INTRODUCTION
1.1 Conducting polymer in wearable biomedical sensors

The increase interests in wearable device market are triggered by healthcare monitoring. Common examples are pulse, heart rate and temperature monitors. Wearable technology has opened up new path for non-invasive diagnostic and therapeutic technologies via sensing of biomarker/drug from the liquid extracted on skin including sweat (Bandodkar & Wang, 2014; Liu et al., 2017). The increasing demand of integrating electronic technology in wearable devices is driven by needs for individual monitoring remotely at home, often called as ubiquitous health care (Choudhuri et al., 2019). In addition, wearable devices allow continuous, long time monitoring at any place, anytime (Bohr et al., 2019).

Mechanical and electrical responsive conducting polymers (CPs), plus high flexibility and stretch-ability contribute to recent accelerate growth of publications involving conducting polymer in wearable and skin-attachable device. Metals and silica (semiconductors) are inorganic materials that are generally regarded as highly conductive but are rigid and inflexible. The concept of organic conductors/semiconductors has arisen since the discovery of highly conducting polyacetylene by Hideki Shirakawa, working along with Alan MacDiarmid and physicist Alan Heeger in 1977 (Shirakawa et al., 1977). The most apparent advantage of organic electronics as compared to inorganic is that they are highly flexibility and they are lightweight. These properties are ideal for wearable sensors. Conducting polymers with long-term electrical and chemical stability such as polypyrrole (PPy), poly(3,4-ethylenedioxy-thiophene) (PEDOT) and polyaniline (PANI) (Figure 1) have gained popularity in this field (Puiggali-Jou et al., 2019; Talikowska et al., 2019).

Doping counterions in close proximity to the extended pi-bond significantly improve CP conductivity. These doped CPs have electrical conductivities ranging from >1 S/cm to >1000 S/cm aligning CPs with the inorganic semiconductors, for example silicon
and germanium, and comparable to conductors such as graphite (Figure 1a) (Fielding et al., 2015). The most common form of doping is via oxidation, where anions are inserted into CPs such as PPy and PEDOT, resulting in positive charge carriers that move through the polymer structure. To date, doped PEDOT has been reported with the highest electrical conductivity, ranging up to >8000 S/cm (Cho et al., 2014). Adding to this, PEDOT is air stable, reasonably inexpensive and easily synthesized, making it attractive in many application fields from transparent electrodes (Yu et al., 2016), to thermoelectric devices (Fan et al., 2019), to sensors (Çetin & Camurlu, 2018) in biology.

Rapid growth of CPs in biomedical engineering areas includes neural networks (Braendlein et al., 2017; Nielsen et al., 2016; Rivnay et al., 2015), artificial muscles(Khan et al., 2019) and tissue engineering (Talikowska et al., 2019). CPs offer synergistic properties of both polymers and inorganic semiconductors, being lightweight, relatively easy to process and mechanically flexible combined with electrically conductive(Chapman et al., 2020; Park et al., 2019; Puiggali-Jou et al., 2019; Talikowska et al., 2019). These properties are highly sought after for wearable electronic systems which leads to increasing trend of adopting organic electroactive materials for wearables utilizing conducting polymers (CPs) (Nezakati et al., 2015), artificial muscles(Khan et al., 2019) and tissue engineering (Talikowska et al., 2019). CPs offer synergistic properties of both polymers and inorganic semiconductors, being lightweight, relatively easy to process and mechanically flexible combined with electrically conductive(Chapman et al., 2020; Park et al., 2019; Puiggali-Jou et al., 2019; Talikowska et al., 2019). These properties are highly sought after for wearable electronic systems which leads to increasing trend of adopting organic electroactive materials for wearables utilizing conducting polymers (CPs) (Nezakati et al., 2015).

As seen in Table 1, PEDOT role in ISE specially for potentiometric sensors minimize the potential drift of system hence enables sensitivity for 50 ~ 60 mV/dec and large detection range (Anastasova et al., 2017; Gao et al., 2016). Comparable performance is also achieved using normal working electrode, but taking advantage of PEDOT’s electroactivity enhancement during electropolymerization in ionic liquid medium (Matzeu et al., 2016). Incorporation HEPES with PPy contributed by its biocompatibility with biorecognition element improves sensitivity to 35.07 mV/dec for urea detection (Varadharaj & Jampana, 2016). Work by (He et al., 2018) through PANi combination with Cu network enables large variety of analytes, which are urea, uric acid, lactate, glucose, Na+ and K+ simultaneously but suffers from detection limit and ranges for each analytes. Another work of PPy on Cu, this time Cu foil has proven for Ca2+ detection with sensitivity of 2.2 × 10-3 g/L and range to ~40.15 g/L (Zhao et al., 2019).

Major works in OECT involved commercial PEDOT:PSS, taking advantage from hydrophilicity of the solution hence enables facile fabrication methods such as spin coated (Keene et al., 2019) or transfer printed (Zhang et al., 2019) or wet spinning (Lim et al., 2019) onto the transistor.

This particular reason also adopted in electrophysiological sensor as such in Table 2. PEDOT:PSS can be directly inkjet printed on thin tattoo paper for ECG and EMG measurement (Bihar et al., 2018; Ferrari et al., 2018). Further incorporation with other elements such as D-sorbitol is also reported (Spyropoulos et al., 2019). Other than PEDOT:PSS, interfacial polymerization between SF and PPy improves maximum strains up to 300% with 1.91Mpa maximum adhesion (Yang et al., 2020). 60% maximum strains with 250 kPa maximum adhesion is achieved through octopus patterned electrode moulded using Carbon black with PDMS and p3ht mixture.

Next, other type of sensing target such as temperature and electrophysiological signals is introduced.

**FIGURE 1** (a) Conductivity chart of conducting polymer showing greater values than Si, Ge semiconductors and comparable to graphite {103 S/cm}. (b) Chemical structure of the most employed ICPs (PANI, PPy and PEDOT) (adapted with permission from Puiggali-Jou et al., 2019)

### 2 | ELECTROCHEMICAL BASED SENSOR

Chemical sensors target analytes such as chemical compounds found in urine (Wong et al., 2020), blood (Liu, Lee, et al., 2018) and human serum (Yang et al., 2017). These body fluids contain biomarkers, drugs and metabolites that can give valuable information about human health. Current demand of minimal/non-invasive sensing especially in continuous monitoring system has encourages deeper studies through various sensing mediums such as dermal (Liu et al., 2020) and eyes (Campigotto & Lai, 2020). Increasing works on skin sensing via sweat of dermal interstitial fluid (ISF) is contributed by the fact that...
| Transducer | CP       | Method                                                                 | Analyte                        | Sensitivity          | Range         | References               |
|------------|----------|------------------------------------------------------------------------|--------------------------------|----------------------|---------------|--------------------------|
| Potentiometric | PEDOT:PSS | PEDOT electropolymerized on ISE                                      | K⁺, Na⁺                        | Na⁺: 64.2 mV/dec K⁺: 61.3 mV/dec | Na⁺: 10–160 mM K⁺: 1–32 mM | Gao et al. (2016)       |
| Potentiometric | pedot    | PEDOT electropolymerized on ISE                                       | Na⁺                            | 56 ± 1 mV/unit       | -42 ± 1.2 mM  | Anastasova et al. (2017) |
| Potentiometric | pedot    | PEDOT electrodeposition on WE in ionic liquids                       | NaCl                           | 55.5 mV/logaNa⁺      | 10⁻⁵ to 10⁻¹ M | Matzeu et al. (2016)     |
| Potentiometric | ppy      | Ppy electropolymerization on WE with HEPES                            | Urea                           | 35.07 mV/decade      | 10 μM to 5 mM  | Varadharaj and Jampana (2016) |
| Amperometry | pani     | Pani electrodeposited on Cu network                                   | Urea, uric acid, lactate, glucose, Na⁺ and K⁺ | NA                  | NA            | He et al. (2018)         |
| Amperometry | pani     | Pani electropolymerization on Cu foil                                | Ca²⁺                           | 2.2 × 10⁻³ g/L       | -40.15 g/L    | Zhao et al. (2019)       |
| Amperometry | ppy      | Ppy electropolymerization on WE with MWCNT and lactate oxidase       | Lactate                        | 0.0778 μA/μM         | 5–60 μM       | Meshram et al. (2015)    |
| Impedimetric | pani     | Pani electropolymerization in pNIPAM microgel                        | Lactate                        | NA                  | NaCl: 3–80 mM Lactate: 1–25 mM | Mugo et al. (2020)     |
| OECT       | ppy      | ppy in situ polymerization on fibre                                   | Dopamine                       | 47.28 NCR/decade     | 1 nM to 1 μM  | Qing et al. (2019)       |
| OECT       | pedot pss| PEDOT:PSS spin coated on ion-selective membrane                       | NH₄⁺ and Ca²⁺                  | NA                  | 0.01 × 10⁻³ and 100 × 10⁻³ mM | Keene et al. (2019) |
| OECT       | PEDOT:PSS| PEDOT:PSS doped DBSA layered with PVA hydrogel to transfer printing | Glucose                        | NA                  | 1 × 10⁻⁶ M to 0.1 M | Zhang et al. (2019)     |
| OECT       | PEDOT:PSS| PEDOT:PSS wet spinning into sulphuric acid                           | NaCl                           | NA                  | NA            | Lim et al. (2019)        |
ISF provides accurate information and is reported to have comparable analytes as in blood (Miller et al., 2018). Less invasive sweat sensing has attracted significant interest recently. Chemical analytes such as electrolytes including salt (Matzeu et al., 2016), metabolites, urea (Liu, Liu, et al., 2018) and lactate (Meshram et al., 2015) or biomarker including glucose (Wang et al., 2017) or even drug like dopamine (Oh et al., 2017) are common target analytes for human health monitoring. These biological analytes (widely referred to biosensor) usually involved biological recognition element such as enzymes, antibodies, living cells and tissues (Torres et al., 2020).

Rapid analysis, low cost and robust operation offered by electrochemical sensing has opened new path for on-skin sensing specially to cater for continuous monitoring of chemical analytes in the sweat. The past decade has seen rapid development of CPs nanostructures and nanocomposites deposited on the working electrode in an electrochemical sensor because of the CPs biocompatibility, chemical stability, electrochemical activity, high electron mobility and high surface-to-volume ratio (Aydemir et al., 2016). The sensor performance is heavily affected by electroactive materials deposited at electrode’s surface to better facilitate electron transfer and improve recognition element immobilization. To date, a variety of CP composites has been reported to decrease the limit of detection (LOD), such as CP incorporating metal-Au (Teixeira et al., 2019), metal oxides-ZnO (Hatamluyi et al., 2019) and graphene oxide (Hadi et al., 2017), thus enhancing biosensitivity. CP compatibility allows it to be deposited with variety of biological composite such as enzyme, DNA or protein.

In general, electrochemical transducers can be potentiometric, amperometric, conductometric or impedimetric. The characterization of these transducers depends on their outputs, which are voltage (potential), current, conductance and impedance, respectively. The study of CPs in electrochemical wearable sensors was outlined in next section according to their transduction mechanism.

### 2.1 Potentiometric sensor

An ion-selective electrode (ISEs) is commonly used in sweat analyte detection due to its simple construction (usually made up by two electrodes), low cost and robustness. The ion-selective membrane improves sensitivity towards targeted ions. Sodium ions in sweat are common in continuous health monitoring sensors because they can detect electrolyte imbalance which impacts athletes working in hot and humid environments (Anastasova et al., 2017). NaCl levels in sweat can be used to help diagnose cystic fibrosis which causes accumulation of mucus in lungs and intestines resulted from malfunction of chloride channel (Jadoon et al., 2015). ISE-based potentiometric sensor for Na⁺ detection using PEDOT has been reported where PEDOT:PSS is deposited on the sweat as ion to electron transducer in the ISE to minimize potential drift of the ISE (Gao et al., 2016). PEDOT is electropolymerized on ISEs working electrode by applying constant current of sweat dipped into EDOT and NaPSS solution. In Gao et al., the potentiometric sensors for ion detection achieved sensitivities of 64.2 mV and 61.3 mV per decade of concentration for Na⁺ and K⁺ sensors within concentration of 10–160-mM Na⁺ and 1–32-mM K⁺. The robust and miniature design also allows integration with many other sensors, for example glucose, lactate and temperature for on-skin multiplexed sweat sensing, all integrated on PDMS layer strapped around wrist as seen in Figure 2.

**FIGURE 2** Schematic of the sensor array (including glucose, lactate, sodium, potassium and temperature sensors) for multiplexed perspiration analysis (adapted with permission from Gao et al., 2016)
2.2 | Amperometry sensor

Amperometry electrochemical sensors monitor current generated when electrons are exchanged in the system especially towards sensing of redox active target. Reduction or oxidation of the target results in electron exchange, which is then detected by the electrochemical sensor as output current. In enzymatic amperometry sensor for biosensing, redox of target is assisted by enzyme immobilized on the electrode. Different enzymes immobilized on PANi deposited working electrodes enables dual function, that is multiplexed sensing and energy harvesting (He et al., 2018). PANi is deposited on Cu network patterned PDMS by electropolymerization. During bending and relaxing, electrostatic induced between PANi and water produces current, namely triboelectric effect as seen in Figure 3.

Each working electrodes consists of PANi deposited enzymes of urease, glucose oxidase, uricase and β-galactosidase to detect urea, glucose, uric acid and lactate from perspiration, respectively. Higher concentrations of analytes resulted in protonated/deprotonated of PANi due to product of the enzymatic reaction induces changes to surface chemical state of PANi, thus influenced triboelectrification effect. The triboelectrification between PANi and water, assisted by enzymatic reaction coupling of enzymes immobilized on PANi electrode, enables energy harvesting from human motion and multiplexed biosensing of metabolites. Plus, rough surface of PANi increases active surface area.

2.3 | Impedimetric sensor

Impedimetric sensor reports analyte activity through the output of electrical impedance, for example inductance or capacitance. A non-enzymatic capacitive sweat sensor for dual sensing of NaCl and lactate is reported (Mugo et al., 2020) using stimuli responsive polymer as switch. First, a salt responsive hydrogel carrier using poly...
(N-isopropylacrylamide) pNIPAM is used to detect NaCl in sweat due to pNIPAM microgel collapse upon increasing of NaCl ions due to electrostatic repulsion. Then, by coating pNIPAM with lactate imprinted PANi/phenylboronic acid (PANi/PBA), the sensitivity of microgel towards salt is switched off, turning to sensitivity towards lactate. The lactate imprinted film consists of conductive nanoporous carbon nanotube (CNT)/cellulose nanocrystal (CNC) and the aromatic cyclic structure of PANi (PANI) polymerized by chemical oxidation with sulphuric acid and ammonium persulphate. Design of the capacitive sensor is seen in Figure 4 where CNT/CNT deposited on PVA sheet before pNIPAM microgel suspension layered on the nano porous CNT/CNC film.

Here, CNT/CNC in microgel enhances electrochemical activity for capacitive sensor. The electrode based on multiwalled (CNT) using electropolymerized PPy for lactate detection has shown good binding of the conducting polymer with biomaterial. Due to PANi filling up nanopores of the CNT matrix, electroactive PANi increases film’s conductivity. The linear concentration range obtained for pNIPAM @CNC/CNT sensor in response to NaCl was 3–80 mM, with a limit of detection (LOD) of 0.23 mM NaCl. The lactate imprinted PANi/PBA-pNIPAM @CNC/CNT sensor resulted in a linear range of 1 mM-25 mM and a limit of detection (LOD) of 0.10 mM for lactate detection in sweat.

3 | ORGANIC ELECTROCHEMICAL TRANSISTOR

The term organic electrochemical transistor (OECT) arises from using conducting polymer as the channel connecting drain to source in a transistor-based sensor. OECT integrates the chemical-electrical transduction from electrochemical and miniature amplifiers by the transistor thus has been adopted in many biocompatible sensing applications (Braendlein et al., 2017). High ionic conduction by conducting polymers has been introduced in OECT for biomedical application (Inal et al., 2017). An electrolyte containing sensing target (usually charged ions) is built between gate to channel when using OECT sensors. When voltage is applied to the gate, target ions injected into the conducting polymer channel, hence causes de(doping) that change charge carrier density in the channel, thus manipulate the output current. In addition, an ion-selective layer can be deposited at gate electrode to allow selected sensing targets to be injected into the semiconductor channel which controls the output current of the transistor (Talikowska et al., 2019). This can increase selectivity and sensitivity of the OECTs towards their respected targets. The design simplicity, robustness and miniature size of OECT made good potential for skin-attachable sensor.

3.1 | Tattoo OECT sensor

A tattoo-based OECT (Zhang et al., 2019) reported using facile hydrogel transfer printing method that allows customization of PEDOT:PSS on rigid substrate before transferred to soft substrate, in this work the tattoo paper. Here, PEDOT: PSS is doped with decylbenzene sulphonic acid (DBSA) which also acts as surfactant before spin coated on the glass. To peel off polymer layer from glass, polyvinyl alcohol (PVA) is layered on the polymer. While surfactant decreases adhesion between polymer film and glass, hydrogel carrier (PVA) has strong adhesion to PEDOT: PSS thus enables polymer to lift off from glass in hydrogel manner before transferred to the tattoo paper. For the design of OECT, the drain and source

![Figure 4](image-url) (a) Structural schematic of the synthesis of the pNIPAM microgel; (b) Schematic of the components of the capacitive sensor depicting PANi@CNC/CNT and PANi/PBA-pNIPAM@CNC/CNT printed in series on a PVA sheet for NaCl and lactate detections, respectively; (c) Photograph showing the set up for testing the miniaturized sensor, with an inset photograph demonstrating the size of the sensor patch compared to a Canadian 5 cent coin (adapted with permission from Mugo et al., 2020)
electrodes are first patterned on tattoo paper before transfer printing of PEDOT:PSS thin film between the electrode. The whole device then easily tattooed on the skin from the tattoo paper. The overall facile procedure is seen in Figure 5. Conductivity of 400 $\text{S/cm}$ was calculated for PEDOT:PSS films transferred on the target substrates which was comparable to those values on glass. This work demonstrates the potential of OECT in glucose sensing with sensing range from $1 \times 10^{-6}$ M to 0.1 M.

Another tattoo-based OECT is reported using highly stretchable and flexible styrene–ethylene–butylene–styrene (SEBS) elastomer with microcapillary sweat acquisition layer to provide sample for sensing. An ion-selective membrane is be deposited on the conducting polymer channel, that is PEDOT:PSS, to enhance OECT selectivity and enable simultaneous detection of $\text{NH}_4^+$ and $\text{Ca}^{2+}$ ions (Keene et al., 2019) with wide working range between $0.01 \times 10^{-3}$ and $100 \times 10^{-3}$ mM. In one example, PEDOT:PSS is spin coated on OECT electrode as channel electrochemical transducing layer functionalized with ion-selective membrane (Figure 6).

### 3.2 | Fibre OECT sensor

A fibre-based OECT is reported for dopamine monitoring in sweat taking advantage of 3D structure of fibre coated with PPy (Qing...
et al., 2019). A commercial PA6 filament and pyrrole is immersed in nanofibre suspension prior to in situ polymerization of the polymer to get PPy nanofibre network deposited on the filament surface, resulted in large surface-to-volume ratio. To use as OECT, pair of fibre strand acts as drain and source (PPy coated on fibre surface as channel) strapped around the wrist or forehead as seen in Figure 7 to contact with sweat as electrolyte. The range of detection is from 1 nM to 1 μM with sensitivity of 47.28 normalized current response NCR per decade with response time of 0.5 s.

4 | TEMPERATURE SENSOR

On-skin temperature sensor employs thermosensitive materials that changes their resistivity correspond to temperature change. In regard to conducting polymer, increasing of temperature leads to increased thermally assisted charge carrier hopping thus decrease in resistivity. Temperature-sensitive of PEDOT:PSS is well adopted in resistor type electronic with varying resistance (Culebras et al., 2016) due to structural change of microstructure within PEDOT:PSS polymer (Vuorinen et al., 2016). In PEDOT:PSS, PEDOT nanocrystal act as core and surrounded by PSS-rich shells which also acts as insulating boundaries. When temperature is high, the insulating PSS boundaries are reduced thus reduced the resistance. Otherwise, at low temperature, charge carriers may not possess sufficient thermal energy to overcome these boundaries thus increasing the resistance. The water soluble, commercial PEDOT: PSS makes it compatible as inductive ink for inkjet printing. Further conductivity enhancement is realizable by adding carbon based conducting material, for example graphene (Vuorinen et al., 2016) or graphene oxide (Zhou et al., 2020).

Skin-attachable temperature sensor with simple approach using inkjet printed the PEDOT/graphene ink on commercial skin formable bandage plaster as seen in Figure 8. The temperature sensor shows sensitivity of 0.06% of normalized resistance per degree Celsius with 20 seconds response time for 2 deg incremental. Faster response time (3.5 s) is reported using PEDOT:PSS nanowires/graphene oxide (Zhou et al., 2020). The PEDOT:PSS/GO ink is dropped on nanogroove PET substrate to realize the nanostructure polymer.

5 | ELECTROPHYSIOLOGICAL SENSOR

Electrophysiology refers to study of electrical signal emits by biological cells and tissue in the body, while electrophysiological sensor refers to detection of electrical activity generated mostly by brain, that is electroencephalogram (EEG), heart, that is electrocardiogram (ECG) and muscle, that is electromyogram (EMG). Most of these electrophysiological sensors require rigid, bulk electrodes attached to part of body such as chest and head using several types of adhesions. They are affixed to skin by tape, cloth or sticky pads. The rigidity and bulkiness of conventional electrode is limiting for long-term monitoring.

In conventional electrophysiological monitoring system, commercial electrode often applied medical gel to maintain stable connection between rigid electrode and deformed skin. However, wet-gelled approaches may be problematic when sweat is present. This would hinder affixation of electrode to skin. Using water to enhance interfacial polymerization
between silk fibroin (SF) and PPy (PPy) result in stretching uniformly plus water (in sweat) enhance adhesion of patch to skin due to mechanical property match between conductive (PPy) and adhesive layer (SF) (Yang et al., 2020). Interfacial polymerization is executed using SF gel, made by formic acid solution of SF (adhesive layer), FeCl₃ (PPy polymerization initiator) and CaCl₂ (adjust mechanical property of SF gel). CAPE refers to conformal and adhesive polymer electrode where interfacial polymerization using SF gel that acts as interlock between PPy/SF. Plus, wet surface of SC increases its adhesion thus during sweaty condition, the electrode shows higher adhesion as compared to other gel electrode. The adhesive strength of 1.91 MPa between PPy and SF gel is 27 times higher than 0.07 MPa of gel and Ag/AgCl in commercial gel electrodes as seen in Figure 9.

As the relative humidity increases (referring to moisturization of skin, i.e. sweating condition), its adhesion energy increased to 40 J/m², while the adhesion energy of commercial gel electrodes decreased to 2.6 J/m². The sensitivity of ECG electrodes is defined as the relative voltage ratio of T wave to R wave measured. The sensitivity of CAPE is 0.50 due to the conformal electrodes, compared to 0.31 achieved by commercial gel electrodes. However, when the RH is below 31%, the adhesion energy of CAPE is smaller than 0.9 J/m² and could not adhere to skin.

The demand for adaption of electrophysiological measurement into wearable devices also been increasing in the recent works due to continuous monitoring may help reduce risk of heart attack or epilepsy. To allow this, further development of flexible and stretchable devices is required to ensure affixation and stable of signal. The common soft adhesive electrolyte gel usually used between solid electrode and soft skin, but they are not elastic. Stretchable hydrogel realized by double network hydrogel has been reported and made into conductive gel using conducting polymer (Nagamine et al., 2016). A double network hydrogel is placed on electrode Au-PEDOT/PU film which then electropolymerized with another layer of PEDOT:Cl to anchor electrode to the hydrogel film. Electrode maintains contact on soft skin by conformable conductive hydrogel double network/Au electrode. PEDOT/PU polymerized on Au patterned glass substrate by thermal vapour polymerization; then, DN hydrogel sheet placed on Au-PEDOT/PU film. Then, another layer of PEDOT:Cl electropolymerized on the film to anchor electrode to hydrogel film. By lowering electrode/hydrogel interfacial impedance, PEDOT/PU/AU has 27 times lower impedance (~100 ohm) compared bare Au electrode (~1000 ohm) and with stable resistance of 35 ± 5 sq−1 even during successive 20% stretching.

Dry contact is reported using tattoo electrode also more stable over time compared to standard wet-gelled electrode. Tattoo paper offers seamless interfacing between skin and electrode due to its
ultrathin and ultraconformable property. In this work, solution of PEDOT:PSS is inkjet printed on tattoo paper followed by lamination process and laser cut to have multilayer and patterned electrode. The inkjet printing method enables low cost, large area and facile patterning of multielectrode arrays (Ferrari et al., 2018). On the temporary tattoo paper, substrate of ethyl cellulose layer had a constant thickness of 450 nm; thus, overall electrode thickness ranged from ∼600 to ∼1200 nm, with a consequent drop of sheet resistance from Rs = 520–50 Ω/sq. While the electrodes have firm and stable contact with skin, however the gap due electrical contact (interconnector) that bridge between sensor and external system, in this work using Pt film with the tattooed electrode might hinders the transmitted signal as seen in Figure 10.

Connection between tattoo electrode and interconnect pad in one step fabrication can eliminates the use of rigid interconnect (Bihar et al., 2018). Similarly, PEDOT:PSS is inkjet printed on temporary tattoo paper before transferred to skin, but in addition the conductive polymer also printed on textile as interconnection between electrode and external system. The electrical conductivity of a 95 ± 5 nm thick printed PEDOT:PSS electrode (single layer) is 240 ± 10 S/cm and can be further increases by layering more printed layer. However, since the ink is aqueous, the printed film is instable on the tattoo paper since the sacrificial layer can be dissolve by the aqueous ink.

6 | CONCLUSION

In this review, the potential of conducting polymers in wearable devices focusing on sensing is discussed. The electrically controlled of CPs through de-doping has been used in chemical sensing of metabolites in sweat. The stretchable and flexible property of CP can match with elastomer or hydrogel for the tactile sensing. Furthermore, conformal and nanostructured CP helps increase stable signal between electrode and soft skin for the electrophysiological monitoring. Aside from that, solution processable and in situ polymerization of polymers contributes to ease of fabrication in various types of wearable devices, for example patch, strap, tattoo and hydrogel. Solution of CP especially commercial PEDOT:PSS is printable on soft substrate. The polymer can be blended and grown with/within other chemical compounds such as hydrogel precursor, acid, surfactant before transferred to other substrates.

CONFLICT OF INTEREST

Conflict of Interest: No conflict of interest has been declared by the author(s).

ORCID

Siti Musliha Ajmal Mokhtar https://orcid.org/0000-0002-3975-2889
Eva Alvarez de Eulate https://orcid.org/0000-0002-1634-5498
Miko Yamada https://orcid.org/0000-0001-8795-4461
Tarl W. Prow https://orcid.org/0000-0003-2892-6238
Drew R. Evans https://orcid.org/0000-0002-1525-2249

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ORCID

Siti Musliha Ajmal Mokhtar https://orcid.org/0000-0002-3975-2889
Eva Alvarez de Eulate https://orcid.org/0000-0002-1634-5498
Miko Yamada https://orcid.org/0000-0001-8795-4461
Tarl W. Prow https://orcid.org/0000-0003-2892-6238
Drew R. Evans https://orcid.org/0000-0002-1525-2249

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