Battery-less Smart Diaper based on NFC Technology

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***Abstract*— Modern disposable diapers are carefully designed to transport fluid away from the skin to prevent irritation and dermatitis. These devices are used not only for babies. In fact, they are increasingly used for the elderly since both average life expectancy and the number of people who need special care are increasing. Diapers must be changed to prevent skin rash caused by the skin being exposed to wetness for a long time and to high the pH levels of urine. However, it is not easy to check the state of diapers (of babies or the elderly) without removing the baby's or the elderly's clothes. In this paper, we propose a smart diaper equipped with a battery-less Near Field Communication (NFC) tag. Moisture is detected by changes in capacitance between two electrodes located on the back sheet of the diaper, while capacitance is determined from the charge time through a high-value resistor using a microcontroller. The change in capacitance as a function of the wet conditions is simulated and measured. The tag is based on an NFC IC with energy harvesting. The power required to feed the electronics and microcontroller is obtained from the magnetic field generated by a smartphone with NFC used as a reader. A simple model is proposed to estimate the volume of urine in the diaper. We compare this approach with others in the literature and study how the capacitance is affected by the body and the materials.**

***Index Terms*—Near Field Communications, smart diaper, capacitive sensors, Radiofrequency Identification, moisture sensor.**

I. INTRODUCTION

THIS paper presents a smart diaper that includes a wet detector based on a battery-less Near Field Communication (NFC) tag. Diapers are used to absorb moisture in infants, adults, and animals. Nowadays, cloth diapers are replacing disposable ones. Diapers are used not only for babies but also for individuals such as the elderly who cannot control their bladders or bowel movements or who are unwilling to use the toilet. This includes people with medical conditions, such as the bedridden and those in a wheelchair. Average life expectancy is increasing. The elderly population has therefore grown and the number of people who require special care is increasing [1], [2]. It is difficult to check the state of the diaper in babies and the elderly without removing their clothes. Nurses and other caregivers spend a lot of time performing this task. Optimizing this process can therefore help to reduce the cost of health services. Diapers must be changed to prevent skin rash caused by the skin being exposed for a long time to wetness and the high pH levels of urine and feces [3]. It is therefore important to reduce the time that the moisture is in contact with the skin by periodically replacing the diaper [4]. Skin rash is often avoided by adding absorbent gels but these may have drawbacks such as irritation and allergies.

Several sensors have been proposed in the literature to detect the degree of moisture in diapers [5]–[14]. Diapers that incorporate these devices are often called smart diapers. Wearable devices are being progressively introduced into healthcare, so interest in these smart diapers is increasing. The most recent devices incorporate methods for wirelessly transmitting the state of the diaper to a mobile phone to set off an alarm [2], [15]–[17]. However, these communication functionalities increase the cost of the device. Moreover, they require batteries that need to be replaced or recharged. These batteries can also be a source of contamination if they are not recycled.

Several passive wireless sensors based on Radio Frequency Identification (RFID) have therefore been proposed for smart diaper applications [9], [10], [13]. Barcodes are progressively being replaced by RFID tags because of their low cost [18]. RFID is nowadays considered a mature technology for tracking items in logistics and a key technology in the expansion of the Internet of Things (IoT). One type of RFID that has received a lot of attention, thanks to its massive use in payment cards, is the NFC device at 13.56 MHz [19]. The number of mobiles that incorporate NFC readers is growing continually [20]. In addition to their traditional application in identification, advanced RFID tags with sensing possibilities have been developed in recent years. Due to their longer read range, active or semi-passive RFID sensors are preferred [21], [22]. Passive Ultra High Frequency (UHF) sensors with a read range of a few meters have been demonstrated in the literature [23]. However, the amount of power that can be extracted from a UHF interrogating waveform is limited to a few μW [24], which restricts the number of electronic devices that can be integrated. Recently available on the market is a battery-assisted passive (BAP) RFID tag IC operating at UHF band. One example is EM Microelectronics's EM4324, which can achieve sensitivities of up to -31 dBm in BAP mode. These devices open the door to long-range sensors such as those proposed in [25], where a resistive sensor for detecting glucose concentration is demonstrated. Chipless passive RFID

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sensors are another recent technology under research [26]–[28]. The main drawbacks of these sensors are the lack of standardization and the lack of low-cost commercial readers [28]. Other methods are based on detuning the antenna's characteristics [23], [30], the antenna load [31], and harmonic tags [32].

NFC is also an interesting technology for developing low-cost low-range sensors. The NFC integrated circuit (IC) can harvest a few mW from the reader's magnetic field. This is enough to feed low-power sensors [20]. The data from the sensors can be read with a smartphone without the need to pair the devices, simply by placing the reader near the tag. The main drawback of NFC-based sensors is the short-read range compared to sensors based on UHF RFID. However, an additional expensive reader, as in the case of UHF RFID, is not required because NFC readers are integrated into conventional smartphones. Moreover, the short-read range offers a degree of privacy and security from undesired readings by nearby third parties [33]. The most important NFC IC harvesting capability, therefore, is that it can provide energy to small sensors and microcontrollers [20], [34]–[38]. Consequently, the use of NFC systems within the Internet of Things (IoT) and Industry 4.0 is growing [39].

This paper focuses on the feasibility of developing a smart diaper that uses a battery-less Near Field Communication (NFC) tag. The tag is designed to be reusable in order to lower the cost of the system and can be installed in conventional diapers without modification. A smartphone with NFC capability can be used to read the tag without pairing delay or configurations with a simple tap reading. The moisture detector is based on a capacitive sensor. This sensor detects the change in capacitance between two electrodes. To our knowledge, this is the first NFC-based smart diaper that: 1) can detect urine without having to insert the tag into the diaper (simply by adhering it to the output layer); and 2) uses a capacitive detection method (which enables the volume of urine to be detected) rather than a resistive switch (which enables the detection of only two states, i.e. wet or dry). The main features of the smart diaper proposed in this study are detailed in section V, where they are also compared to other battery-less and battery-assisted devices in the literature.

This paper is organized as follows. In section II we describe the tag and the measurement method. In section III we present the simulated capacitance variations as a function of the moisture and dimension of the electrodes. In section IV we provide our experimental results. In section V we present a discussion and a comparison with previous approaches. And finally, in section VI we draw our conclusions.

II. NFC TAG WITH A CAPACITIVE MOISTURE DETECTOR

Capacitive sensing is becoming a popular alternative to other methods in applications for proximity detection, material analysis, and liquid level sensing. The main advantages of capacitive sensing are that it is a contactless, low-cost and low-power solution. Fig.1 shows a block diagram of the NFC tag embedded in the smart diaper. The capacitive moisture detector proposed in this study consists of two electrodes that sense the variation in capacitance between them due to the presence of urine. The electrodes are connected to a low-power microcontroller (Atmel ATTINY 85) that is powered by an NFC IC with energy harvesting capabilities. This study uses the M24LR04E NFC IC from ST Microelectronics. In the presence of a Radio Frequency (RF) field, the NFC IC rectifies the AC interrogating signal and converts it to a DC voltage that is used to feed the IC electronics and the microcontroller. The sensor data from the microcontroller is saved in the internal EEPROM of the NFC IC and read by the NFC reader (a smartphone with NFC). The I2C serial communication interface allows communication between the microcontroller and the NFC IC. The I2C interface consists of two wires, known as serial data (SDA) and Serial Clock (SCL). These are connected via pull-up resistors.

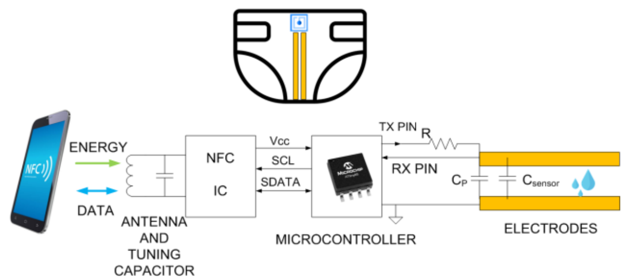

Fig.1. Block diagram of the smart diaper NFC tag.

A prototype of a flexible tag has been designed on flexible Rogers Ultralam 3000 substrate (ε_r=3.14, tan δ=0.0025, thickness 100 µm) [40]. The antenna is a 6-turn loop with an area of 25 × 25 mm. The width of the strips (*W*) and the spacing (*S*) between them are 0.5 mm. A prototype of an antenna was measured with a vector network analyzer, obtaining an inductance of 1.15 µH and a quality factor of 61 at 13.56 MHz in free-space. However, the presence of the smartphone's metallic case (when brought closer to the tag to read it) reduces the inductance to 0.9 µH and the quality factor to 48 and also detunes the antenna [41]. The influence of the body is low because the diaper acts as a spacer. The resonance frequency of the tag was adjusted by adding a 100 pF capacitor in parallel to the input of the NFC IC. Fig.2 shows the average magnetic field H_{av} in the tag measured as a function of the distance between the tag and the smartphone used as a reader. H_{av} is measured with the method described in [36]. The maximum read range for the harvesting output activation and therefore, up to the distance that the sensor can be fed, is 14 mm. This distance corresponds to a minimum magnetic field of 0.64 A_{RMS}/m. However, previously data saved in the NFC IC memory can be read up to 26 mm which corresponds to a threshold magnetic field of 0.21 A_{RMS}/m. The measurements have been performed with a Xiaomi Mi Note 2 smartphone where the NFC antenna is integrated around the camera and has a metallic case. A key commercial aspect is the cost of the prototype. For large quantities (>1000), the cost of the M24LR04E and the ATTiny85 is 0.3 $ and 1 $, respectively (prices have been taken from the digikey distributor). In commercial products, the Ultralam substrate can be replaced by another low-cost dielectric, such as polyamide. The cost can therefore be below 1.5$. The typical price of an adult diaper package in the supermarket, although

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it depends on the brand, is roughly 15 $ (with 10 units) and that of a baby's diaper package is roughly 20 $ (100 units). Therefore, the commercial device, at least the electronic part, should be reusable. Another option is for the electrode to be printed using conductive ink on the diaper surface and for the tag and microcontroller to be purchased as stand-alone parts.

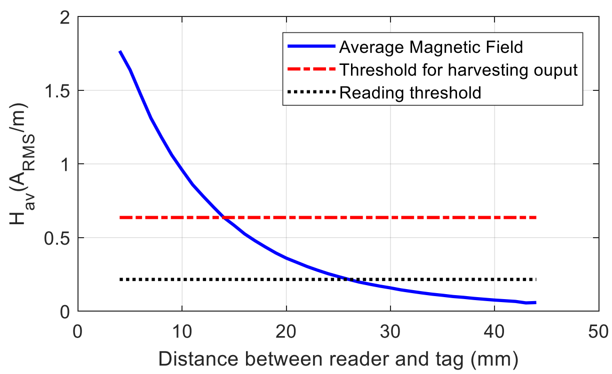

Fig.2. Measured average magnetic field as a function of the distance between the smartphone and tag. The threshold limits for harvesting output activation and reading range are shown.

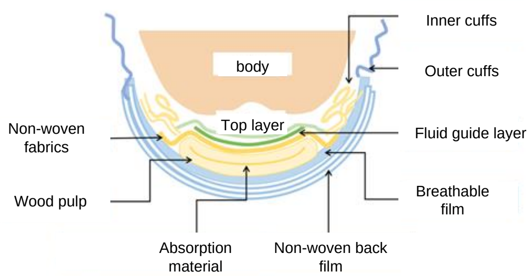

Fig.3. Cross-section of a diaper.

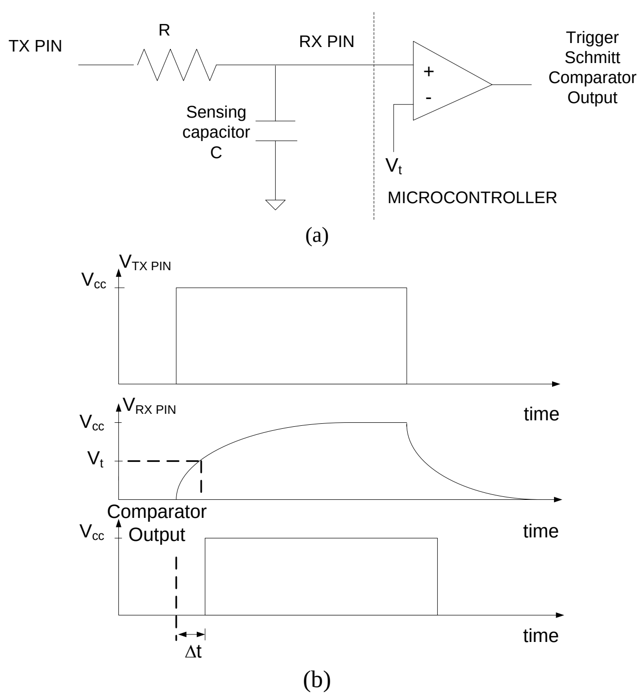

Fig.4. (a) Scheme of the circuit used for measuring the capacitance. (b) Waveforms at TX PIN, RX PIN and at the output of the internal microcontroller comparator.

Fig.3 shows a cross-section of a typical diaper made up of several layers [42]. The top layer is the one in contact with the skin. Its aim is to transfer the fluids to the core while remaining soft and dry to the touch. Some diapers include a skin-care lotion that protects the skin from over-hydration and reduces irritation. The layer between the top sheet and the absorbent core is called the acquisition layer. This channels the fluid away from the skin and spreads it over the entire diaper core for better absorbency. The diaper's most internal layer is the absorbent core. This generally consists of a blend of cellulose fluff pulp and polyacrylate (polyacrylic acid sodium salt, $(C_2H_3NaO_2)_n$) granules. The cellulose material quickly absorbs and transfers the liquid to the polyacrylate superabsorbent material, where it becomes trapped. This material is used because it can absorb and retain large quantities of liquid relative to its own mass. After it absorbs the liquid it solidifies into a gel. The diaper's outer layer is the back sheet, which is designed to be water-resistant to prevent the liquids from leaking onto the clothes. This layer is made of polyethylene or a cloth-like film that allows water vapor and air (but not liquids) to pass through, thus reducing moisture and keeping the skin drier. To avoid any modification of commercial diapers, the NFC tag was designed on a flexible substrate to be attached to the back sheet on the outer surface.

The aim of this study is to detect whether the capacitance between electrodes is higher than a threshold value that indicates the urine content in the diaper core. An absolute capacitance measurement is therefore not necessary and only a differential measurement is required. To increase the system's robustness, the power consumption of the microcontroller and sensors must be optimized. Although the NFC IC can provide up to 5 mA at the maximum magnetic field, it is better to reduce this value to enable small loop tag antennas to be used. To this end, we reduced the frequency clock to 1 MHz and achieved a maximum current consumption of 300 µA at 3 V. In addition, the measurement method should be implemented in a low-power microcontroller with a low-frequency clock (1 MHz in our case). In this study we used a capacitance measurement method that only requires the General Input/Output (GPIO) pins of the microcontroller and an external resistance *R*. It therefore does not add any active components that could increase power consumption. The method uses the charge time of the capacitive electrode, which is driven by a large resistor *R* (4 MΩ in the prototype) connected between a GPIO microcontroller digital output (TX PIN) and the electrode (see Fig.4). The microcontroller program toggles the TX PIN from a low to a high state, reads an input pin (RX PIN), and waits until its state is the same as that of the TX PIN. An integer variable is incremented inside a while loop until the receive pin's state changes. The output of this counter is therefore proportional to the delay between the states. The delay Δt between the TX PIN change and the RX PIN change is determined by the *RC* time constant, where *C* is the total capacitance at the receive pin, including the electrode and parasitic capacitances. The delay can be computed as:

$$\Delta t = -R \cdot C \cdot ln\left(1 - \frac{V_t}{V_{cc}}\right) \quad (1)$$

where V_t is the threshold voltage level that toggles a digital pin in the microcontroller and V_{cc} is the digital voltage power supply provided from the energy harvesting output of the NFC

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IC. If the RX PIN is configured as a digital input, this threshold corresponds to the logic threshold (usually $V_{cc}/2$ in CMOS). Another threshold voltage can be chosen if RX PIN is configured as an analog-to-digital converter input. The counter value (NFC reading) is proportional to the delay Δt and therefore, to the capacitance (1). However, in this application, the absolute value of capacitance C is not required because we are only interested in comparing the counter value between the dry and the wet states. The resolution is determined by the minimum time step the microcontroller can measure, which is limited by the clock frequency and delays in the internal microcontroller. Although our proposed method is simple, it provides enough precision for this application. The accuracy involved in measuring C can therefore be increased by increasing the resistor value. This method is often used successfully in low-cost touch-button libraries in microcontrollers to avoid the need to add specialized integrated circuits based on measuring capacitance using the charge transfer method [43]–[45].

III. Capacitance simulations

Several electromagnetic simulations were carried out to estimate the change in capacitance between the dry and the wet diapers. The S parameters of two coplanar electrodes were simulated with the Keysight Momentum simulator as a function of the frequency. The capacitance between the two electrodes was obtained from the imaginary part of the input admittance. Fig. 5. shows a simplified structure of the diaper with the coplanar electrodes. The height of the back sheet including the flexible PCB substrate is h_1. The permittivity of this layer (ε_{r1}) is assumed to be 3, which is the typical permittivity of polyamide. The permittivity of the absorption layer (ε_{r2}) is considered to be 2 when dry, which is the permittivity of the cellulose layers. The permittivity of urine is roughly ε_r=50, and the conductivity is 1.75 at 1 MHz [46]. For the laboratory experiments, urine was simulated with a solution of water with salt (NaCl). The relative permittivity of the solution is close to that of water (ε_{r2}=78) and the conductivity can be adjusted with salt concentration. In accordance with Stogryn's model [47], a solution of normality 0.175 N (43 g/l) at 25ºC produces conductivities that are close to those of urine. When the core layer is saturated with liquid, it is assumed that the permittivity is close to that of the liquid. The dielectric properties of the body are frequency dependent. At low frequencies, the body has a high permittivity. Permittivity is difficult to estimate because there may be a gap of air between the body and the top layer of the diaper, which reduces the effective permittivity. The maximum bandwidth of a pulse depends on its rise-time. However, it is considered to be 10 MHz (10 times the microcontroller's clock frequency). It is considered equal to 170 in the simulations, which is the value of muscle permittivity at 10 MHz [48]. Note that this layer is included in the simulation in order to study the effect of the proximity of the body and estimate the threshold capacitance between the dry and the wet states. Table I shows the main parameters used in the simulations.

The sensor's electrodes were designed with geometrical parameters selected to ensure sufficient sensitivity. When the diaper is dry, the effective permittivity of the multilayer structure is expected to be low and so a low capacitance value between electrodes is obtained. The effect of the high permittivity of the body should be small if h_2 is higher than the penetration depth of the fields, T (see Fig.5). However, when the diaper is wet, the urine is trapped in the absorbent core layer and the capacitance increases by several times. The field penetration depth of the sensor can be estimated using conformal mapping analysis (see for example [49]):

$$T = \frac{S}{2}\sqrt{\left(1+\frac{W}{S/2}\right)^2 - 1} \approx W + S/2 \qquad (2)$$

The penetration depth of the fields is therefore a function of the width and the spacing between electrodes. T should be of the order of the height h_2 in order to maximize the ratio between the wet and the dry capacitances.

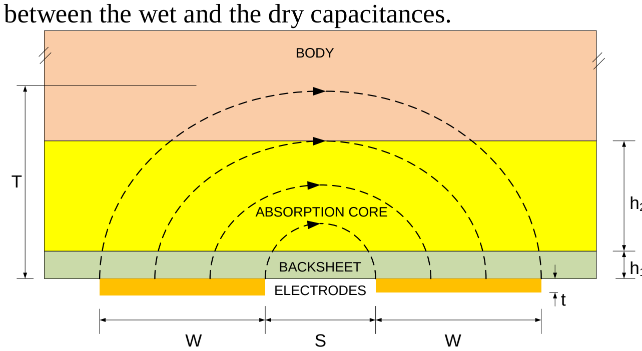

Fig.5. Cross-section used for capacitance simulation showing the electric field distribution inside the diaper.

TABLE I
PARAMETERS USED IN ELECTROMAGNETIC SIMULATIONS

Parameter	Symbol	Value
Permittivity of back sheet and flexible substrate	ε_{r1}	3
Height of back sheet	h_1	200 μm
Permittivity of the core (dry)	ε_{r2}	2
Permittivity of the core (wet with water)	ε_{r2}	78
Height of the core	h_2	5 mm
Permittivity of the body	ε_{r3}	170

Figures 6-9 show the simulated results in function of the main parameters. To avoid excessive bending of the electrodes and improve contact with the diaper, the maximum width of the electrodes is considered to be roughly a quarter the width of the diaper (2–2.5 cm). Two cases are considered in all figures: (1) the diaper is in contact with the body, and (2) the diaper is in the air with no contact with the body. The variation in the capacitance per unit length of the electrode (in pF/mm) between the dry and the wet states is analyzed according to one parameter, while the other parameters remain constant. Fig.6 shows that the change in capacitance increases when the width of the electrodes W increases. As expected, C increases because the area of the electrodes increases. The capacitance is not affected by the body for widths that are less than the height of the core layer (h_2=5mm). For greater widths, the field penetration depth is greater than h_2 and there is a dependence on the permittivity of the body. The increase in capacitance normalized with respect to the dry capacitance is shown in Fig.6b. For an absorption core layer saturated with water (ε_{r2}=78), the ratio of capacitances ΔC/C can reach values of 8. Fig.7 shows the capacitance as a function of the

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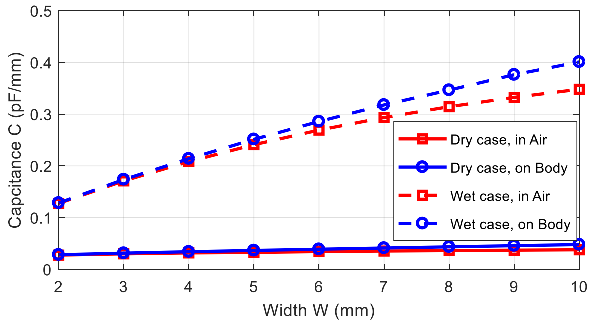
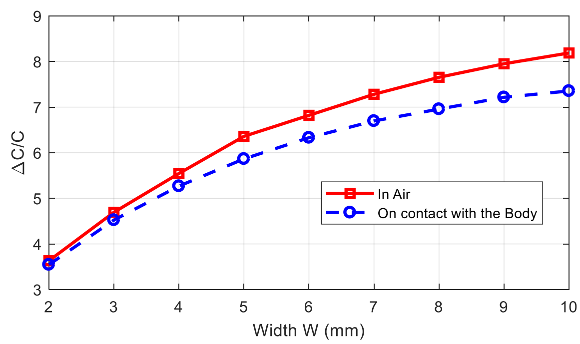
separation between electrodes *S*. As the separation is increased, the capacitance decreases steadily. *W* is assumed to be 5mm in this figure ($T<h_2$ with a small effect of the body). However, an increase in the relative capacitance occurs when *S* increases (Fig.7b).



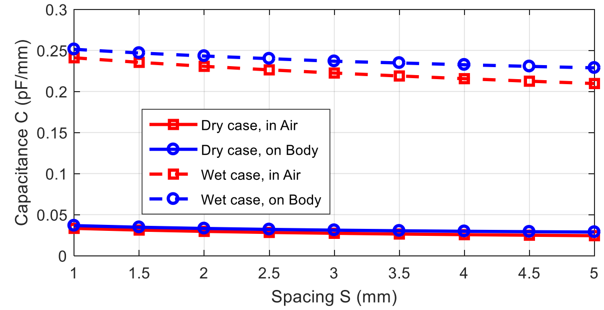
(a)



(b)

Fig.6. Simulated capacitance per unit length (a) and normalized increment in capacitance (b) between the wet and dry cases as a function of the electrode width *W* with the diaper in contact with the body and the diaper in the air (*S*=1 mm, h_2=5 mm).

Fig.8 shows relative capacitance as a function of the permittivity of the absorption layer. This permittivity is a function of the urine content, the pressure and the diffusion of the urine along the electrodes. Capacitance is expected to behave linearly with permittivity. However, this linear effect is observed only up to 10. This effect is due to the curvature of the field lines for high values of the dielectric constant in the interface between the back sheet and the core. This behavior is not often taken into account in analytical formulae derived using conformal mapping in the literature [50] but it can be computed numerically with electromagnetic simulators. These formulae are often used and verified for a low permittivity range, in which they are accurate. However, this conformal mapping does not take into account this strong field change. The same behavior has been found in interdigital electrodes used in measurements of soil moisture [35]. It is important to note the strong dependence of the normalized increment of capacitance on the effective dielectric permittivity of the absorption layer. Finally, the influence of the height of the absorption layer is shown in Fig.9. If the height h_2 is higher than the field penetration depth, the capacitance remains constant. For lower heights, *T* increases. For thicker layers, the normalized increment is almost constant. The height of the absorption layer depends on the diaper manufacturer and plays an important role in the price of the diaper and the maximum volume of liquid the diaper can absorb. The absorption material increases in volume when the trapped urine content increases. Therefore, the diaper is thicker when it is wet than when it is dry and the cellulose layers are compressed.



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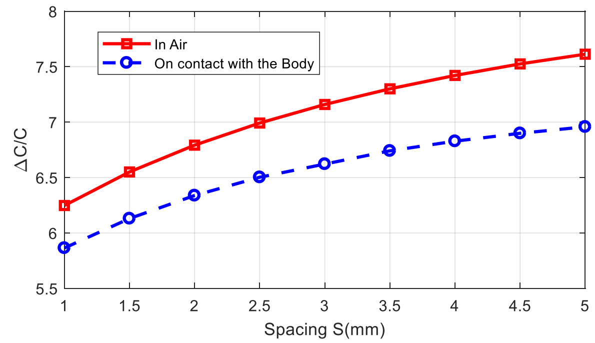
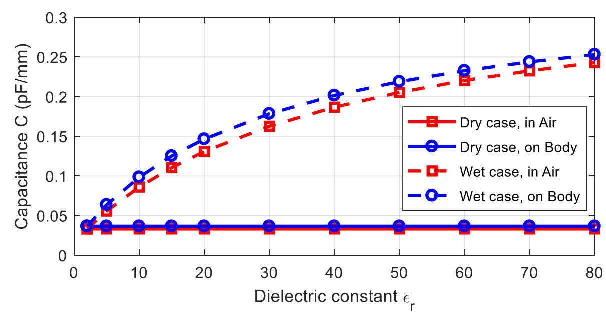


(b)

Fig.7. Simulated capacitance per unit length (a) and normalized increment in capacitance (b) between the wet and dry cases as a function of the electrode spacing *S* with the diaper in contact with the body and with the diaper in the air (*W*=5 mm, h_2=5 mm).



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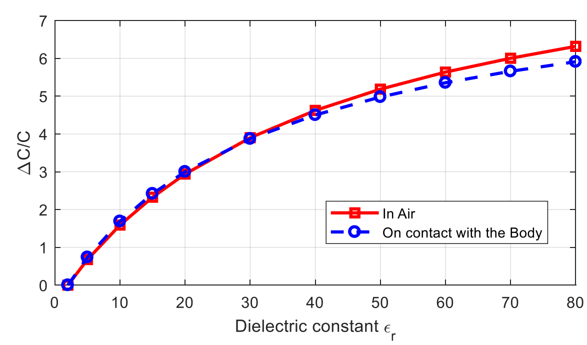
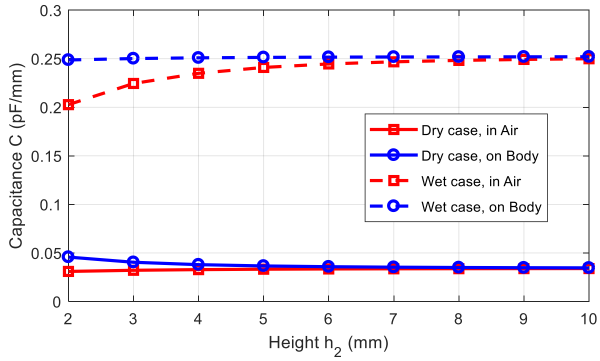
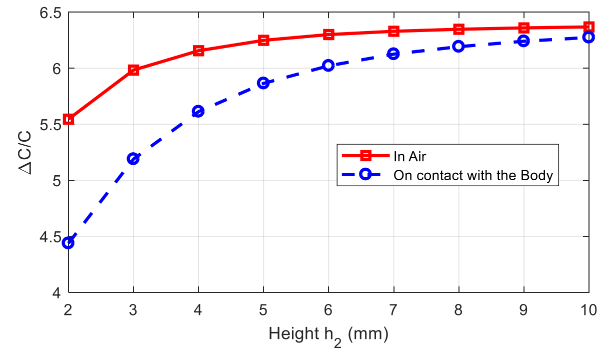


(b)

Fig.8. Simulated capacitance per unit length (a) and normalized increment in capacitance (b) between the wet and dry cases as a function of the effective permittivity of the absorption layer with the diaper in contact with the body and with the diaper in the air (*W*=5 mm, *S*=1 mm, h_2=5 mm).



(a)



(b)

Fig.9. Simulated capacitance per unit length (a) and normalized increment in capacitance (b) between the wet and dry cases as a function of the height of the absorption layer with the diaper in contact with the body and with the diaper in the air (W=5 mm, S=1 mm).

IV. Experimental Results

To check the accuracy of the proposed capacitance measurement method, we conducted a set of measurements. Fig.10 shows the count value measured by the tag when a discrete Surface Mount Device (SMD) capacitor was used instead of the electrodes. A resistor of 5 MΩ was used in these measurements. A linear regression was performed and linear behavior was obtained. The error between the model (line) and the measured values was less than the tolerance of the capacitors (5%). The repeatability of the measurement was below 1 count or 0.18 pF. The measurements were read by the NFC with two mobiles and the readout was identical. The offset error associated with parasitic capacitances and systematic errors was 24.2 pF. This accuracy is enough for the threshold capacitance detector.

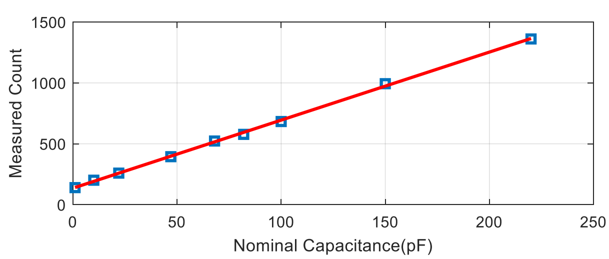


Fig.10. Measured count with the NFC as a function of a discrete SMD capacitor C.

Fig.11. Layout of the tag and the screen of the mobile app.

Fig. 11 shows a photograph of the tag, a photograph of the tag when adhered to a diaper, and several screens of the mobile APP designed to interface with the user. The experiments were performed with different diaper models and a saline solution (0.175M) was used to simulate urine. Fig.11 shows the sensor response as a function of time to a volume of 120 cm^3 injected uniformly with a pump to simulate urine discharge. We chose an adult night diaper. We can see that the capacitance, and therefore the counter output (NFC reading), increases rapidly. However, there is a transitory period of roughly 100 s before the value stabilizes to the end value due to the diffusion of the liquid in the core. The rise time is of the order of 3.66 s, which demonstrates that the diaper can absorb urine flow rates of 33 ml/s. This is higher than the peak urine flow rate of roughly 27.3 ml/s for normal adults [51].

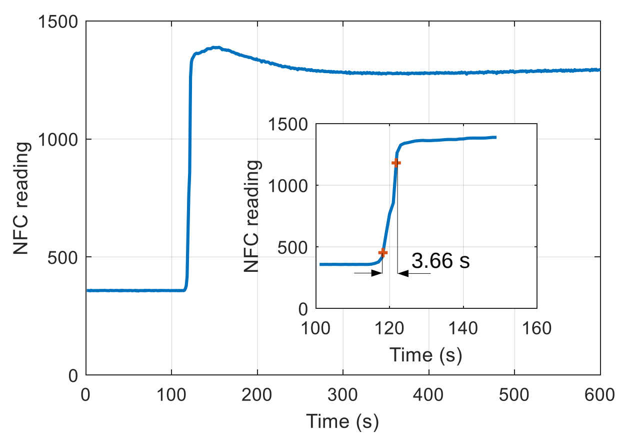


Fig.12. Time-dependent sensor response to a 120 cm^3 step injected into the diaper. The inset shows the rise time response between 10% and 90% of the end values.

Fig.13. shows the raw data output of the NFC sensor (output of the time counter) as a function of the volume of saline water. Two electrode widths are compared (W=5 mm and W=10 mm). The capacitance is proportional to the length of the electrodes. However, these electrodes must cover the active region where the absorption material is located. The loop antenna is located in the belt region to make reading easier. In the experiments, the spacing (S) and length (L) of both electrodes were 5 mm and 20 cm, respectively. The

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increase in capacitance between the dry and the wet diapers was roughly 1.5 times higher when W=10 mm than when W=5 mm, which is in agreement with the simulations in Fig.6. Fig.14 compares the measurement of the NFC sensor and the capacitance of an LCR meter (Keysight U1733C) at 100 kHz. According to the manufacturer's datasheet, the accuracy of the LCR meter is 2%. The capacitance measurement error is therefore between 0.14 pF and 8 pF. A correlation coefficient of 99.5% was obtained between the two measurements. The small discrepancies along the line were mainly due to the diffusion of the water in the absorption layer. This effect is especially visible when the liquid concentration is low and the liquid does not cover all the electrode area. To mitigate this effect, the measurements were taken 5 minutes after the liquid was poured onto the diaper in order to ensure the diffusion of the liquid into it. A range of measured capacitance variation between 7 pF and 395 pF was observed for water volumes between 0 and 400 cm^3 when W=10 mm. Our simulated and experimental results show that wider electrodes are preferable because they provide higher capacitance values and a higher capacitance ratio between saturated and dry states. Also, it is easier to cover the surface covered by the absorption layers. Depending on the number of absorption layers in the core, the Super Absorbent Polymer (SAP) mass ratio and fluff pulp used [52], and the expansion of the diaper in the wet state, the capacitance (and therefore the NFC reading) varies as a function of the model. Diapers designed to contain high volumes (up to 1000 cm^3) can support several discharges during the night and contain several absorption layers. Diapers for babies can support up to 400 cm^3–500 cm^3 and the height is smaller than those designed for adults at night. However, when the volume reaches 200–250 cm^3, the urine covers all the area and the absorption layers start to expand, thus increasing the height of the diaper. The capacitance then tends to saturate, as we can see in Fig.8, where the capacitance remains almost constant when the height is greater than the penetration thickness. In the case of electrode 1, the resolution is roughly 0.16 cm$_3$ for the adult night diaper and 0.8 cm^3 for the baby's diaper. These resolutions are several times lower than for a normal urine discharge.

Fig.13 shows that the readings (and capacitance) increase steadily as the level of liquid increases until saturation is reached and the diaper is considered full. However, the values tend to saturate when the core is saturated with liquid. The capacitance or NFC reading can therefore be used to estimate the volume of liquid in the diaper. We therefore propose the following model:

$$V(\%) = 100\frac{X^{\alpha}-X_{dry}^{\alpha}}{X_{sat}^{\alpha}-X_{dry}^{\alpha}} \quad (3)$$

where V is the percentage of liquid volume in the diaper, X is the normalized NFC reading, and α is the slope in logarithmic scale. X_{dry} and X_{sat} are the normalized NFC readings for the dry and water-saturated cases, respectively. Note that parameters X_{dry}, X_{sat} depend on the diaper model but they can be saved in the NFC message that is sent to the mobile.

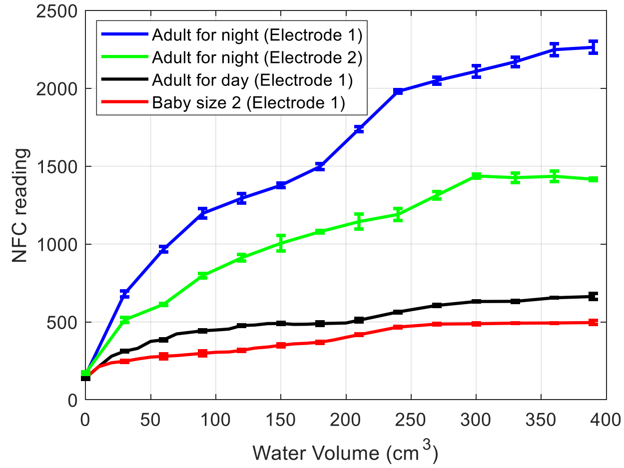

Fig.13. NFC reading as a function of the level of saline water for two electrode widths (Electrode 1: W=10 mm, S=5 mm, L=20 cm; and Electrode 2: W= 5mm, S=5 mm, L=20 cm) and different diapers (adult for night, adult for day, and baby size 2). The standard deviations are shown in the errorbars.

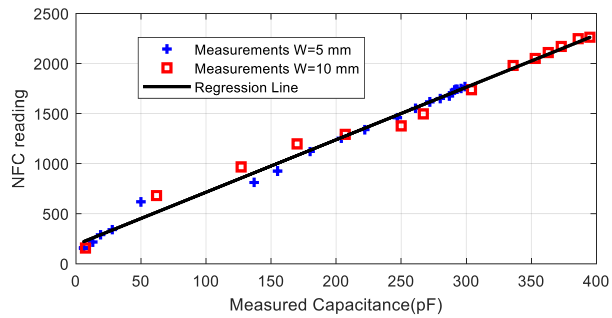

Fig.14. Comparison of the capacitance between the electrodes and the NFC reading for two electrode widths (W=5 mm and W=10 mm). The length of the electrodes is L=20 cm and the electrode spacing is S=5 mm. A regression line is included. The correlation coefficient is 99.5%.

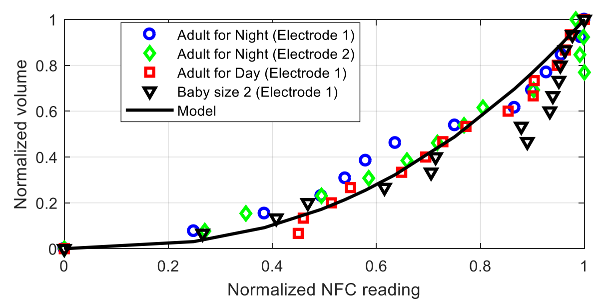

Fig.15. Comparison of the models proposed for the normalized volume (3) as a function of the normalized NFC reading with the measurements for the diapers in Fig.13.

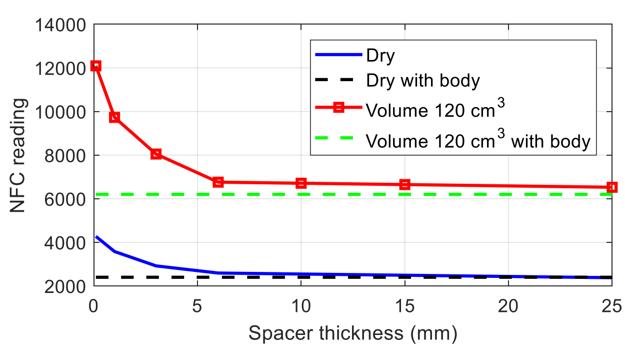

Fig.16. Comparison of the NFC readings as a function of the distance to metal. The dashed lines show the values measured in contact with the body.

Fig.15 compares the proposed model (3) and the measurements for the diapers in Fig.12. Agreement was good for α=2.5. The correlation coefficient between the model given by (3) and the measured normalized volume was between 96% (baby diaper size 2) and 99% (adult day diaper). The standard deviation of the error in the estimation of the volume was between 4.1% (adult day diaper) and 10.3% (baby diaper size 2). Although, as we can see from Fig.13, the dry and saturation values are a function of the diaper model, these values can be stored by the diaper manufacturer in the EEPROM of the NFC IC and saved in the NFC message read on the smartphone app.

Normal urinary frequency depends on how much fluid is taken in a day, the composition of the fluids taken, and whether the person is taking medication. For most people, the normal number of times to urinate per day is 6-7 over a 24-hour period [53]. Normal urine output depends on age [51],[53]-[54]. The normal output is 2–3 cm^3/kg/hour for neonates, 1–2 cm^3/kg/hour for infants, and 0.5–1 cm^3/kg/hour for adolescents and adults [53]. Urine outputs below these limits are considered oliguria [53]. For an adult who weighs 70 kg, a normal urine output is therefore between 120 cm^3 and 280 cm^3. As adult night diapers are designed for a maximum capacity of 1000 cm^3, they can support more than two discharges. Fig.13 shows that for volumes above 250-300 cm^3, the reading is almost constant, thus indicating a risk of overfilling. For a neonate who weighs 4 kg, normal output is roughly 30–50 cm^3. Our experimental results in Fig.13 show that these threshold volume values may be detectable.

To investigate the robustness of the readings when external materials are found below the diaper, we conducted several experiments. The worst case is the presence of metal under the diaper (e.g. a metal chair), which can increase the capacitance between the electrodes. Fig.16 compares the NFC readings as a function of the low permittivity spacer (cartoon, $\varepsilon_r \approx 1.8$), which simulates the thickness of clothes or chair cushion. An adult night diaper with electrode 1 was used in Fig.16, but similar conclusions have been found when other diaper models were used. As a reference we also include the measurement taken when the diaper was in contact with the body. In the measurements in Fig.16, a phantom consisting of a thick block of foam weighing 4 kg was used to simulate body pressure. In agreement with simulated results, differences of roughly 4.5% were observed between the low permittivity phantom and the body. From Fig.16 we can conclude that a spacer of roughly 5 mm is sufficient to offset the effect of the metal. However, even when the metal is very close to the electrode, it is possible to at least determine whether the diaper is dry or wet. Maximum differences of 1.1% were observed between the diaper when flat and when it had a curvature radius of 8 cm. Therefore, if the electrode is attached correctly to the diaper surface, the effect of bending is not very significant. The repeatability of the measurements is illustrated in Fig.17, which shows the histograms of 100 measurements for five diapers of the same model (adult night diaper) with the same volume of liquid (120 cm^3). The difference in the measurements for each diaper is less than 0.5% and the maximum difference with respect to the mean is ±2%. The measurements (in steps of 30 cm^3) have been repeated for five diapers and the standard deviation is shown in the error bars in Fig.13. Typically, the standard deviation is below 5% of the mean value in all the range for all diapers models. These differences can be attributed to small differences due to the location of the electrode, the distribution of the liquid in the absorption layers, and the surface roughness.

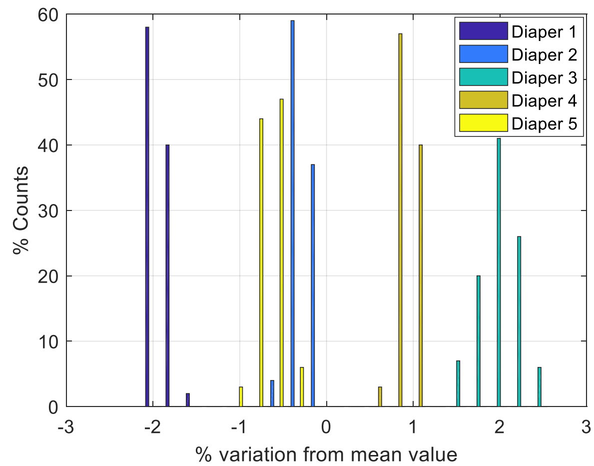

Fig.17. Histogram of variation in NFC reading from the mean for five adult night diapers with a volume of 120 cm^3.

V. Comparison with Other Technologies

Table II compares different passive (battery-less) and active smart diapers that use different technologies. Some techniques for sensing moisture in diapers are based on detecting changes in the color of a reactive in contact with urine. Such techniques are low-cost but they cannot detect the state of the diaper under clothes. However, because of their simplicity these technologies are commercially available in some diapers. Moreover, by using a nitride, a protein and leukocyte reactive, they can detect the presence of certain infections based on a change in color [55], [56]. Other studies propose passive tags that are detuned in the presence of moisture. A UHF tag was used in [13], whereas an HF tag was used in [5] and [10]. The main drawback is that the read range is reduced in the wet state. This reduction may also be due to other reasons, such as misalignment between the reader and the tag antennas. Moreover, in [10], the tag must be inserted in the core layer. In [5], an NFC tag was used with energy harvesting based on the SL13A IC from AMS, which uses a resistive switch. The main drawback here is that the electrodes must either be inserted into the core layer (perforating the back sheet) or be in contact with the skin. In both [10] and [5], the commercial diapers must be modified and only a binary reading (wet or dry) can be made. In [7], the wet stage produces a shift in the resonance frequency of an LC resonator that is read with a non-standardized custom reader.

Systems based on active devices have longer read ranges than passive alternatives. However, a battery is needed and so the cost is higher. The first devices reported in the literature,

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such as [6] and GSM devices [12], used a custom TX transmitter. However, these communication systems do not usually have low power and battery life is limited. With the appearance of smartphones with Bluetooth Low-Energy (BLE), several approaches based on this technology have recently been developed [17], [57], [58]. However, most of these approaches are based on the resistive switch and so an electrode must be in contact with the core of the diaper. In [17], a method based on measuring the external temperature of the diaper was proposed. This noninvasive method can detect a rise in the temperature of the outer sheet layer as a result of urine accumulation at a temperature above ambient temperature. Unfortunately, this detection method cannot be applied to battery-less devices such as the one proposed in this paper since it requires the temperature to be monitored periodically. However, it can be used if a battery is included as in a data logger. Unlike the method we propose in this study, the smart diapers reported in the literature provide information only if the diaper is dry or wet and do not estimate the volume of liquid that can be used to determine certain pathologies associated with oliguria [59]. The device proposed can be used to introduce small games in combination with mobile apps oriented to control the urinary continence in children at night.

TABLE II
COMPARISON OF SMART DIAPER DEVICES

Year/Ref.	Sensor technology	Communications technology and comments
This work	Capacitive	NFC based on M24LR04E, estimates the volume of liquid, do not require modify the diaper
Yambem et al. 2008 [7]	LC resonance shift	Inductive coupling, only dry or wet states detection, modification on the diaper is required, custom reader
Siden et al. 2011 [5]	Resistive switch	NFC based on SL13A, only dry or wet states detection, modification of the diaper is required
Sajal et al. 2014 [13] Nilsson et al. 2011 [9]	UHF antenna detuning	UHF RFID, measurement based on the horizontal and vertical polarization reading distances, only dry or wet states detection
Chen et al. 2014 [54] Karlsen et al. 2014 [14]	Colorimetric	Not available, It cannot be used with clothes
McKnight et al. 2015 [11]	Paper based biopotential	BLE, requires external amplifiers and impedance analyzer (AD5933)
Ziai et al. 2015 [10]	Resonance frequency detuning	Passive, HF at 13.56 MHz, only dry or wet states detection
Siden et al. 2004 [6]	Resistive switch	Tx at ISM 13.56 MHz activated by the sensor switch and custom reader
Simik et al. 2014 [12]	Resistive switch	GSM, only dry or wet states detection, modification of the diaper is required
Khan et al. 2018 [17]	Temperature	BLE, requires continuous monitoring of the temperature
Rahman et al. 2017 [57]	Resistive switch	BLE, only dry or wet states detection, modification on the diaper is required
Yu et al. 2016 [58]	Resistive switch and colorimetric nitride sensor	BLE, urine-activated urinary nitride sensor

Unlike other HF RFID solutions for smart diapers (such as [7][10]), our proposed method is compatible with a commercial smartphone and does not require a specific reader.

One important point is the safety of the system, especially when it is used with babies. According to the ISO 15693 standard, the maximum operating field is 5 A_{RMS}/m. Assuming that the tag is illuminated sporadically for about 1 second, the average exposure of the body over 6 minutes will be 0.0139 A_{RMS}/m which is less than the reference level for human exposure to RF (0.0728 A_{RMS}/m) recommended in [60]. In addition, Fig.2 shows that the induced magnetic field decays very fast with the distance (as the inverse of the square of the distance) and therefore, the level on the body will be lower than the limit. Therefore, the RF levels can be considered safe for the human.

VI. CONCLUSIONS

In this study, we have designed a prototype for a smart diaper that measures moisture by measuring the change in capacitance between electrodes. Capacitance is noninvasively determined by measuring its charge time when driven by a high-value resistor using a microcontroller. The tag is based on an NFC IC with energy harvesting. The power needed to feed the electronics and the microcontroller is obtained from the magnetic field generated by the reader, a smartphone with NFC. As an alternative to urine detection based on a resistance switch, we propose a tag based on capacity sensing that can be adhered to the outer layer of commercial diapers. With a resistance switch, the tag must be placed inside the diaper – a more invasive approach that requires either the manufacturer to make a special modification to the diaper or the tag to be placed in the inner sheet (which must be separated from the skin to avoid irritation). The proposed moisture measuring method can be implemented in the internal microcontroller of a Bluetooth LE module, in applications that require overcoming the inherent limitations of reading range of NFC technology. However, in this case, the cost of tag increases and needs a rechargeable battery and furthermore the privacy is lower than in the tag based on NFC technology.

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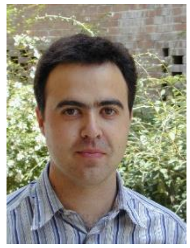

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