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► **To cite this version:**

Dau Thi Ngoc Nga, Giorgio Mattana, Vu Thi Thu, Ronan Roussel, Benoît Piro. A simple flexible printed capacitive pressure sensor for chronic wound monitoring. *Sensors and Actuators A: Physical*, 2022, 338, pp.113490. 10.1016/j.sna.2022.113490 . hal-03667167

HAL Id: hal-03667167

<https://u-paris.hal.science/hal-03667167>

Submitted on 22 Jul 2024

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A Simple Flexible Printed Capacitive Pressure Sensor for Chronic Wound Monitoring

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In this work, we describe the fabrication of a conformable capacitive pressure sensor obtained by inkjet-printing silver electrodes onto polyimide foils subsequently glued face-to-face onto a double-sided tape. The sensor operates for pressures between ca. 1 Pa and 2 kPa. Its response time is 0.25 s. It presents a non-linear response as a function of the capacitance variation with the pressure, namely $\Delta C/C_0 = P^n$ with $n = 0.58$. The sensitivity is higher than 0.5 kPa^{-1} for pressures below 200 Pa, and 0.08 kPa^{-1} for pressures up to 2 kPa. The sensor is included into a commercial dressing for direct measurements on the body of a volunteer. It is used to monitor how firmly a dressing is glued on the (skin) surface, the main application being the reduction of ischemia risk. It is also used to monitor the movement of the ankle of the patient while walking, and to monitor the pressure exerted on the foot sole when left at rest down on the floor or lifted.

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† To the memory of Prof. Ronan Roussel, a leader, a builder, concerned with advancing the care of diabetic patients

Keywords: Pressure sensor, Inkjet-printed device, Dressing, Wound monitoring

1. Introduction

Worldwide, one adult out of eleven suffers from diabetes mellitus. If not correctly treated, it may lead to diabetic foot ulcers, one of the most important causes of leg amputation. Treatment of such chronic wounds is therefore among the most important health challenges in terms of concerned population, quality of life of patients, and costs [1]. Offloading the wound is one medical answer to improve the healing rate as it reduces the risk of ischemia by allowing good blood circulation at the wound site. For that, the patient must not take long stationary posture where the body's weight is applied on the wound [2]. The healing rate is also very dependent on the pressure the dressing itself applies on the wound (typically when the sticking bandage is too tightly set on the skin). Currently, because no convenient tools are available, this pressure is estimated only purely qualitatively by the medical staff when the dressing is applied, and the patients have no possibility to monitor it by themselves. The simplest way to monitor the pressure applied on a wound is to integrate pressure sensors into dressings [3], so that the pressure can be monitored when the dressing is set on the wound, informing the medical operator that it is too tight or informing the patient that, after a certain amount of time spent in a fixed posture, he/she must change his/her position to let the blood flow more freely. Because dressings are flexible and disposable, ideal pressure sensors should be flexible and disposable as well, and of reasonable cost compared to that of the dressings taken alone [4]. Some original pressure sensors proposing an optical detection are based on the change in resonance frequency of resonant circuits [5], or on field-effect transistors (FETs) [6,7]. Among the variety of pressure sensors integrated into dressings for diabetic ulcers, most belong to one of the three main types, however: capacitive, piezoresistive or piezoelectric [8,9].

In general, any sensor whose working principle relies on capacitance variations are considered as capacitive sensors. The most common capacitive pressure sensors are made of simple

parallel conductive plates separated by an ideally elastic dielectric. Upon application of a pressure perpendicular to the two plates, the thickness of the dielectric decreases, resulting in a capacitance increase. There is literature available concerning capacitive dressing sensors [10] or printed capacitive sensors [11,12] but very few papers address the two challenges together. Shi *et al.* used Ecoflex® as dielectric in a screen-printed flexible capacitive sensor made with PEDOT:PSS conductive contacts. The operating range was 0-20 kPa with sensitivities ranging between 10^{-3} and 10^{-2} kPa⁻¹ [13]. Masihi *et al.* described a screen-printed capacitive sensor made of thermoplastic polyurethane substrate and using microporous PDMS as dielectric, demonstrating sensitivities between $2 \cdot 10^{-3}$ and 10^{-2} kPa⁻¹ for pressures ranging from 1 kPa to 1 MPa [14]. In the field of e-skin applications, Palaniappan *et al.* also used a micro-structured dielectric to address a lower pressure range. The limit of detection was about 10 Pa, with sensitivity of 0.16 kPa⁻¹ at 10 kPa [15]. Ntagios *et al.* described a 3D-printed sensor made of printed Ag electrodes and Ecoflex® as dielectric for applications on robotic hands, with sensitivity of about 10^{-3} kPa⁻¹ for pressures between 1 and 50 kPa [16]. None of these works investigated dressing applications.

Resistive (or piezoresistive) sensors are also made of two parallel plates, but separated by a resistive material (e.g., metallic nanoparticles, nanowires, carbon nanotubes, reduced graphene oxide sheets, etc.), whose resistance depends on the distance between the conductive particles: the higher the pressure, the shorter the distance between particles and the smaller the resistance. There are few works dealing with resistive sensors for applications in wound monitoring. Chung *et al.* reported a conductive fabric for applications in ulcer monitoring [17]. Mehmood *et al.* reported a very complete work dealing with a Kapton foil-based flexible sensing device for chronic wound diagnostics, but they used commercial piezoresistive sensors [18,19]. More recently, Da Silva *et al.* described a piezoresistive device for dressings by assembling polypyrrole and CNT on a polyurethane membrane. The maximum sensitivity

was ca. 0.1 kPa^{-1} within an operating range between 1 and 30 kPa [20]. Garlach *et al.* proposed an array of screen-printed piezoresistive pressure sensors (carbon nanotube/PDMS inkjet-printed on PET foil) operating between 1.5 kPa and 15 kPa [21]. Xiao *et al.* described a screen-printed resistive array (bottom Ag interdigitated electrodes and carbon black top electrodes, both on PET) operating in a pressure domain from 10 to 200 kPa, which they applied for measuring wrist pulse but also foot plantar pressure [22].

Piezoelectric pressure sensors have a significantly different functioning. They are made of a material which can generate an electric potential difference between two conductive plates upon application of a pressure between these plates [23]. If we only consider flexible materials, the most popular flexible piezoelectric material today is poly(vinylidene fluoride-trifluoroethylene) (P(VDF-TrFE)), with many applications in the literature but not for dressings, these sensors being more adapted for monitoring movements rather than static pressure.

In this work, we chose the capacitive transduction principle because of its advantages compared to other working principles, for instance its low dependence on temperature (which is not the case of resistive sensors), and because these sensors are easier to fabricate than piezoelectric ones, which typically require complicated procedures for properly polarizing the piezoelectric layer. We propose here an extremely simple method to fabricate the flexible pressure sensor: the only technique used is inkjet printing, with only one type of silver ink; the only material used is a commercially available polyimide glued tape, used as substrate for printing, dielectric intercalator and sticking plaster to secure the sensor onto the dressing or the bandage. The other important feature of this work is that it reports *in-situ* measurements: the device was tested on a volunteer to reproduce real conditions. Our sensor is able to monitor the tightness with which a dressing is applied onto the skin and is also able to measure the pressure that is externally applied on the dressing.

2. Experimental Section

2.1. Materials

Commercial dressings (Hansaplast, 72 mm x 19 mm or 60 mm x 70 mm) were bought in a local pharmacy. 3MTM double coated tape 9495MP was used as dielectric layer and purchased directly from the 3M Company (Minn, USA). DuPontTM (Del., USA) Kapton® HN or Kapton® Black were used as printing substrates. Inkjet printing was performed from a commercially available Ag nanoparticle-based ink (SicrysTM I40DM-106, metal loading: 40% wt, PV Nano Cell, Israel), or from a lab-made Au nanoparticle-based ink [24]. Weights were small plastic-covered metal pieces from 5g and more, 3D-printed polymer pieces from 0.5 to 5 g, and small pieces of polyimide (PI) or paper sheets for the lightest ones (< 500 mg). Conductive wires were contacted to the sensor's electrodes with a conductive glue from MG chemicals company (UK) or with an anisotropic conductive tape 3MTM 7393.

2.2. Characterizations

The surface morphology of the dielectric as well as conductive layers was examined by scanning electron microscopy (SEM, Zeiss SUPRA 40). An Agilent E4980A precision LCRmeter was used to measure the change of the sensors' capacitance when applying an external pressure; measurements were performed at 4 kHz under a bias of 1 V and small amplitude oscillations of 10 mV (peak-to-peak). An Arduino Uno was also used to measure capacitance changes. In this case, no pulsed signal was used; the capacitance C was derived from the time constant $\tau = RC$ of the capacitor and the resistance R of the circuit.

2.3. Fabrication of the capacitive pressure sensor

The objective was to fabricate a capacitive pressure sensor easy to produce and to include into a commercial dressing. This is the reason why we did not print directly the sensor onto the dressing's material (cotton fabric or sticky tape) but we rather developed an independent device that could be simply glued on the dressing as a patch. The device has a sandwich-like

structure with two parallel silver plates (7 x 7 mm in lateral dimensions for the largest one, **Figure 1A**) printed onto a Kapton® HN tape (13 μm thick; this material was selected for its high temperature resistance and its flexibility) put face-to-face on each side of a 140 μm thick double-sided 9495MP 3M tape serving as dielectric layer (side-view of the structure shown on **Figure 1B**). The sensor's total thickness was precisely measured by SEM and varies between 140 and 170 μm, consistent with what was expected from the thicknesses of the individual components (**Figure 1C**); At higher magnification, the structure of the layers appears clearly, with the conductive top and bottom Ag layers, the PET tape in the middle and the two glue parts (of kaleidoscopic appearance) on both sides (**Figure 1D**). Before printing of the Ag electrodes, the PI tape was cleaned with acetone and deionized water in sequence, followed by heating at 150°C for 60 minutes. The top and bottom Ag electrodes were obtained from one layer of printing, using a Dimatix DMP-2800 inkjet printer. The plateau temperature was fixed to 40°C and the cartridge temperature to 28°C. The waveform was the standard Dimatix one, and the drop-spacing was set to 15 μm. After printing, electrodes were annealed in an oven at 150 °C for 90 minutes to evaporate the solvent and sinter the Ag nanoparticles (NPs). Under these conditions, the electrode surface appears as partly coalescent Ag nanoparticles (**Figure 1E**). These printing and curing parameters typically allow the deposition of a 300 nm thick Ag layer. The final sensor resulting from the assembly of two printed electrode onto the double-side tape is shown on **Figure 1F**. Some experiments were made with sensors placed on the skin of one of the author (N.T.N. Dau), with her written consent. Experiments were short (a few minutes of test) and done by herself.

[Insert here Figure 1; two columns]

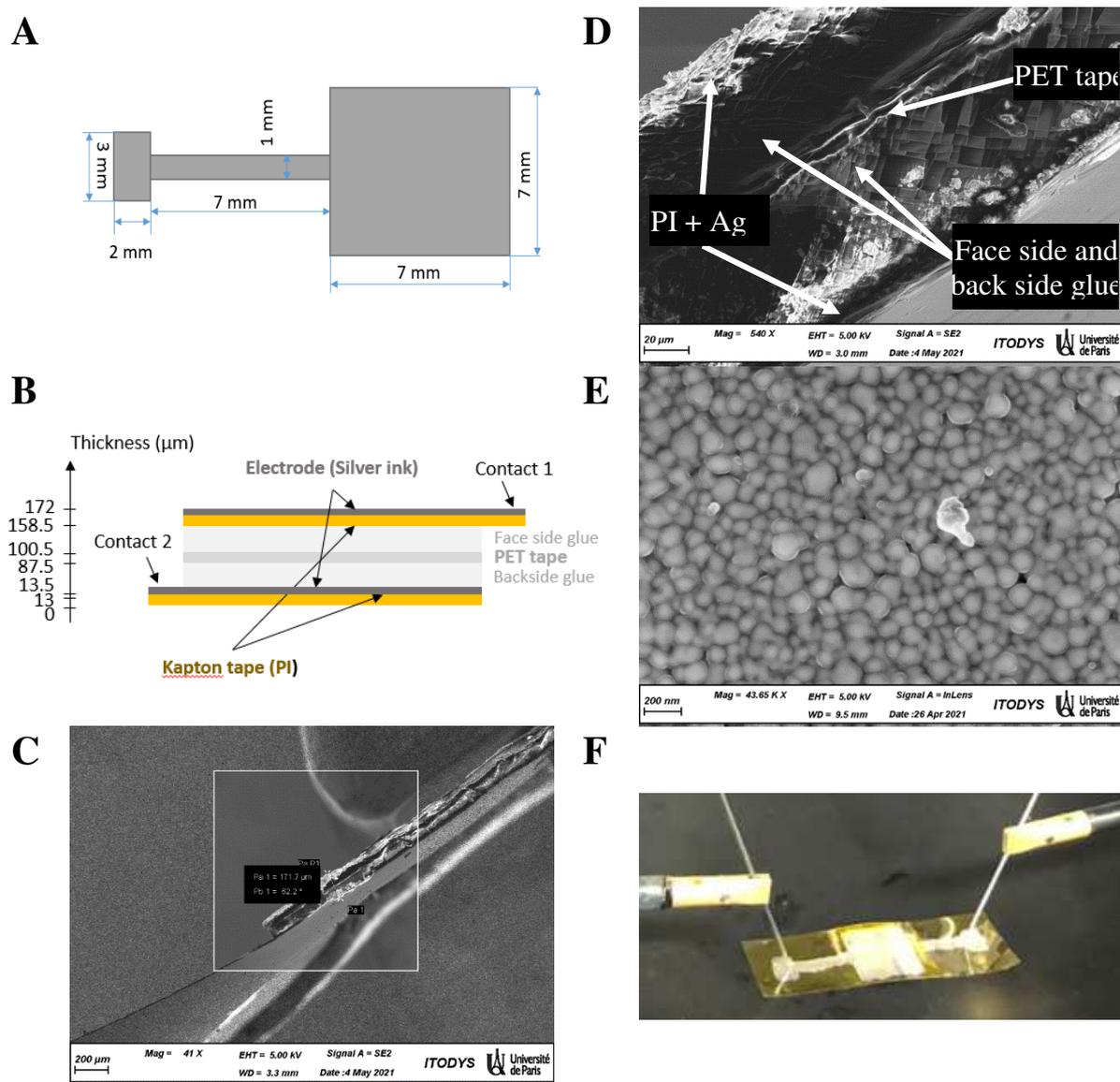


Figure 1. (A) Scheme showing the lateral dimensions of a printed Ag electrode and its connectors. (B) Structure of the capacitive pressure sensor, obtained by assembly of two Ag-printed electrodes on both sides of a double-side PET tape. The expected thickness of the device is 172 μm , made of 2 \times Ag plates (0.5 μm), 2 \times PI tapes (13 μm), 58 μm of face side glue, 13 μm PET tape and 74 μm of backside glue. (C) SEM images (side-view) of the pressure sensor, showing the various layers. (D) Zoom-in showing details of the various components (conductive layers, PET tape and glue layers). (E) Surface aspect of the printed Ag electrodes, formed of Ag grains of ca. 50-150 nm in diameter, partly coalescent. (F) Resulting device on the probe station.

3. Results and discussion

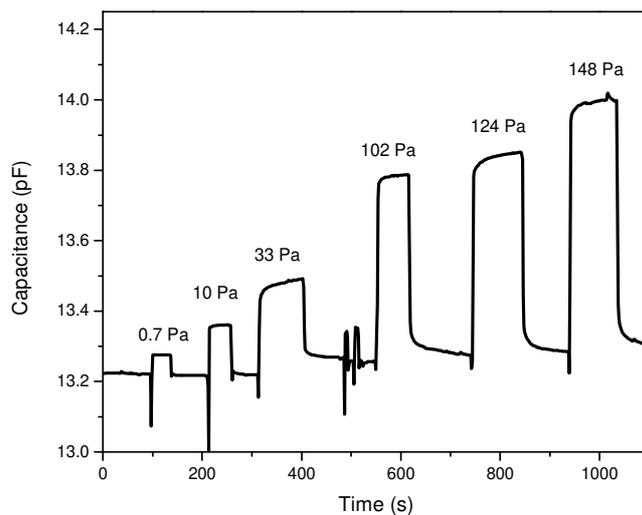
3.1. Sensor's figures of merit

The sensor is expected to sense dressing movements (without necessity to quantify their amplitude), the pressure applied on the dressing when the patient is at rest, or sensing the pressure applied on the dressing when it is set on the wound by the nurse or the patient. The timescale of these measurements is relatively long, so that the response time of the sensor is not critical. In our case, it was estimated to be 0.25 s (see **Figure S1** for details) from the capacitance change resulting from the application of a 200 Pa pressure.

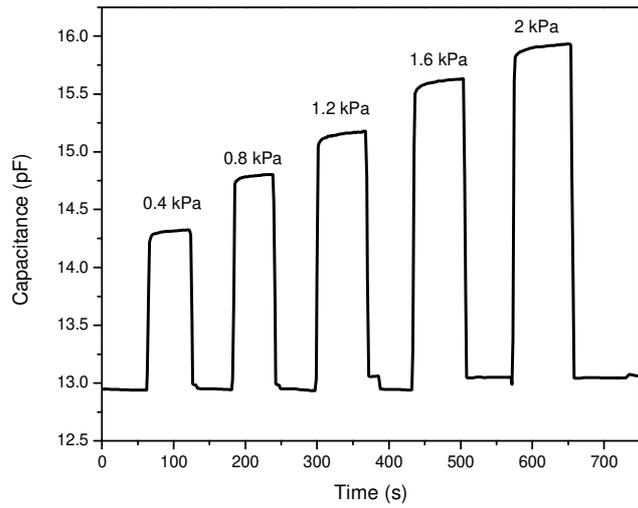
The limit of detection of our sensor was estimated of around 0.5 Pa, for a S/N ratio of 3 (see **Figure S2**). A calibration curve was established for pressures ranging from 0.7 Pa up to 2 kPa (**Figures 2A,B**). As expected, the dependance of the capacitance on the pressure, $C=f(P)$, is not linear. Two domains can be identified: one corresponding to low pressures below 150 Pa, the other corresponding to higher pressures (**Figure 2C**). With ΔC the capacitance changes and C_0 the nominal capacitance without application of any external pressure, the plot of $\log(\Delta C/C_0) = f(\log P)$ is shown to be linear, with a slope of 0.58 (**Figure 2D**).

[Insert here Figure 2; single column]

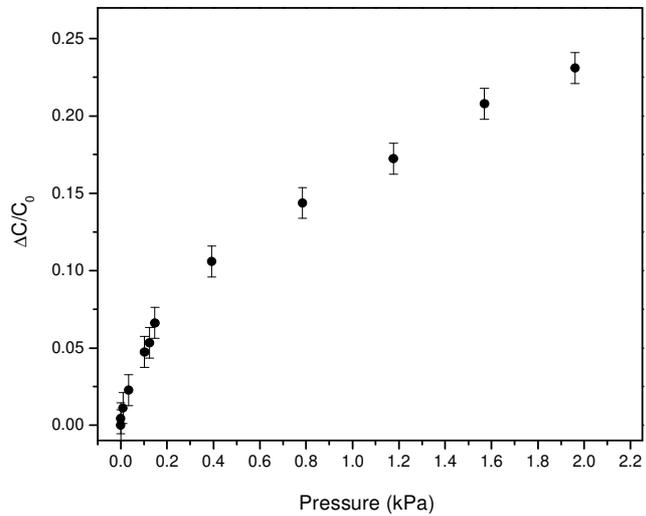
A



B



C



D

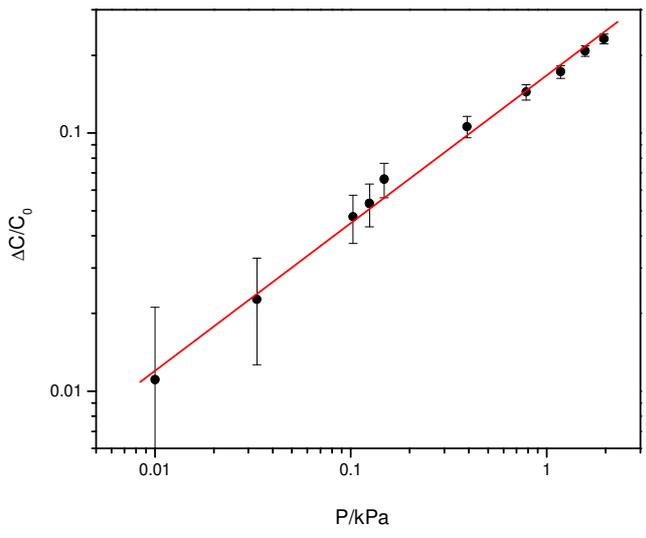


Figure 2. Changes in capacitance C of the sensor upon application of (A) pressures from 0.7 to 150 Pa then (B) from 400 Pa up to 2kPa. (C) Calibration curve of the relative capacitance changes obtained from the previous measurements. (D) Linear log-log plot of the relative capacitance change, showing a slope of 0.58.

The stability of the sensor's response was evaluated under repetitive application of a pressure (see **Figure S3**). Reported sensitivities for such kind of sensor range from few hundredth of kPa^{-1} to 1 kPa^{-1} , depending on the architecture and the materials used. Our device shows sensitivities in line with these values, of ca. 0.5 kPa^{-1} for pressures below 200 Pa, and of 0.08 kPa^{-1} for pressures up to 2 kPa. These values are compatible with those expected for a multilayered structure composed of polymeric foils with relatively low dielectric constants (3.4 and 3.5 for Kapton® and PET, respectively).

3.2. Measurements in a dressing

Measurements were performed with the pressure sensor included within two different commercial dressings, one of 72 mm x 19 mm and the other of 60 mm x 70 mm. The **Figure 3A** below shows an example of setup with the sensor inserted on the internal side of the dressing, or below the cotton fabric of the dressing. In these conditions (the dressing being glued on the platen of the probe station), the capacitance at rest is 14.8 pF while it is of 13.7 pF taken alone, which corresponds to a pressure exerted by the dressing of about 200 Pa. When a scotch tape is applied above the dressing to maintain it more firmly on its support and simulate its gluing onto the skin (see **Figure 3B**), the capacitance increases up to 16.7 pF, which corresponds to a pressure of 2 kPa. It should be noticed that since information on body movements depends on capacitance variations ΔC rather than on absolute capacitance values, detection of movements will not be much affected by the dressing presence. At the same time, this experiment shows that the sensor is able to quantify how tightly the dressing is secured on a wound.

[Insert here Figure 3; single column]

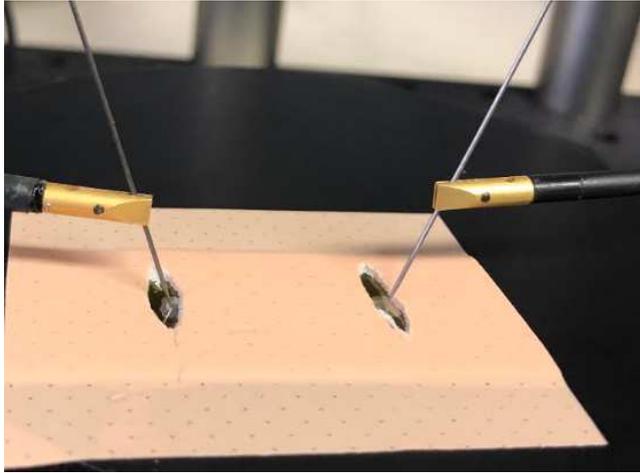
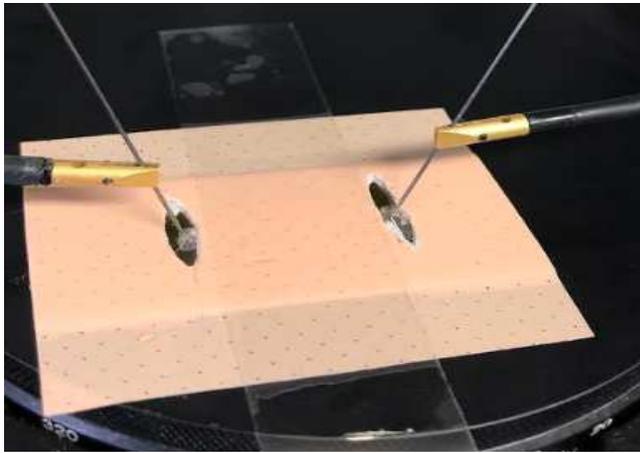
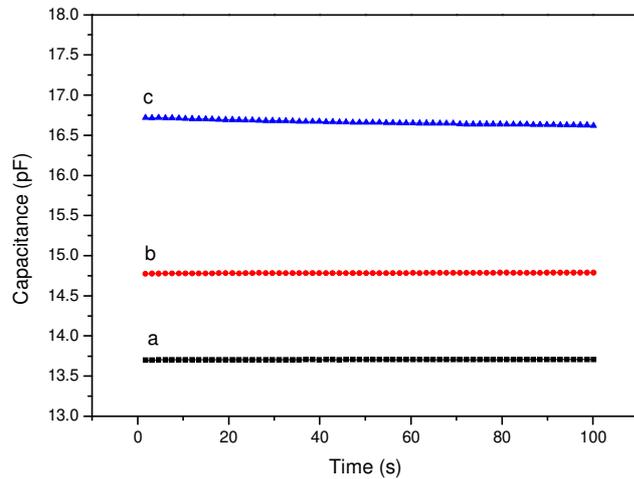
A**B****C**

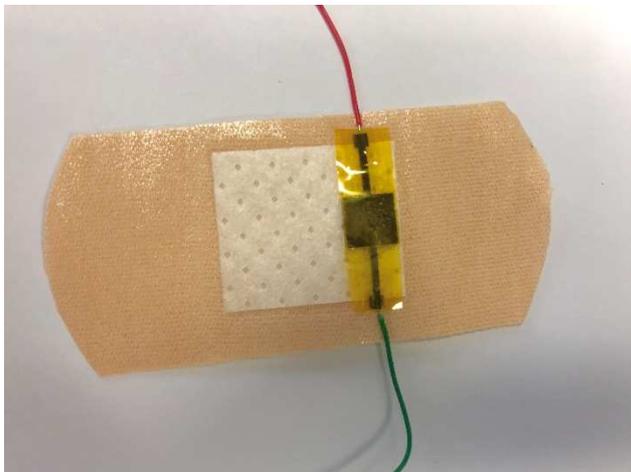
Figure 3. (A) Example of a measurement setup with the sensor included into the dressing, which is loosely glued onto the platen. (B) Same setup, with a scotch tape firmly fixed onto the dressing, to simulate a firm gluing of the dressing. (C) Records of the capacitance for **a**- a sensor taken alone, **b**- the same sensor included into a dressing, and **c**-the same sensor included into a dressing, onto which a scotch tape is firmly glued.

3.3. Measurements on the body

For measurements on the body, the setup utilized above was changed to a setup where the electrodes were no more contacted with pins but with copper wires and an anisotropic conductive tape. To avoid a direct contact between the external electrodes and the skin, a glued Kapton tape was added atop (**Figure 4A**). The sensor was placed on the ankle of a volunteer at various places corresponding to various radii (r) of curvature of the surface, from $r = 15$ cm (the least curved area of the ankle) to $r = 3$ cm (the most curved area). Measurements were made with the volunteer immobile. As shown, the capacitance changes due to radius variations are measurable but remain low ($< 1\%$) compared to changes resulting from the application of a direct pressure. We can conclude that the device remains functional whatever the curvature radius (between 3 to 15 cm) of the surface where it is placed.

[Insert here Figure 4; single column]

A



B

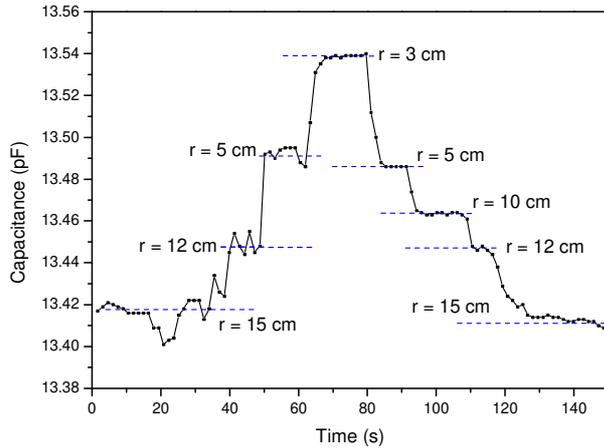


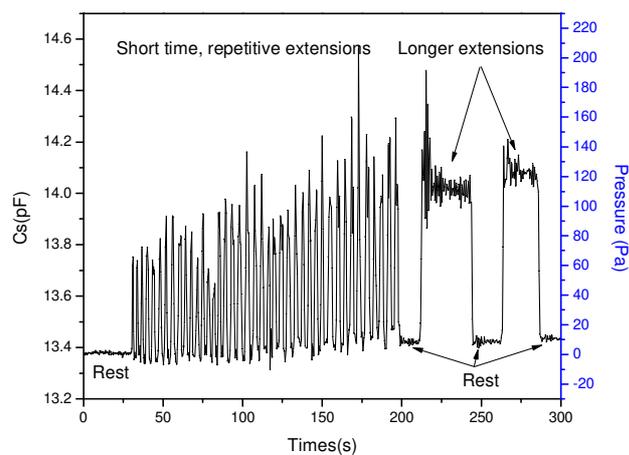
Figure 4. (A) Picture of the sensor with its wires, included in a commercial dressing. (B) Capacitance changes resulting from the placement of a sensor at different places on a volunteer's ankle, corresponding to various radii (r) of curvature, without application of any external pressure. The relative change of capacitance stays lower than 1%.

The sensor was then evaluated for the monitoring of ankle movements, typically during walking. It was placed on the external side of the ankle of a volunteer who, for the first 30 s stayed still, then simulated a walk as on a stepper for about 170 s, then made similar but slower movements i.e., extension of the ankle followed by a rest position (in both cases, no external pressure is applied on the sensor other than that resulting from the change of position of the foot). The capacitance changes and corresponding pressures are given on **Figure 5A**. Pressures are into the 80-160 Pa range, which is expected for ankle movements that do not correspond to direct pressures applied onto the dressing but to shear forces. The same sensor was also embedded into a dressing then glued below a volunteer's plantar for a simulated walk (positions illustrated on **Figure 5B**). The capacitance changes and corresponding pressures are given on **Figure 5C**. At rest, the volunteer's foot was set flat above the ground (no contact); foot down corresponds to the foot laid flat on the ground (leg's weight being applied, highest capacitance value); foot up corresponds to the ankle in extension (foot's fingers on the ground, heel up). As shown, the noise generated by connections is limited to less than 0.1 pF. In this configuration, the dressing is directly subject to the foot pressure if in contact with the floor. In practice, this pressure varies depending on the situation (place of the

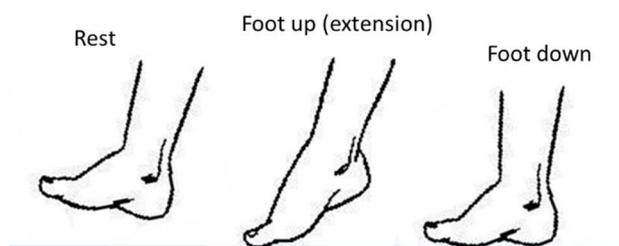
wound under the foot, on a bare foot or in a shoe, etc.) so that foot plantar pressure values may go up to 1,900 kPa. However, it is generally recognized that pressures higher than normal capillary pressure i.e., 12 mmHg (equivalent to 1.6 kPa) compromise blood circulation and cause injuries after a few hours [25]. Therefore, our sensor is well adapted for detecting these pressures up to 2kPa. If higher pressures have to be monitored, the sensor design may be easily changed by intercalating a thicker polyimide foil ($> 13 \mu\text{m}$) or thicker double-sided tapes.

[Insert here Figure 5; single column]

A



B



C

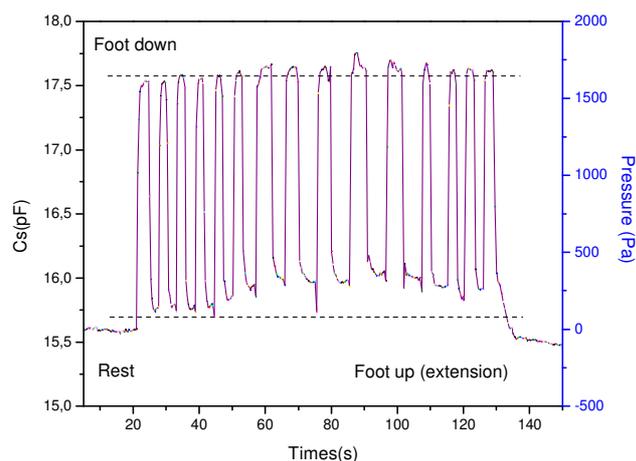


Figure 5. Capacitance changes resulting from a simulated walk as on a stepper for (A) a sensor placed on a volunteer's ankle, at rest (first 30s), for a walk (up to 200s) then for a series of extension of the ankle followed by a rest (up to 300s). (B) Illustration of the three foot positions. (C) Capacitance changes for a sensor embedded into a dressing and glued on the foot plantar, at rest, the foot up, and the foot down on the ground.

The stability of the sensor was also evaluated over a longer period (more than 4h) during which 750 repetitive actions of 650 Pa were applied (**Figure S4**).

4. Conclusion

The inkjet-printed and conformable capacitive sensor reported here is able to measure how tight the dressing into which it is inserted is set in place. It can also measure the amplitude of the movements of an ankle onto which it is glued and can measure the pressure exerted on the foot sole on the ground. Its poor sensitivity to the curvature radius of the support onto which it is glued allows to place it at any place on the body where chronic wounds generally appear. It is fabricated very easily from a commercial PI substrate, a commercial double-side tape and a commercial Ag ink. Its size (for the reported experiments, of 16 mm in length / 7 mm in width) could be reduced down to 2 mm in length / 1 mm in width if necessary, with the same fabrication protocol, depending on the size available on the dressing, the area to monitor and/or the number of sensors to place on the dressing. The forthcoming experiments will

concern the inclusion of several types of sensors on a single dressing (e.g., pressure, humidity, pH...).

Acknowledgements

N.T.N. Dau thanks the French Embassy in Hanoi, Viet Nam, for a PhD scholarship. The authors thank the University of Science and Technology of Hanoi (USTH, grant number USTH.AMSN.01/21-23) as well as the Consortium USTH and the French Embassy in Hanoi for funding part of this work through the 2021's collaborative S&T project funding.

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A fully-printed, flexible pressure sensor is fabricated, included onto the internal face of a commercial dressing, interfaced with a portable Arduino Uno and glued onto various parts of the body of a volunteer, e.g. to follow the pressure permanently applied on a wound by the dressing itself, or due to the movements of the patient.

A Simple Flexible Printed Capacitive Pressure Sensor for Chronic Wound Monitoring

