# Epidermal RFID Passive Sensor for Body Temperature Measurements

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***Abstract*—Real-time and continuous wireless measurement of human body temperature could enable a better control of many pathologies such as the wounds infection after surgery and the evolution of epidemics involving fever rush, as well as the monitor of athletic activities. This paper describes an RFID passive UHF epidermal sensor suitable to be directly attached onto the human skin by means of a bio-compatible transpiring Poli($\varepsilon$-caprolacton) (PCL) membrane. The antenna elements provide a broad matching band and even a post-fabrication tuning mechanism to better manage the specific placement over the body. The temperature is directly measured by the EM4325 microchip, also providing RFID communication capabilities. The epidermal sensor, that can be read up to 35 cm in case of 0.5 W EIRP emitted by the reader, has been moreover thermally calibrated versus a thermocouple and then applied to the measurement of human body temperature in both static and dynamic conditions with an accuracy of about 0.25°C with respect to reference measurements.**

## I. INTRODUCTION

The wireless and continuous measurement of human body temperature may open new frontiers in the monitor and control of pathologies involving fever rush, such as the SARS epidemic [1], and to take care of the skin infection around wounds and lesions. Moreover, variations of skin temperature are common indicators of brain activity and of particular psychological states. Long-term assessment of skin temperature can also provide accurate profiles of the circadian-system activity (for instance skin warming is functionally linked to sleep propensity).

Current technologies to quickly and non-invasively collect the body temperature are mostly based on non-contacting infrared thermometers (NCIT) [2], [3] that however require to place the instrument's probe very close to the body district. These devices moreover need the effort of an operator, thus generating a huge workload for nurses and clinicians in case of concurrent intensive temperature sampling rate of many patients. Automated wireless monitoring of temperature would instead enable data readings to be automatically stored, retrieved, and analyzed for trends and statistics. Battery-assisted temperature autonomous sensors for both clinical and personal use are nowadays being commercially available [4], but the involved high costs and the not negligible volume (mainly due to the presence of battery) are currently an obstacle for their true diffusion as disposable and nearly transparent devices able to conformally adhere to the human skin.

Different measurement options may originate from the *epidermal electronics* (or skin electronics) [5], [6], a new cross-discipline merging together the material science, the conventional electronics and the communication background. Conformal devices are fabricated directly onto thin and bio-compatible membranes for direct application over the human body, like a tattoo, with the purpose to collect human parameters like pressure, blood glucose level, heartbeat, temperature and so on. Most of the presented works are related to the design of materials and of electronic components, while less attention has been devoted till now to the efficient communication of the epidermal object with the interrogating device. A few papers have however addressed the issue of tattoo-like antenna design for UHF Radiofrequency Indentification (RFID) which can be considered as the natural technology to interact with epidermal device from a distance large enough to avoid a near-contacting interrogation. In particular, [7] describes a substrate-less nested slotline antenna suitable to direct application on the skin, i.e. without any dielectric substrate or air gap between the tag metallization and the skin. The proposed prototype, made of conductive paint profiled using the stencil techniques, provides a useful 80-120cm read ranges. The same authors in [8] presented a mouth-mounted passive RFID tag which acts as a tongue-touch controlled switch for assistive devices, helping patients with severe movement impairments.

This paper investigates the possibility to collect the body-temperature by using the epidermal UHF RFID technology in a fully passive mode with particular attention to provide the tag with re-tuning capability after manufacturing at the purpose to improve the read distance depending on the placement site over the body. The tag is fabricated onto a synthetic skin, already used in biomedical applications, with remarkable body-compatibility features. The temperature sensing mechanism relays onto a new family of RFID microchips equipped with an integrated temperature sensor and providing the temperature data straight away in a digital form. An experimental champaign is presented at the purpose of sensor calibration and experimentation in real conditions.

## II. TUNABLE EPIDERMAL TAG

Measuring temperature by an RFID tag attached onto the body requires the tag itself not to modify the local temperature of the skin. In other words, the epidermal tag should preserve the natural transpiration of the skin. At this purpose the substrate has to be not only bio-compatible but even transpiring and, for the same reason, the amount of antenna conductor

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has to be minimized. Therefore, among some possible options for wearable antenna layouts, such as the dipole [9], [10], the slotted patch [12], [11] and the loop [13], the latter configuration has been considered throughput the paper for the required small amount of conductor and for the known superior radiation performance close to lossy materials (as the human skin is). In particular, the sensor layout is derived from a one-wavelength rectangular-loop excited by a smaller inner loop whose shape factor is properly selected to achieve conjugate impedance matching to the microchip (Fig.1). The external dimensions are fixed to $L_1 = L_2 = 50mm$ (like a medium-size plaster) in order to meet a possible trade-off between the antenna gain and the overall size of the sensor, that should be as low as possible for user's comfort. The two loops include additional elements (in darker gray) that permit to increase the antenna bandwidth and to slightly modify the impedance matching of the tag (for instance to compensate possible changes depending on the placement region and on the different body mass of users). To clarify this point, let assume the tag without the tuning elements (the only shape in dark grey in Fig.1) to be connected to a microchip of impedance $Z_C = 23 - 145j\Omega$. Fig.2 shows the simulated realized gain of the tag (computed by the Method of Moment solver in FEKO) when it is attached over a layered model of human body [12] simulating a thin and thick thorax. There is a significant shift of the tag's response depending on the mass index of the phantom, and, accordingly, a significant loss of reading performance will be expected when moving from a configuration to another.

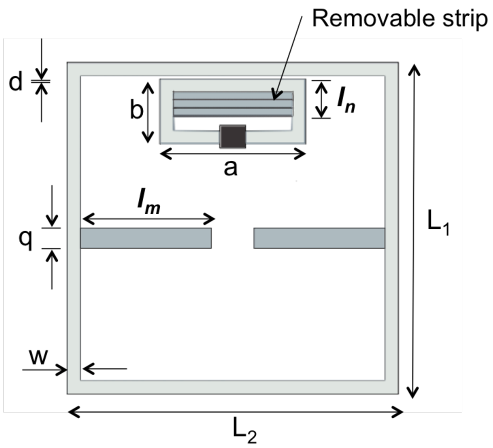



Figure 1. Layout of the epidermal tag. Size [mm]: $L_1 = L_2 = 50$, $w = 2$, $q = 2.5$, $d = 0.5$, $a = 26$, $b = 10$. The parameters in bold $l_m$ and $l_n$ can be modified after fabrication for impedance fine tuning.

A post-fabrication impedance tuning can be achieved by making the size of the feeding inner loop variable as shown in Fig.3. The width $l_n$ of the longest segment of the loop can be indeed modified by removing strips of equal size in order to softly reshape the path of the excitation currents and hence to shift the resonance frequency of the device. In particular, the peak of the power transfer coefficient $\tau$ moves from right to left on removing one, two or three strips of conductors.

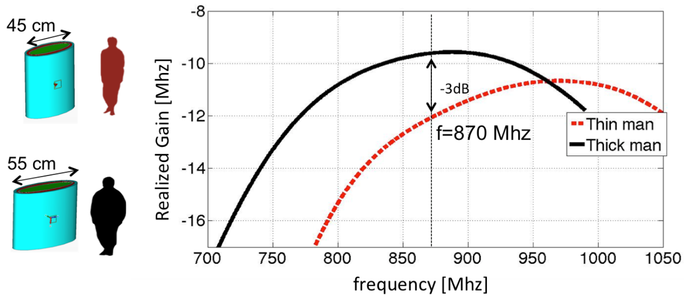



Figure 2. Example of tag response (realized gain) when the tag of parameters as in Fig.1 (without considering the dark gray elements) is placed onto two different layered human body elliptical phantoms .

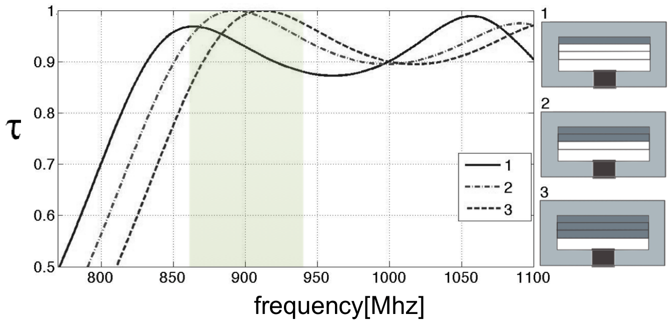



Figure 3. Tuning the power transfer coefficient of the epidermal tag by modifying the size of the feeding inner loop through removal of pre-carved strips of conductors .

It is moreover visible in the same Fig.3 that there are two peaks of the power transfer coefficient: a significant band enhancement could be hence achieved by packing those peaks into the useful RFID band. This is obtained by adding two whiskers in Fig.1: on increasing the length of the whiskers, the two peaks get closer and closer (Fig.4) and a nearly unitary $\tau$ is obtained in the [780-950]MHz band for $l_m$=14mm. The resulting layout is hence potentially suitable for world-wide application. Finally, the ratio between the total area of the conductor and the size of the tag is roughly 20%, with great benefit to skin transpiration. The residual surface of the skin could be moreover used to host other devices, chemical sensors and even drug delivery mechanisms, for enhanced capabilities.

## III. PROTOTYPE

A prototype of the epidermal RFID tag was fabricated onto a Poli($\varepsilon$-caprolactone) (PCL) synthetic membrane [14]. PCL is a semicrystalline bio-reabsorbable poly $\alpha$-hydroxyester with slow degrading rate due to its hydrophobic nature and high crystallinity degree. The membrane was produces by the electrospinning technique [15] by using PLC granules solved in $CHCl_3$ and THF:DMF (1:1) solutions. Electrostatic fiber-spinning, also denoted as "electrospinning", is a straightforward, cost-effective methodology to produce non-woven micro and/or nano-fibrous fabrics for several biomedical applications, e.g. scaffolds for tissue engineering or wound dressings. The electrospinning process is activated by applying high voltages

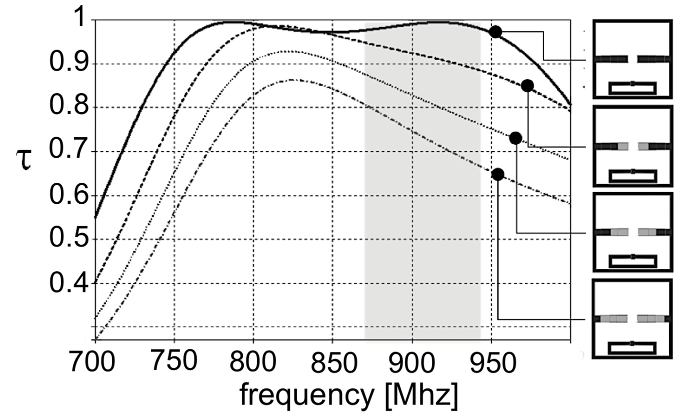


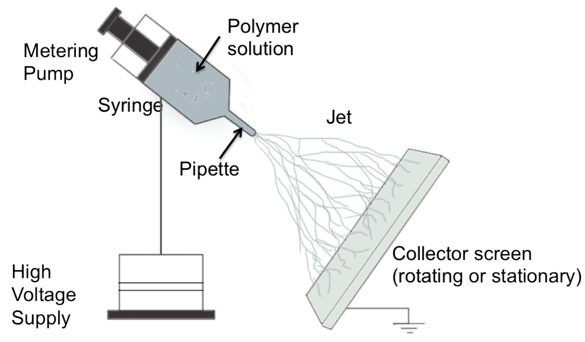

Figure 4. Broadbanding the epidermal tag by adding whiskers-like strips. .

between a polymeric solution flowing through a capillary and a collector plate, in order to generate an electrically charged jet, as shown in Fig.5a. As the voltage is increased, the electric field intensifies hence causing a force to build up on the pendant drop of polymer solution at the tip of the needle. This force acts in a direction opposing the surface tension that holds the solution. When the electric field reaches the critical value, the electrostatic force overcomes the surface tension at the tip of the capillary and a continuos charged jet is ejected. When the energized suspension moves away from the needle toward the collector screen, the jet rapidly thins and dries as the solvent evaporates. At the end, the polymer is randomly deposited onto the grounded target forming a dense non woven membrane (see fig.5b) that is completely bio-compatible (free of toxic residues). PCL is moreover flexible and stretchable for a comfortable application over any body curvature and, thanks to its fibrous structure, the natural transpiration of the skin is preserved.

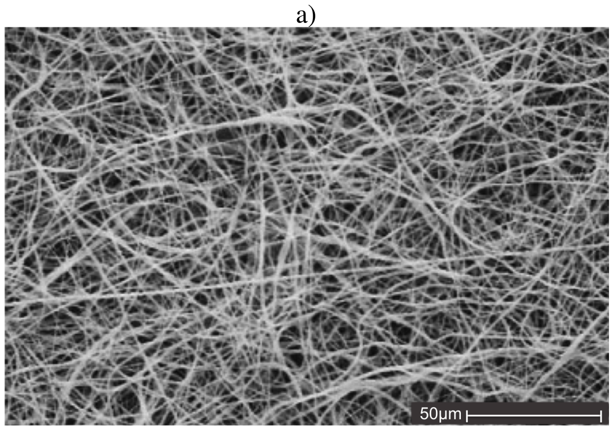
The considered RFID microchip was the EM4325 IC [16] (impedance $Z_C = 23 - 145j\Omega$, power sensitivity ($P_{chip} = -4.5\,dBm$) able to work as conventional RFID transponder as well as to provide on-chip temperature measurements in the [-40°C,+64°C] range (in passive mode) with a resolution of 0.25°C.

The antenna layout was fabricated on adhesive copper (conductive paints will be considered in the near future) carved by a two-axis digital-controlled cutter. The tuning elements on the inner loop were pre-cut in order to simplify the removal of the desired strips during the manual re-tuning of the antenna, if required. The RFID IC was hence soldered onto the antenna and the whole layout was finally transferred over the PCL membrane to give the prototype in Fig.6.

The communication performance were characterized by the measurement of the realized gain of the tag when it was attached onto the volunteer's arm by using a hypoallergenic cosmetic glue. The measurement set-up comprised a Thing-Magic M5 long-range reader, connected to a 5dBi broad-band PIFA antenna. The system is controlled by a custom software implementing the turn-on measurement procedure




a)




b)

Figure 5. The Poli $\varepsilon$-caprolactone membrane used as substrate of the epidermal tag: a) fabrication scheme by electrospinning; b) SEM image of the membrane .

[17]. Results for the case of reader-tag alignment are shown in Fig.7 in comparison with simulated data obtained once again by FEKO. The average antenna gain is of the order of -10dB in the [840-940] MHz band, in reasonable agreement with simulations. Accordingly, the maximum read range (by applying the simplified free-space Friis model) can be estimated as 80cm and 30cm in case of reader power emission of 3.2W EIRP (long-range reader) and 0.5 W EIRP (short-range reader), respectively. Such distances are computed by considering the low sensitivity of the microchip when the temperature data is read from the IC memory. Nevertheless these reading ranges could be fully compatible with a remote temperature monitoring.

## IV. Thermal Characterization

The interrogation of the EM4325 chip provides a thermal reading with an offset uncertainty to be determined by comparison with reference data. Thus, the thermal response and accuracy of the epidermal sensor were preliminarily evaluated by a reference test-case, wherein the RFID sensor outcome was compared with data provided by a thermocouple. The set-up comprised a water-filled plastic box (size $20cm \times 20cm \times 10cm$) wherein a heating electrode was inserted. The RFID sensor was attached onto the external surface of

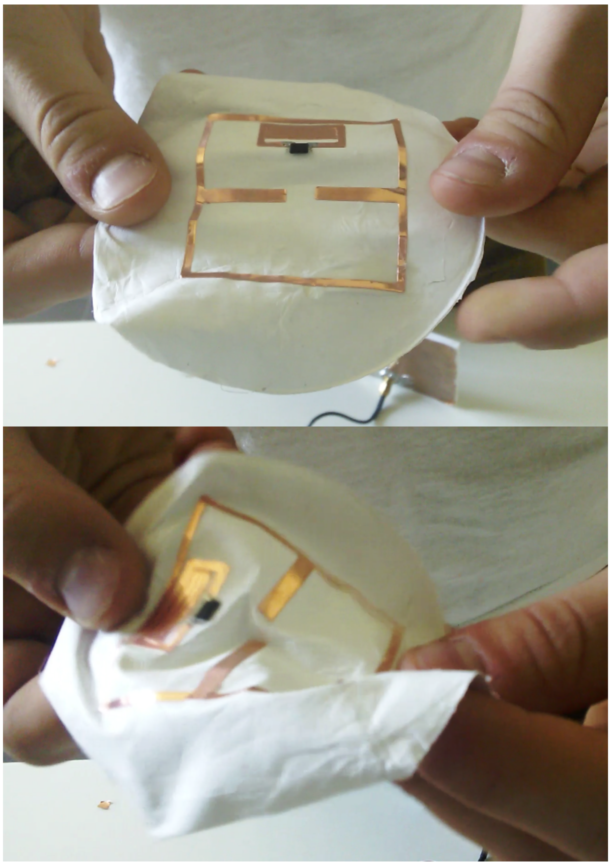


Figure 6. Prototype of the stretched epidermal RFID temperature sensor over poli $\varepsilon$-caprolacton membrane. .

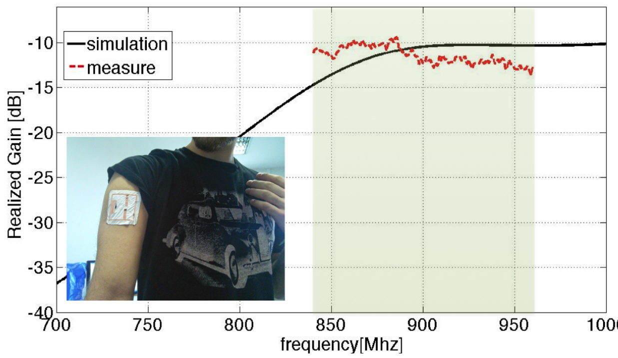
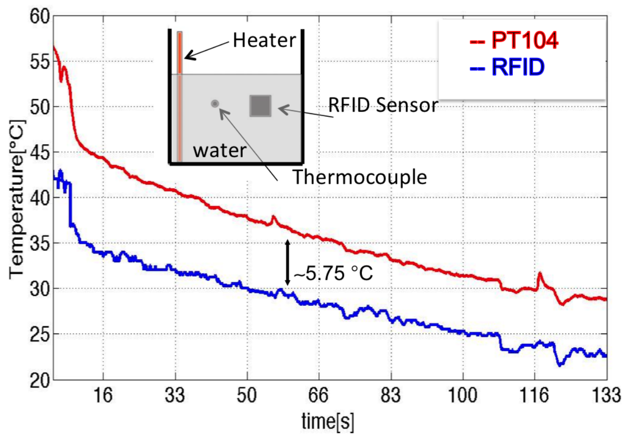
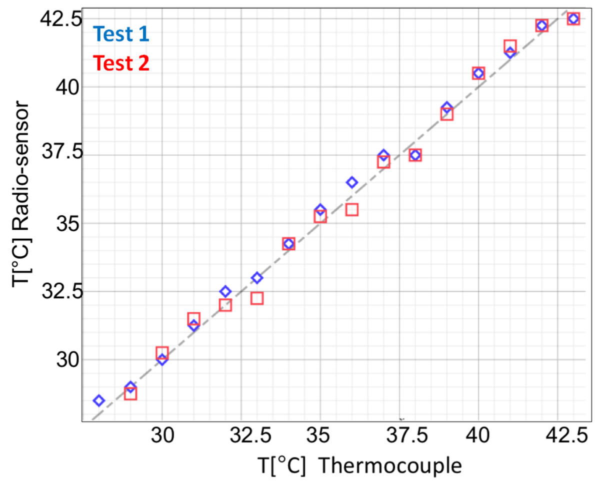



Figure 7. Measured realized gain of the fabricated epidermal RFID sensor when applied, by a natural glue, onto the arm of a volunteer. .

the container and interrogated by a CAEN-QUARK short-range reader (implementing the EM4325 non-standard reading protocol), emitting 0.25W and connected to a 3dB PIFA antenna, for an overall 0.5W EIRP. A PT104 thermocouple was placed close to the tag and connected to Picologic data-logger. The temperature of the water was increased up to 55°C; the heater was removed and the recovery curve was hence recorded by the two systems down to the ambient temperature. A 5.75°C temperature offset was estimated from the results of Fig.8a and was then used for the calibration of the RFID sensor. Fig.8b finally shows the true-measured temperature diagram derived from two further independent experiments. The resulting standard deviation of the calibrated system is lower than 0.25°C, which is comparable with the resolution of the temperature measurement of the EM4325 chip.




a)




b)

Figure 8. Sensor calibration by a reference thermocouple measurement. a) Identification of the temperature offset; b) RFID vs. true (thermocouple) measured temperature profile in further tests by using the calibrated RFID sensor.

## V. Application Example

The calibrated epidermal sensor was finally applied to the measurement of body temperature in both rest condition and under physical stress. The sensor was attached onto the right side of the abdomen (Fig.9) and again the thermocouple was placed at a close distance from it to provide reference data. The volunteer wore his usual fitness clothing. A first measurement was taken for 5min in rest condition and the detected temperature was 35°C. As expected, the measured values correspond to the superficial skin temperature which is lower than the core body temperature (about 37°C). Indeed, the surface temperature changes according to the different part of the body, even in the same subject, and is strongly affected by the blood flow at a particular time as well as by the surrounding ambient temperature [18].

During the second experimental session the volunteer rode an exercise bike for about 1h. During the physical activity the temperature followed a slightly ciclic profile due to thermoregulation mechanisms [19] (e.g. sweating changes in skin blood flow). A substantial agreement with the thermocouple data can be appreciated in both the experiments.

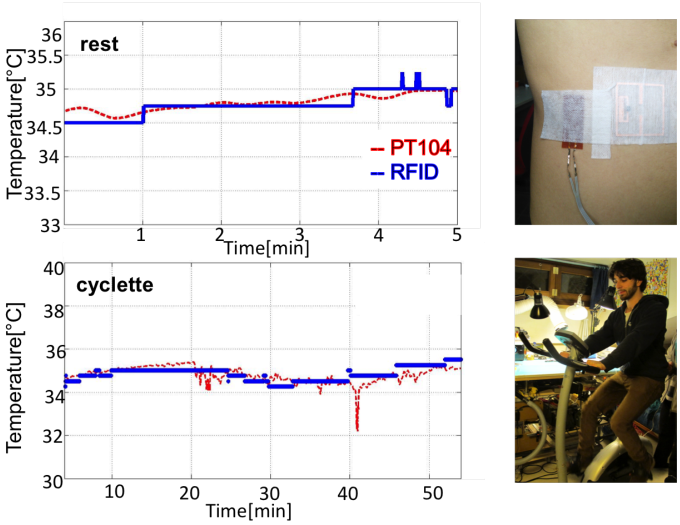



Figure 9. Temperature measurements over the body in rest condition (top) and during physical activity with exercise bike (bottom) .

## VI. Conclusions

This paper has considered for the first time (according the authors' best knowledge) the direct temperature measurement of the human body by an epidermal-like tag made on a Poli($\varepsilon$-caprolactone) membrane. The results are comparable with those produced by a conventional thermocouple with the non-negligible advantage to enable a full wireless interrogation up to 30cm when using a low-power interrogating device, virtually integrable into a mobile phone. The proposed tag has a broadband response and moreover includes a retuning capability for an easy adjustment over different human subjects and body districts.

For the sake of a better manufacturing and placement, the antenna layout should be however directly printed onto the membrane. Investigations on conductive paints suitable to the deposition over the PLC membrane is currently in progress and some results will be given at the Conference.

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