



# Micromechanical Analysis of Soft Tactile Sensors

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Today, soft materials (e.g., elastomers, polymers, fabrics, etc.), with compliances and extensibilities not allowed by rigid components, are investigated in order to develop artificial tactile sensors that can emulate the adaptability of the natural tissues to objects and environment in general. This research is fueled by the increasing demand for conformable, yet functional and robust, systems in many fields from soft robotics to wearable systems and biomedical devices (Kim et al., 2013).

In the past decades, remarkable results were obtained in the design and fabrication of tactile sensors and, more in general, of *electronic skins (e-skins)* (Someya et al., 2004; Sekitani et al., 2008; Kim et al., 2011; Park et al., 2014; Xu et al., 2014; Zhang et al., 2015). These achievements were fundamental for a plethora of new applications, such as human-machine interfaces, body-integrated electronics, and medical/health-monitoring devices. Also, they offered very advanced technological solutions for robotics and wearable systems. From a mechanical point of view, flexible and stretchable substrates were chosen, while the integrated components went from rigid (Yang et al., 2010; Kim et al., 2011; Xu et al., 2014) to semi-rigid (Engel et al., 2005; Yang et al., 2008) and finally to flexible (Tien et al., 2014; Park et al., 2015; Zhang et al., 2015) and stretchable (Someya et al., 2004; Sekitani et al., 2008; Park et al., 2014; Chou et al., 2015) devices.

In particular, pressure sensors are widely applied in classical robotics, i.e., where the materials of the robot body are rigid. Thus, when positioning the sensors on the hosting structures, the sensors' functionality is not affected by the mechanical characteristics of the robot itself, since the rigid and elastic materials do not store any mechanical energy. Hence, tactile sensors were developed and characterized as independent components that were integrated in the robot at a later stage. On the other hand, especially for integrating pressure sensors in a soft robot or in a wearable system, this approach would not give the desired results. Indeed, the mechanical characteristics of the soft robots or the natural skin, which act as hosting substrates, could reduce or even totally impair the correct functionality of such mechanical sensors. In particular, due to the substrate softness, the largest compression can occur in the substrate rather than in the sensing device. The abovementioned weaknesses can be overcome with an *integrated approach* (i.e., considering the mechanical characteristics of both sensing device and substrate) in the modeling, fabrication, and characterization of such systems. For instance, in the case of wearable systems, the properties of human skin, and underlying tissues and bones, should be taken into account; whereas in soft robotics, the materials used for fabricating them [e.g., mainly elastomers such as polydimethylsiloxane (PDMS) and Ecoflex<sup>®</sup> (Shepherd et al., 2011; Morin et al., 2012)] must be included as part of the working system. From the mechanical point of view, human skin can be approximated as a bilayer tissue standing on a soft substrate, with non-linear and viscoelastic properties (Hendriks et al., 2003) depending on several factors, such as the age, the sex, or the body region (Escoffier et al., 1989). The lower layer (*hypodermis*) has an average thickness around 0.8 mm, with a Young's modulus around 2 kPa. Otherwise, the upper (*dermis*) has a thickness between 1.2 and 1.5 mm, with a Young's modulus around 35 kPa, while the most superficial layer (*epidermis*) can be neglected from a mechanical point of view, due to its very low thickness (around 80 μm). Finally, the subcutaneous tissue (*muscle*) has an average Young's modulus of 80 kPa (Pailler-Mattei et al., 2008). Looking at the soft robotics side, PDMS and Ecoflex<sup>®</sup> present non-linear and

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viscoelastic behaviors for high strain rates, while the stress–strain relationships are almost linear for strain under 40% (Case et al., 2015). In particular, PDMS is a transparent silicone with a Young’s modulus varying from 100 kPa to 4 MPa, mainly due to its curing temperature and to the mix ratio of prepolymers. Also, Ecoflex® has a Young’s modulus around 30–40 kPa (Case et al., 2015), and it is one of the very few materials available for mimicking the human skin mechanical behavior (Amjadi et al., 2015).

Hence, modeling the behavior of a soft pressure sensor on a multilayer soft substrate can be not trivial. In our opinion, analyzing the correlation between the mechanical and functional (e.g., electrical) behavior of sensors made of soft materials is mandatory. In particular, it is important to understand how the mechanical properties of the constituent materials of sensors affect the electrical response and to evaluate the influence of the soft structures and substrates in which they will be embedded. Then, these characteristics can be exploited to encode tactile cues more efficiently and with simpler designs. This requires the development of non-linear mechanical models, also considering the way the mechanical response of such materials is typically described in literature, i.e., with hyperelastic models like the Mooney–Rivlin, Ogden, or Fung (Fung, 1993; Holzapfel, 2000).

In the overall vision of designing innovative smart devices for soft robotics and wearable applications in the near future, here we present a case study evaluating the behavior of a skin-like soft pressure sensor when stimulated by external loads quasi-statically (i.e., the indentation velocity is low). This way, some theoretical insights can be given for future analyses of more complex stimuli. The sensor, shown in **Figure 1A**, is made of different soft layers. It is highly sensitive and can detect normal forces in the 20 N range. Since the sensor area is 8 mm × 8 mm, the pressure range is around 300 kPa. The device structure, sketched in **Figure 1B**, is built by integrating in between them two non-stretchable conductive textile electrodes (Zelt fabric, Mindsets Ltd., Whaltam Abbey, UK) and a thin film of Ecoflex®. It is a super soft silicone elastomer (Smooth-On 0010, USA), which acts as a deformable dielectric layer. The whole structure is embedded between two PDMS (Sylgard 184, Dow Corning, Midland, TX, USA) layers that insulate the textile electrodes while providing a soft and

compliant surface to outer stimuli. In our case, we used a 1:10 mix ratio, with the films cured at room temperature for at least 24 h before the use. Regarding elastomeric layers, the corresponding Young’s modulus is around 3 MPa for PDMS and 30 kPa for Ecoflex. Both values have been measured by means of indentation tests. Moreover, a particular aspect that affects sensor response is given by the textile electrodes: they consist of copper/tin-coated woven fabric. The architecture of a woven fabric is such that small air volumes (i.e., sheds) are embedded among the warp and weft that confer unique properties to the tactile sensor. Indeed, a microporous structure is obtained in the active region of the device. When a pressure is applied, together with the compression of elastomeric layers, the sheds’ volume in the stimulated area is reduced relevantly, thereby resulting in an enhanced sensor response for a wider pressure range (Viry et al., 2014).

A pressure applied on the sensor surface causes the variation of the dielectric layer thickness  $d$ , with respect to  $d_0 = 300 \mu\text{m}$ , giving

$$\frac{\Delta C}{C_0} = \frac{d_0 - d}{d} = \frac{\varepsilon_c}{1 - \varepsilon_c}, \quad (1)$$

being  $C_0 = kA_0/d_0$  the capacitance for null pressure,  $k$  and  $A_0$  the permittivity and the sensor area, respectively, and  $\varepsilon_c = (d_0 - d)/d_0$  the compressive strain of the dielectric layer. The best fitting of the compressive stress ( $\sigma_c$ )–strain ( $\varepsilon_c$ ) experimental curve of the multilayer film is given by the Fung model, with an exponential stress–strain constitutive equation (Fung, 1993), valid for uniaxial response

$$\sigma_c = c_1 \left( e^{c_2 \varepsilon_c} - 1 \right), \quad (2)$$

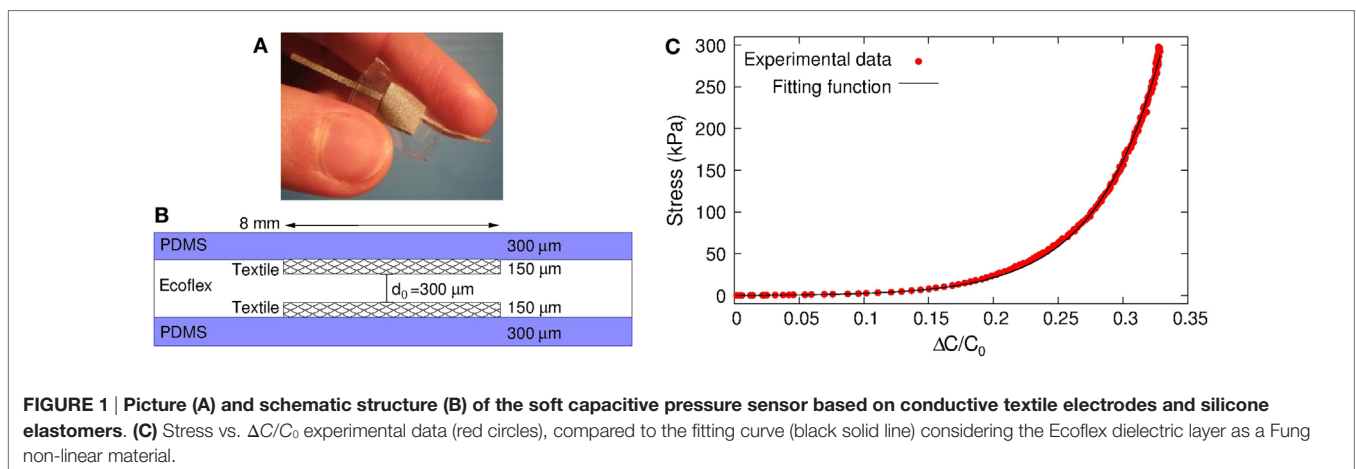
with  $c_1$  and  $c_2$  two material constants.

Then, combining Eqs 1 and 2, we obtain

$$\sigma_c = c_1 \left( e^{c_2 \frac{x}{1+x}} - 1 \right), \quad (3)$$

where  $x = \Delta C/C_0$ .

Equation 3 is exploited for the evaluation of experimental data obtained by indentation tests. In particular, a flat Delrin



8 mm × 8 mm probe was connected to a 6-axis load cell (ATI Nano 17, ATI Industrial Automation Inc., Apex, NC, USA), which acquired force data during the tests. The indentations were made at constant low velocity (10 μm/s) by means of a servo-controlled micrometric translation stage (M111.DG, Physik Instruments, Karlsruhe, Germany), while the capacitive output signals were acquired simultaneously with a custom made printed circuit board based on 24-bits resolution capacitance-to-digital converter (CDC) (AD7747, Analog Devices Inc., Norwood, MA, USA) and PIC32 microcontroller that sends data to a control PC via USB interface. The CDC resolution is 1 fF, while the RMS noise is reduced to around 2 fF by implementing a differential capacitance reading strategy and using coaxial cables. Thus, the minimum detectable signal is around 6 fF (three times the RMS noise), which allows a readout force resolution at the millinewton scale.

In **Figure 1C**, the  $\sigma_c - \Delta C/C_0$  curve, averaged over 12 tests, is fitted using Eq. 3, showing a very good agreement. We obtain  $c_1 = 0.0660$  and  $c_2 = 33.847$  kPa.

To study the influence of each different layer, a first approach could be to limit the analysis to small deformations. In this case, linearizing Eq. 2,  $\sigma_c \simeq c_1 c_2 \varepsilon_c = E \varepsilon_c$ , and the product  $c_1 c_2$  represents the Young's modulus for small deformations. Then, for small strain, the apparent Young's modulus of the whole structure is  $E_a = c_1 c_2 = 2.24$  kPa, well below the nominal value of Ecoflex  $E_0 = 30\text{--}40$  kPa. A direct measurement of the Young modulus of textile by means of indentation tests is not trivial, since the experimental errors can affect the results considerably. Thus, we made tests on a multilayer composed of Ecoflex embedded between two textile films (which correspond to the active region of the sensor), obtaining a multilayer modulus  $E_m = 4$  kPa. Finally, comparing the stiffness of the whole structure to the series of three layers, the Young's modulus  $E_t$  of the textile electrodes can be estimated. Indeed, considering

$$\frac{d_0 + 2d_t}{E_m} = \frac{d_0}{E_0} + \frac{2d_t}{E_t}, \quad (4)$$

where  $d_t = 150$  μm is the textile thickness, we obtain  $E_t = 2.22$  kPa, almost equal to  $E_a$ . This demonstrates the relevance of the textile material in the mechanical behavior of the sensor. Also, as

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already shown in Viry et al. (2014), the air sheds enhance the sensor response at low pressure ranges and considerably enlarge its operating range. Moreover, it is noteworthy to mention that, even if during compression all different layers are subjected to deformation, the capacitance variation is due mainly to the thickness variation of the central sensor region. Therefore, due to the strong non-linear behavior of the whole structure, it is not possible to relate directly the total compressive strain to the measured capacitance. Indeed, the output signal is due to the compression of the central region only. By contrast, this example suggests that such electrical measurements could also be exploited as a tool to evaluate the effective deformation of a single layer embedded in a composite structure, even in the presence of several non-linear materials.

In conclusion, we provided some insight about the kind of complexity introduced by the presence of soft materials in modeling and designing a tactile sensor. Starting from this analysis, future work will be dedicated to quantitatively describe the effect of mechanical behavior of the constituent materials (especially textile) on the sensor response, by using stimuli with different stiffness and shape. In addition, the effect of the soft substrates, such as human skin for wearable systems or elastomers (i.e., PDMS, Ecoflex®) for soft robotic applications, should be included, as mentioned above. In this way, it will be possible to better understand how to encode different tactile features, thus not only force but also, e.g., shape and hardness, with improved but simple sensor designs.

## AUTHOR CONTRIBUTIONS

MT, NP, and LB developed and discussed the model. MT performed experiments. LB supervised the work. All the authors discussed the results and wrote the manuscript preparation.

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