of the capacity to the total length *l*, showed in Figure 4-7, points towards a direct correlation between *C* and *l*.

An interesting information from chapter 4.2.1.4. is, inter alia, the plausibility regarding the material parameters, e.g. the permittivity constant ε_r, and the correlation of the calculated capacity, based on this constant and the measured capacity. The values for ε_r, shown in Table 4-5, have different material constancies, shown in the literature. As comparative material, the permittivity value for rubber was used with ε_r = 2.5 – 3. The results for ε_r of the examined sensors were, for example, 8.61 and 11.38. The deviation is to declare with the enclosed secreted water ($\varepsilon_r \approx$ 81) between the VHB tape and the hydrogel strip. Additionally, some air ($\varepsilon_r \approx$ 1) inclusions are possible during the production procedure. [23] With the inserted permittivity values in the relevant formula (8), the calculated C_{real} could be achieved and are pictured in Figure 4-10. As this figure implies, the values differentiate at the second position after the decimal point, with around ± 0.008. This indicates, that the assumption, that the flexible capacitor elongates in a linear way, and the width $b_{real} = b_{model}$, can be supposed to be constant over the entire length, independent of the stretching length.

The last observations, were made on the application of the novel measurement method on real test persons. The results were compared with the outputs of the Cutometer, implemented on the same test persons. In order to see, if the Hydrogel-Sensor has a sufficient sensitivity, four different test groups were defined, that should present different specific elasticity values. As Figure 4-11 and Figure 4-13 displays, no correlation with the Cutometer values R2 or R7 could be demonstrated. Supplementary, Figure 4-12 does

not show a correlation between the capacity outputs and the caused changes of length. As can be seen, the single outputs per group are to find in an area of the diagram, that suggests a very small, but existing coherence between the values. The arbitrariness can be explained on the one hand with the inaccurate way of positioning the sensor onto the skin. The pre-stretching was done by hand, which creates big differences in the initial length from sensor to sensor. The flexible capacitor was stretched to the predefined length, drawn on the skin and subsequently mounted on the skin. Because the sensor starts to stick on the fingers, in the moment the sensor is stretched, the glued areas are already compromised. Thus, the adhesive force of the capacitor on the skin shows inequalities. Furthermore, this fact influences the precision of the equality of the sensors initial length. On the other hand, the Cutometer shows no connection between the outputs and the expected specifications for the different groups, likewise. Conversely, in comparison of the resulting averages for each group, done with the Hydrogel-Sensor and the *Cutometer*, the novel method demonstrates more clear graduation compared to the suction method (Table 4-8). This fact leads to the conclusion, that by changing the application process by itself, more accurate results are expectable. Because both methods show unspecific outputs, other influences need to be considered as well. Not only the measurement circumstances, but also the biological and physical parameters, influencing the skin elasticity are factors, that can change the results. As mentioned in chapter 3.1.4, sex, age, temperature, skin area and humidity are influencing the elasticity.

Out of the preceding conclusions, it can be said, that especially the first hypothesis H0 was absolutely confirmed and proven. Regarding H1, the results are at times contradictory. The examinations done with the setup described in chapter 3.4.1, are clear and can affirm the connection of C, Δl and E. In contrast, the results in Figure 4-12 are not clear and could disprove H0. However, to the mentioned explanations and uncertainties have to be considered. Hypothesis H2 was supposed to confirm a relation between the measurement results of the Hydrogel-Sensor and the Cutometer. It was expected, that a higher elasticity would lead to higher outputs of the suction method. Otherwise, with greater elasticity, the Δl and the ΔC should result at the measurement with the Hydrogel-Sensor. This could be confirmed in the first measurements done with the Hydrogel-Sensor, as explained above. The average for the measurement done in group 1, are compared to the values of group 3, higher, which would confirm the literature. The same observations could be made in group 2 and 4. However, the influencing parameters on such measurements are of great dimension, as explained above. In general, it is to

realize, that the measurement methods have different way of measurement direction. The *Cutometer* is a suction method, which displaces the skin out of its plane and in all directions. In contrast, the Hydrogel-Sensor elongates the skin only in one direction and within the skin plane.

6. Conclusion

Concluding, the idea to detect elasticity by measuring the length change done by a force impact, is already well investigated. The results of this thesis confirm the implementation of this method to the human skin. H0 could be verified as well as H1. Therefore, all the important requirements, regarding a reasonable measurement method are fulfilled with the Hydrogel-Sensor.

Nonetheless, several improvements are necessary. For example, to improve the sensibility and to direct the scale of the flexible capacitor, the recipe and the production process need to be changed and realised in a standardized procedure. One step would be, to directly pour the hydrogel onto proper formed areas to avoid the process step of cutting.

This brings us to the next implementation. The geometry of the sensor could be changed, to get a bigger cross-sectional area of the capacitor. Out of it, again the sensibility could be improved. Additionally, the failure cases may be minimalized, because weak-point within the gel are avoided.

The VHB tape shows many of the needed characteristics, necessary to realize the examined method. However, as this method is intended to be used on humans, it needs to fulfill the legal provisions for medical products. The double-sided adhesive tape used needs, which is flexible, biodegradable and skin-compatible.

Another major problem in this examination, with the most extensive uncertainties, is the way of stretching and mounting the sensor on the skin. As mentioned, it was all done by hand. Other ideas and methods to accomplish this measuring step need to be found. One idea is to bring the sensor in an already stretched state onto the skin. This could be implemented by a kind of template.

Hypothesis H2 could not be confirmed. One possible explanation is the difference in the deflection of the methods, as explained in the previous chapter. It could be considered, to compare the Hydrogel-Sensor with a more similar measurement method, for example, the extensometry (chapter 3.2.2). This method displaces the skin in the same longitudinal direction and works in the skin plane as well.

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