Oscillator based bio-sensor for skin sweating detection: A feasibility survey

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Oscillator Based Bio-Sensor for Skin Sweating Detection: A Feasibility Survey

Naveen Srinivassane1, Aïcha Saïd[2], Frédéric Hameau[2], Rémy Vauché[3], Florence Podevin[1], Sylvain Bourdel[1]
[1]Univ. Grenoble Alpes, Grenoble INP, TIMA, 38000 Grenoble, France
[2]Univ. Grenoble Alpes, CEA leti, 38000 Grenoble, France
[3]Univ. Aix Marseille, IM2NP, 13013 Marseille, France

***Abstract*—This paper is a feasibility survey of the design of an oscillator-based bio-sensor for skin physiological parameter detection. The sensor aims to detect stress through skin sweating measurement. The proposed architecture uses an oscillator coupled to an on-skin resonator. The oscillator is based on a cross coupled pair (CCP) which is directly connected to the resonator. The on-skin resonator has been designed to have an oscillation frequency sensitive to the sweat. Oscillator outputs are driven by two buffers that act as 50 Ω impedance matching. Two test benches, one with a dry skin and one with a wet skin, have been used for the study. The oscillation frequency of this oscillator is equal to 27.876 GHz for the dry skin, and 27.911 GHz for the wet one. The difference between the two oscillation frequencies, which is equal to 35 MHz, can be easily distinguished which validates the feasibility survey. The survey provides the sizing of each block and the simulated results in 55 nm BiCMOS technology.**

Keywords*—*voltage control oscillator, sweat sensor, wet skin, dry skin, oscillation frequency

I. INTRODUCTION

Nowadays, the progress in healthcare brought a fast evolution for the devices dedicated to physiological measurements. The major exploited solutions are based on magnetic resonance imaging (MRI), X-ray, and ultrasound. However, all those techniques are either invasive for the human body or consume a lot of energy. Thus, even if these techniques are effective, they are not applicable to new healthcare applications such as mobile monitoring or home diagnosis. For such a new application field, the challenge is to design compact, low-power, easy to use and robust devices.

The radiofrequency (RF) domain is an interesting solution to answer all these challenges. Indeed, RF solutions can be not invasive for the human body and low power. In particular, frequencies from 20 to 120 GHz show shallow penetration in the skin [1]. There is also no physical impact. In parallel, sweat only involves a superficial change in the dielectric behaviour of the skin. Thus, skin sweating detection using RF can be considered using a low power sensor since there is no need for deep penetration.

It has been shown that the water content in tissues differs from healthy and cancerous skin. Indeed, malignant lesions have around 26 % higher water content than healthy tissues [2]. This change affects the dielectric parameters of different tissues so they can be electromagnetically characterized. Several approaches have been investigated to detect the variation of the reflected signals from healthy and cancerous skin. In [3], they showed a difference of 0.3 dB in $|S_{11}|$ and a 6° phase variation between normal and cancerous skin at 30 GHz using a simulated waveguide probe. Moreover, the use of a vector network analyser (VNA) to measure the reflection coefficient of a waveguide closed by the skin [1] or a coaxial sensor similarly closed [4] have firstly been explored. An output voltage variation of 16 mV at 40 GHz is detected in [1] and a phase variation of 9 ° is detected at 1.5 GHz in [4] between dry and wet skin. To improve the compactness of such systems, [5] proposes an integrated 125 GHz reflectometer based on a 6-port structure. Instead of monitoring reflection coefficient, some authors propose the use of integrated radar systems [6]. However, both are complex architectures whereas oscillator-based sensors are much simpler to design and operate. For example, their use has been investigated to detect changes in the heartbeat [7] [8]. The system operates in near-field around 2.4 GHz and measures the antenna impedance modification through an oscillator connected to the antenna. The system is quite simple and can use an autoregressive spectral estimation method [7] or a PLL [8] to retrieve the frequency shift which occurs when antenna impedance is modified.

In this study, we propose to investigate the use of an oscillator based sensor working around 30 GHz for which the dielectric behaviour of the skin is significantly modified by the sweat (+1.55% for conductivity and +14.4% for permittivity). The frequency deviation estimator is not studied here. The paper focuses on the design of the oscillator and especially the design of a cross-coupled pair (CCP) and its connection to a resonator designed and simulated using HFSS software. As the resonating frequency is different between the two environments (wet and dry skin), the oscillation frequency is modified according to the skin state. The sensitivity of the bio-sensor is directly related to the offset frequency (Δf) between the two states and has to be maximized. The main objective is to design a miniaturized contactless system that can discriminate the two states (wet and dry) of the skin.

This paper is organized as follows. Section II presents the different blocks of the oscillator: the resonator, the CCP, and the output buffers. The design methodology is detailed and simulation results are provided for each block. In Section III, the study is validated by simulations of the oscillation frequency and the phase noise for wet and dry skin. Section IV gives the conclusion and the perspectives.

II. SYSTEM DESIGN

A. Architecture

The studied system presented in Fig. 1 consists of a sensor designed on a Teflon substrate connected to a CCP integrated in 55 nm BiCMOS technology and assembled with flip chip process to minimize interconnection effects. The role of the CCP is to compensate for the losses of the resonator. These are a key parameter design because small CCP can compensate for low losses, which enables low power consumption and low parasitic capacitance. Indeed, any parasitic capacitance added to the resonator by the CCP reduces the oscillator Δf which reduces the sensor sensitivity. Thus, the resonator shall be optimized to achieve low losses at the resonant frequency and a large Δf between dry, and wet skin. The CCP is composed of transistors N_1 and N_2 which are biased through $\frac{\lambda}{4}$ meandered lines to isolate DC from RF as shown in Fig. 1. Two buffers at the outputs enable matching to perform measurements on a 50 Ω impedance environment. Both the CCP and the buffers, are biased with a current mirror embedding a noise filtering capacitor C_{dec}.

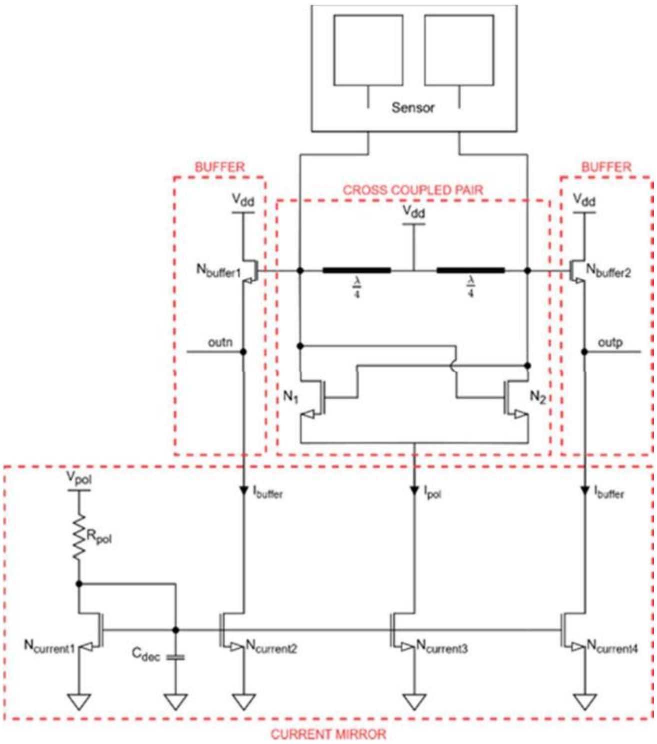

Fig. 1. Proposed system architecture

B. *Sensing resonator*

The sensing resonator is based on two-planar resonating patches, which act as a unique differential near field sensor for sweat detection. One patch is presented in Fig. 2. It consists of a metallic square shape deposited on a grounded Teflon layer with a thickness equal to 200 μm and dielectric permittivity equal to 2.1. Indeed, a substrate with a low relative permittivity allows to maximize the variation of the antenna impedance. The sensor is placed at 1 mm on top of the skin to ensure near-field radiation. Two access lines (not represented in Fig. 2) are used to feed the resonator (one access line per patch). The resonator is optimized to maximize Δf at 30 GHz, which leads to a patch size of 2.2 mm by 1.3 mm. The resonator has been simulated and optimized with HFSS software. As shown in Fig. 2, an infinite box, where the permittivity and conductivity are chosen regarding the skin state, is placed at 1 mm away from the patch to emulate the skin. The dielectric parameters of the wet and dry skin are given in TABLE I. [9].

TABLE I. DIELECTRIC SKIN PROPERTIES AT 30 GHZ [9]

State	Dry	Wet
σ	27.099 S/m	27.52 S/m
ε_r	15.51	17.743

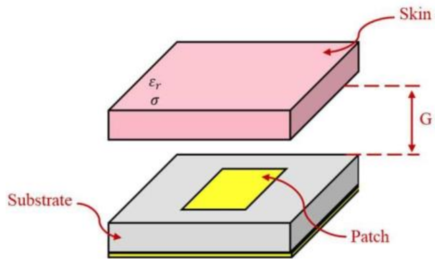

Fig. 2. Schematic of the sensor

The simulation results of the sensor based on two-planar patches are shown in Fig. 3 where the resonant frequency is read at $\varphi = 0°$. The resonant frequency for the dry skin is 31.701 GHz whereas it is 31.830 GHz for the wet skin, which leads to a Δf = 130 MHz. The sensitivity is relatively good between the two states of the sensor. This first observation suggests a large margin for the design of the CCP.

The real part of the sensor impedance can be extracted from Eq. (1) and is presented in Fig. 4. At the resonant frequency associated to each state, losses are extremely low since a high parallel resistance R_p, between 80 to 90 kΩ, has been found. This second observation definitely states a relaxed constraint on the CCP design which is favorable to Δf maximization. In addition, the value of R_p keeps almost constant around the resonance frequency, which ensures robustness in the design. The variation of R_p is about 10 percent around the resonance frequency for wet and dry skin respectively (TABLE II.). To keep the distance between the sensor and the skin, it will be considered to make a sensor with a cavity on its edges or add an interface layer on the patch for practical implementations.

$$R_p = \Re\left(\frac{1}{Y_{11}}\right) \quad (1)$$

TABLE II. SENSOR PARAMETERS

State	Dry	Wet
Resonance frequency	31.701 GHz	31.830 GHz
R_p **at resonance frequency**	81.89 kΩ	90.77 kΩ

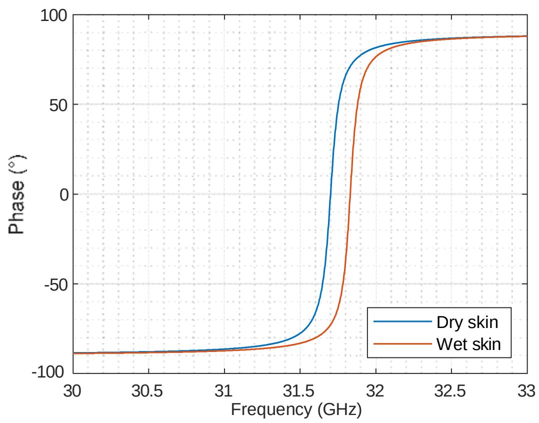

Fig. 3. Phase of the sensor with dry and wet skin

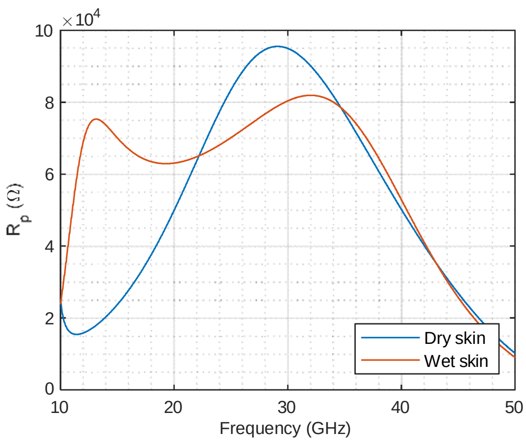

Fig. 4. Real part of the sensor impedance with dry and wet skin

C. *Cross coupled pair transistors*

The CCP has to compensate for the resonator losses. To ensure the oscillations, the system needs to respect the Barkhausen criteria [10], which is equivalent in a LC oscillator to the following condition:

$$|R_{neg}| = \frac{2}{g_m} \leq \frac{R_p}{A} \tag{2}$$

where g_m is the transconductance of the CCP, R_{neg} the negative resistance achieved by the CCP, R_p the real part of the resonator at the resonant frequency, and A a real positive value greater than one (often named compensation factor). The A parameter is typically equal to 3 or 4 for perfectly mastered integrated resonating tanks which ensures good loss compensation through the CCP. For bio-sensors, with flip-chipped die on the Teflon board, as the proposed one, it seems much more reasonable to aim at higher values of A, in the order of tens. Knowing R_p, providing A, it is then possible to size the CCP with the following relationship giving g_m as a function of the biasing current (I_{pol}), the transistor width (W) and its length (L)

$$g_m = \sqrt{2K_n \frac{W}{L} I_{pol}} \tag{3}$$

where $K_n$depends on the mobility of the electrons and the oxide capacitance. A large R_p is interesting as it enables large R_{neg} , or small g_m achievable with low biasing current. To isolate RF signal from DC path, a $\frac{\lambda}{4}$transmission line is preferable to a choke inductor in order to minimize the parasitic capacitance and then maximize Δf. The transmission line is first sized, before considering the CCP effect, as it is an important contributor of the resonator frequency shift. A 1 µm width has been chosen. The length of 1518 µm corresponds to $\frac{\lambda}{4}$ at 30 GHz. Such line can be meandered which leads to a reasonable surface of 150 µm by 180 µm if 20 µm spaced sections of 150 µm length are considered.

To size the CCP, parametric simulations are run out to extract the negative resistance (Fig. 5) and the parasitic capacitance (Fig. 6) for different transistor widths in function of I_{pol}. According to the curves in Fig. 5, a CCP designed with the minimum width of $W = 3$ µm achieves a negative resistance of −4.030 kΩ for $I_{pol} = 1\ mA$ which gives a compensation factor equal to 77 which is sufficiently large to compensate for extra-losses due to layout and interconnection between resonator and CCP (not taken into account in this feasibility study) and also starts oscillations. The parasitic capacitance of the CCP, associated to $\frac{\lambda}{4}$ transmission lines, seen from the resonator side is about 4.34 fF. The characteristic of the CCP are given in TABLE III.

TABLE III. CHARACTERISTICS OF THE CROSS COUPLED PAIR

W	3 µm
L	55 nm
I_{pol}	1 mA
V_{dd}	1 V
W_{line}	1 µm
L_{line}	1518 µm
$\|R_{neg}\|$	4.03 kΩ
C_{par}	4.34 fF

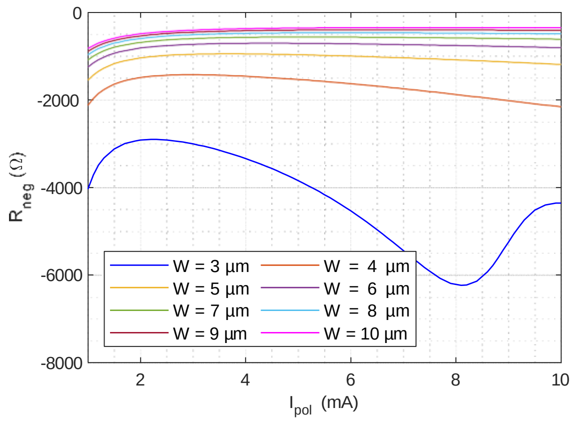

Fig. 5. Negative impedance variation

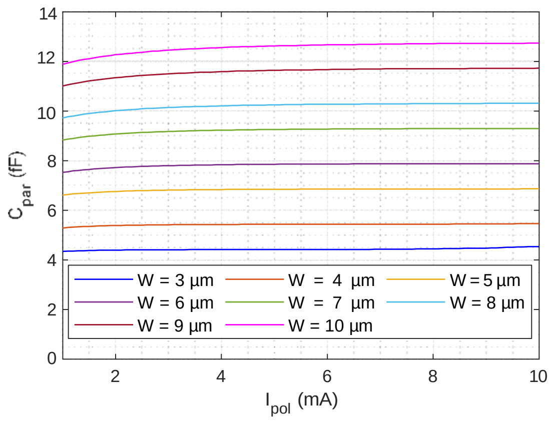

Fig. 6. Parasitic capacitance variation

D. *Buffers*

In Fig. 1, the transistors $N_{buffer1}$ and $N_{buffer2}$ are source follower amplifiers acting as output buffers. They are biased by the V_{dd} voltage and a specific current source (I_{buffer}). The goal is to synthesize an output impedance of 50 Ω. Knowing that [10]:

$$\Re(Z_{out}) = \frac{1}{g_m} \tag{4}$$

the transistors have to be sized to provide g_m = 20 mS. The main parameters of the buffer are listed in TABLE IV. The width W = 8 µm leads to a parasitic capacitance of C_{par_buff} = 0.85 fF which leads to a total parasitic capacitance of 6.04 fF. It is worth mentioning that such kind of buffer is mandatory only for measurement purpose. In the case of a full implementation with a frequency deviation estimator such as a PLL, the constraint on the required buffers might be relaxed.

TABLE IV. CHARACTERISTICS OF THE BUFFERS

W_{buffer}	8 µm
L_{buffer}	55 nm
I_{buffer}	1 mA
V_{dd}	1 V
C_{par_buff}	0.85 fF

III. SIMULATION OF THE OSCILLATOR

The complete oscillator has been simulated in the two states (wet and dry skin) using two different S-Parameter boxes obtained from electromagnetic (EM) simulations (HFSS) of the resonator. As shown in Fig. 7, two separated frequencies are observable at 27.876 GHz and 27.911 GHz for a dry and wet skin respectively, leading to a Δf= 35 MHz. Compared to the resonator performances, a frequency shift of 11.7 percent is observed due to the loading effect of the CCP and the buffers. A Δf reduction down to 26 percent of the initial value is also observed. By the way, 35 MHz stays a frequency deviation that can be easily detected. As presented in TABLE V. the output power is 4.7 dBm which corresponds to a 1.1 V voltage dynamic at the buffer outputs. Output buffer voltages are shown in Fig. 8. The phase noise at 35 MHz of the carrier is equal to -129 dBc from simulations. The frequency deviation, the output dynamic and the phase noise show the potentiality of the system. These performances allow us to consider this bio-sensor architecture for future implementation.

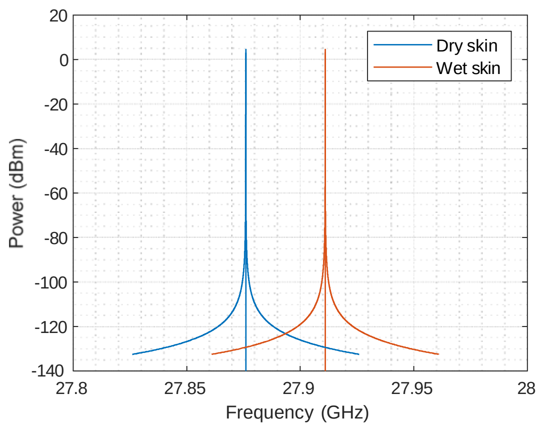

Fig. 7. Harmonic balance simulation with the phase noise

TABLE V. SIMULATION RESULTS

State	Dry	Wet
Frequency	27.876 GHz	27.911 GHz
Power	4.75 dBm	4.794 dBm
Phase noise at Δf	-129.1 dBc	-129.1 dBc

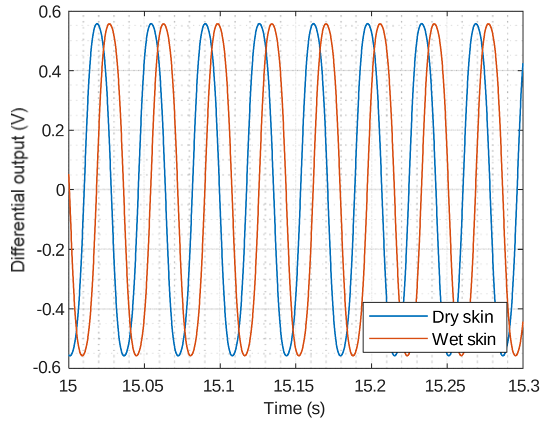

Fig. 8. Transient simulation

IV. CONCLUSION

This paper has presented a feasibility survey of an oscillator based bio-sensor for sweat detection. The oscillator is composed of an on-skin resonator coupled to a CCP with buffers integrated in 55 nm BiCMOS technology. The study is performed on the basis of EM simulations for the resonator and design kit based electrical simulations for the active integrated parts of the sensor. Depending of the skin sweat (dry or wet) the bio-sensor provides an output frequency deviation of 35 MHz around a frequency value of 27.9 GHz, with an output power of 4.7 dBm and a phase noise of −129 dBc at 35 MHz. The CCP has been oversized to ensure oscillations even with additional losses due to interconnects and layout. This first study shows the potentiality of this architecture for bio-sensor and paves the way toward further implementation.

REFERENCES

[1] S. Kharkovsky *et al.*, « Near-field microwave and mm-wave noninvasive diagnosis of human skin », *IEEE International Workshop on Medical Measurements and Application*, p. 5-7, 2009

[2] A. Taeb *et al.*, « Millimetre-wave waveguide reflectometers for early detection of skin cancer », *IET Microwaves, Antennas & Propagation*, vol. 7, nº 14, p. 1182-1186, 2013

[3] U. Khokhar *et al.*, « Near-field tapered waveguide probe operating at millimeter waves for skin cancer detection », *IEEE Trans. Antennas Propag. USNC/URSI National Radio Science Meeting*, p. 795-796, 2017

[4] P. Mehta *et al.*, « Microwave reflectometry as a novel diagnostic tool for detection of skin cancers », *IEEE TIM*, vol. 55, nº 4, p. 1309-1316, 2006

[5] B. Laemmle *et al.*, « A 125-GHz Permittivity Sensor With Read-Out Circuit in a 250-nm SiGe BiCMOS Technology », *IEEE TMTT.*, vol. 61, nº 5, p. 2185-2194, 2013

[6] J.-K. Huang *et al.*, « A 5.8-GHz radar sensor chip in 0.18-µm CMOS for non-contact vital sign detection », *IEEE RFIT*, p. 1-3, 2016

[7] S.-G. Kim *et al.*, « Wireless RF Vital Signal Sensor Using Autoregressive Spectral Estimation Method », *IEEE Antennas Wirel. Propag. Lett.*, vol. 11, p. 4, 2012.

[8] Y. Hong *et al.*, « Advanced non-contact near-field proximity vital sign sensor using phase locked loop », *EuMC*, p. 605-608, 2013

[9] D.Andreuccetti *et al.*,"An Internet resource for the calculation of the dielectric properties of body tissues in the frequency range 10 Hz - 100 GHz", IFAC-CNR, 1997. Based on data published by C.Gabriel et al. in 1996. [Online]. Available: http://niremf.ifac.cnr.it/tissprop/

[10] E. Sharma *et al.*, « Design of a 77-GHz LC-VCO With a Slow-Wave Coplanar Stripline-Based Inductor », *IEEE TCAS-I: Regular Papers*, vol. 67, nº 2, p. 378-388, 2020