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Review

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Convergence of biocompatible printed electronics and sensing in wound dressings: a leap forward in sustainable health monitoring

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Manoj Jose[1,2,3] ✉, Mani Teja Vijjapu[1], Lukas Neumaier[1], Lukas Rauter[1], Anna Heba Chakkunny[1], Daniel Corzo[1], Ronald Thoelen[2,3], Antoni Picard[4], Jürgen Kosel[1] & Wim Deferme[2,3] ✉

The healthcare system is moving away from traditional hospital-centric models towards a more personalised, patient-centric approach driven by the concept called 'lab on wearables'. The nucleus of this concept is grounded on the translation of biological signals into actionable healing information with the help of soft, conformable and biocompatible sensors. This soft flexible electronic platform development is more leaning towards unconventional electronics fabrication routes like printed electronics over clean room based micro-electronics manufacturing. Printed electronics can harness the potential of stretchable foils, bio-derived functional materials and organic electronics, enabling the development of biodegradable and bioresorbable wound monitoring systems that are conformable with the skin. The review explores the potential of sustainable and biocompatible printed electronics in transducing wound biomarkers into actionable healing insights, enabling timely interventions. This work also provides a roadmap for printed electronics-based wound monitoring and on-demand treatment solutions, offering a glimpse into the future promises of the technology.

Wound healing especially for chronic wounds is a severe physical, biological, psychological and economic burden for human mankind. 18% of the diabetes population have chronic wounds which correlates to at least 6 million people in Europe alone. The mortality rate due to diabetic foot ulcer complications is similar to that due to common types of cancer, which demonstrates the seriousness of this issue[1]. Moreover, industrialised countries spend 2–4% of their total healthcare budget on chronic wound treatments[2]. Even today, wound diagnosis and treatments are highly dependent and vary upon the physicians' experience and ability to recognise the wound status from its colour, size and odour. Although these are clinically proven practices, this is inadequate to identify and interpret the biomolecular and biophysical changes deep inside the wound and early detection of infection[3–6]. Such aspects of wound healing are key for effective diagnosis and in-time treatments for better wound management. As the complexity in wound dynamics rises, diagnosis extends to wound swab collection or tissue biopsy from the wound site, further assessment and analysis at molecular testing labs which is a time-consuming, labour intensive and inconvenient approach[6]. In every wound evaluation occasion, health practitioners opening the dressing for inspection can disturb the wound, leading to impairment and delays in curing and increasing the chances for infection[7]. To comply with the next generation healthcare, wound treatments demand non-invasive wound monitoring point-of-care devices to offer better wound management[8,9]. And as the world's aged population is growing, the chronic wound complications are expected to rise and the responsible agencies are required to invest millions of euros for the research and development of smart wound dressings. Such smart and intelligent wound dressings could make comprehensive changes by informing about the wound status in real-time, its closure and infections if any and assisting to keep records of the wound healing history of that patient. Further, it benefits to prevent repeated dressing changes, unwanted stress on the wound, frequent hospital visits and aids the shortening of the nursing time, reducing the wastage of dressing material and improving patients' ease of life. This review paper focuses on the advancements in printed flexible electronics, sensing modalities and communication technologies to facilitate a paradigm shift from

[1]Silicon Austria Labs GmbH, Villach, Austria. [2]Hasselt University, Institute for Materials Research (imo-imomec), Hasselt, Belgium. [3]imec, imo-imomec, Diepenbeek, Belgium. [4]Department of Informatics and Micro Systems Technology, University of Applied Sciences Kaiserslautern, Zweibrücken, Germany. ✉e-mail: manoj.jose@silicon-austria.com; wim.deferme@uhasselt.be

hospital-rooted health diagnosis towards wearable, personal health monitoring labs and to overcome the black box status of a wound.

Wound monitoring biomarkers

A wound on the skin can be defined as the injury or disruption of the anatomy and the functioning of the skin and ranges from a simple cut occurred on the epithelium tissue layer extending to damages on the integrity of the subcutaneous tissue. Wound healing involves the spatial and temporal organisation and coordination of a span of cell types with diverse roles in the different steps of the healing progress (Fig. 1a). Effective healing demands interaction and interplay of many different cell types and this process needs to be composed and controlled at multiple levels[10]. The main goal of wound management is real-time monitoring for timely intervention. The wound monitoring and wound healing optimisation can be carried out on the basis of biophysical, biochemical feedback evolving from the wound.

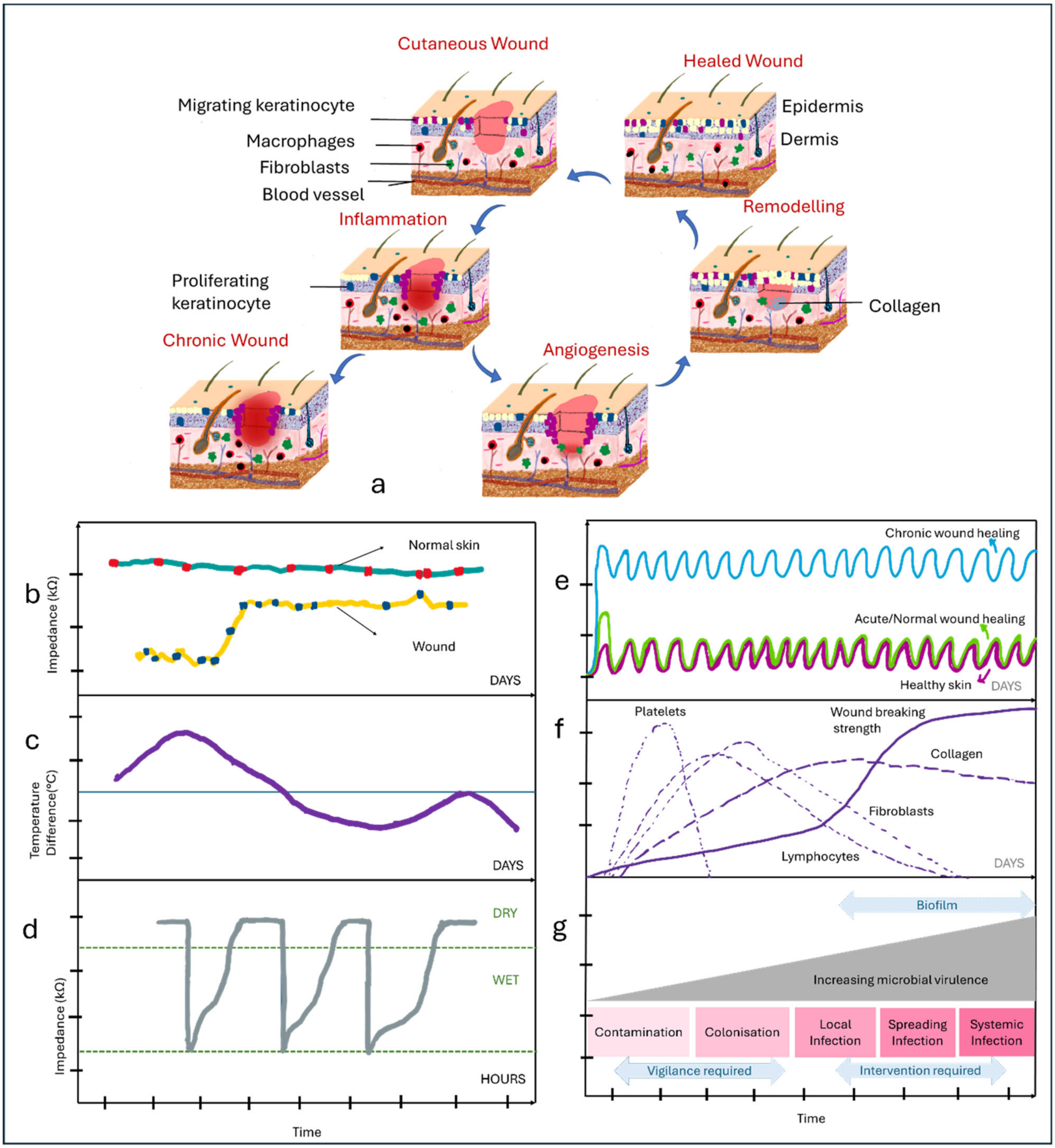


Fig. 1 | Important wound healing biomarkers. The biological wound healing stages (**a**) and the important wound biomarkers are depicted in qualitative plots (**b**–**g**). The electrical properties of a wound will change over its wound-healing period. Typical biological cell has some electrical characteristics, and it can be used for wound monitoring with a simple impedance measurement across the wound. It can show the healing progress where the reference measurement is shown for normal skin (**b**),[480]. The local temperature of a wound varies during wound healing. The plot indicates the variation of wound temperature over several days of healing (**c**)[481,482]. Moisture measurement of wound dressing is plotted (**e**)[483]. pH variation of acute and chronic wounds are shown in (**c**)[484]. The representation of cellular changes once the wound is formed is shown in (**f**)[485]. Finally, different stages of wound bacterial infection are depicted[486] (inset)(**g**).

Biophysical biomarkers. The biophysical properties of the wound varies over the wound healing phases and it has great significance in wound monitoring. The biophysical biomarkers include wound size, temperature, moisture and strain are discussed here.

The primary wound diagnosis parameters are size, colour and odour of the wound as the basic tool for diagnosis[11]. As the healing progresses, wound size (opening and depth) decreases, and infection causes variations in its colour and odour and causes swelling[12,13]. However, these methods do not provide detailed information about the internal dynamics of the wound. Additionally, the physician needs to frequently open the wound for diagnosis. Meantime, these wounds make changes in the bioelectrical properties of the skin during healing. The bioelectrical properties of a wound originate from living cells where the electronic equivalent models describe the tissue cells as constant phase elements where the cell membrane forms a capacitive element and the intracellular and extracellular fluid act as resistive elements[14,15]. The bioelectrical properties of the wound can provide evidence regarding the size of the damage occurred on the tissue and the progress in wound healing (Fig. 1b). Likewise, it is a marker for monitoring the epithelization, proliferation and maturation of new cells through the bioelectrical profile of the wound[16–18]. Monitoring the bioelectrical properties of the wound contributes to

a. Tracking of wound size and closure;
b. Minimise the wound dressing opening;
c. Monitoring of the growth and maturation of new cells.

The temperature of the wound is another pertinent wound monitoring parameter. Studies of Horzic et al. have shown a rise in temperature for the first 3 days after the wound formation, which is an indication of the inflammatory period (Fig. 1c). Generally, the temperature decreases in the following four days. The persistence of an elevated temperature even after the first 3 days is considered to be an infection or disturbance in the wound[19]. Fierheller and Sibbald explored the relation of peri-wound temperature and infection in chronic wounds and the studies show that the infected area causes an average temperature difference of 2.46 °C compared to the normal skin[20]. It is also found that wound temperature decreases with respect to the rest of the body, which might be a sign of low collagen deposition and reduced late-phase 'regenerative' inflammatory cells and fibroblasts[21]. The temperature of the wound plays a vital role as a monitoring parameter and for optimising the healing[22] and it can be used as follows;

a. To understand the cellular dynamics of the wound;
b. To detect infection ;
c. To ensure the wound's critical temperature level is maintained, for regular cellular activity[23].

In 1962 G. De Winter for the first time emphasised the significance of moisture in the wound site[24].The moisture presence in the wound could enhance the healing process, particularly it encourages new tissue growth and ease the pain. A balanced moist wound environment facilitates the action of growth factors, cytokines and chemokines and boosts cellular growth and collagen proliferation[25]. However, the surplus of moisture/wound exudate in the wound bed can impede the healing process and damage the surrounding skin, which could lead to peri wound maceration[26,27]. Monitoring wound moisture (Fig. 1d) can shorten the healing time and the regularity of dressing changes, which in turn reduces nursing time, material and improves patient comfort[7,25]. Moisture measurements of wound helps to;

a. Adequate moisture balance for improved heling ;
b. Wound exudate volume tracking;
c. Timely dressing change.

Studies also suggest that the defined mechanical forces on the micro-environment of the wound site regulates the healing quality and speed. The mechano-transduction process enhances the proliferation and migration of fibroblasts and collagen synthesis. In vivo and in vitro studies shows that mechanical tension induce the fibroblast to myofibroblast differentiation[28,29]. As a novel mechanotherapy, negative-pressure wound therapy has shown importance in the treatment of wounds.

Bio-chemical and biological biomarkers. Biomolecular and cellular biomarkers are interlinked to wound healing stages and infection. These biomarkers include pH, multiple biomolecules contained in the wound exudate and bacterial presence/byproducts are mainly considered here.

The pH of a wound/wound exudate is influenced by microbial activity, enzymatic functions and cellular responses. pH is a wound status indicator belongs to biochemical class of biomarker. the normal skin pH is below 5 and that of a fresh wound is 7.4[30,31]. Studies have shown the use of pH as an evaluating parameter for the assessment of wound healing and the wound's pH was found to drop as wound healing progresses (Fig. 1e). The pH tracking of the wound can help envisage the evolution of healing as pH can be a suggestive pointer of the biochemical changes. Based on the scientific evidence, it is clear that pH serves as an important biomarker for monitoring chronic and acute wounds[32,33]. The pH in wounds has significance in:

a. Identifying the wound healing stage and progress[34];
b. Indicating the occurrence of infections. An abrupt increase of wound pH is an indicator for infection and high pH levels for an extended time indicates chronicity of wounds;
c. Indicating optimised wound healing conditions. This is related to the regulation of fibroblast and keratinocytes proliferation and controlling matrix metalloprotease (MMP) activities[31,35].

Wound exudate is a heterogeneous fluid with dissolved contents such as electrolytes, nutrients, inflammatory mediators, growth factors, uric acid and waste products[36]. Wound exudate also contains various types of cells like neutrophils, macrophages and platelets and it is undeniably a key biomarker for wound monitoring (Fig. 1f). This fluid can help to detect wound healing complications since the acute wound exudate is rich in leucocytes and nutrients, whereas chronic wound fluids have elevated levels of proteases, pro-inflammatory cytokines and higher level of Matrix metalloproteinases (MMPs), which diminish the healing progress[25,37]. MMPs are enzymes that are involved in, and promote, the healing process at multiple stages of healing. MMPs regulate immune cell influx, cleanse the wound from damaged extra cellular matrix (ECM) and tissues, enable migration of fibroblasts and keratinocytes, and remodel the scar tissue. Among these pleiotropic functions in the healing process, the timely expression, activation and suppression of the respective MMPs are vital for successful wound repairing. An imbalanced production of MMPs cause persistent degradation of the extracellular matrix and suppression of growth factors will lead to chronicity in wound healing[38,39]. Likewise, the elevated levels of uric acid in wound exudate can also act as an accurate marker of wound severity and the S. aureus or P. aeruginosa colonisation in chronic wounds significantly diminish the concentration of uric acid within wound exudate due to bacteria metabolization[40–43].

a. Identify the chronicity of wounds;
b. Provide the biochemical profile of the wound;
c. Wound healing progress;
d. Optimise wound dressing management.

Bioburden in wounds is the major cause to untimed inflammation, wound infection and chronicity. If the bacterial attack is not identified in time, the locally affected infections can grow and spread into the deep tissues and facilitate the potentially fatal systemic infection (Fig. 1g). The bacterial colony encased in an extracellular polymeric substance protects the microbes from immune attacks and become a drug barrier, called biofilm. The biofilms are capable of promoting anaerobic bacteria growth, synergism between different bacteria, generating MRSA-resistant proteins and it can inhibit cell proliferation, prevents cell migration and cause cell death in multiple ways. Zhao et al. studied the bacterial biofilm impact on diabetic mice with a control group. On the 4th week assessment, there was a huge delay observed in the wound healing with biofilms with 56% against 97% in control groups. It was also observed a 10-fold increase in interleukin (IL) 1β

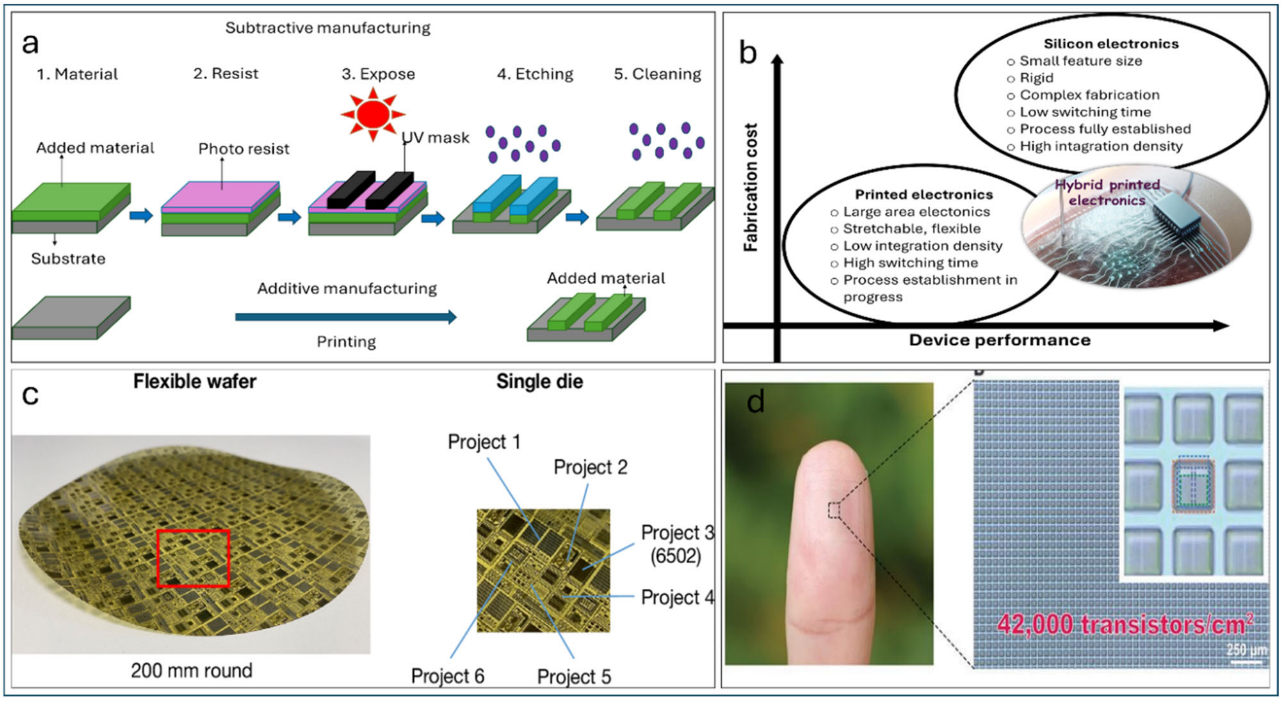


Fig. 2 | Hybrid printed electronics. Printed electronics vs silicon electronics fabrication strategy is depicted in (**a**), the necessity of hybrid printed electronics in their context of device performance and fabrication cost is shown in (**b**), The multi project approach of flexible chips production based on thin film transistor (**c**), reused from ref. 69. Elastic logic circuits with high integration density of transistors based on organic materials are shown in (**d**), reproduced with permission from ref. 70.

and IL 6 and MMP-10 expressions in the wounds with biofilms which is an indication of inflammation at the wound. The high-level expression of MMP-10 was indicating the collateral matrix damage resulting in delayed healing[44,45]. Timely infection diagnosis could help,

a. On time targeted medication;
b. To avoid becoming chronic wounds;
c. Accelerated healing.

There are some other parameters which are considered as important for optimising the wound healing conditions. Oxygen is one of such parameters and the disruption on the micro-macro vasculature may cause an oxygen deficiency case such as hypoxia that may occur in the wound. Hypoxia in wounds can impair healing; however, it can stimulate the vascular regeneration. Nevertheless, the severity and depth of chronic wounds can prevent adequate oxygenation for regeneration, causing wound ischaemia. Sufficient oxygenation is vitally important for the rebuilding of new vessels, connective tissue and also prevent infection[46,47].

The transduction of biomarkers/wound parameters into sensor signals is essential for effective wound monitoring, requiring careful selection of sensing principles based on electrochemical, bioelectrical, electromechanical, electrothermal and optical methods. Compatibility of sensors to interface with living tissue and the sensitivity of target biomarkers administer these choices. Wearable sensors must exhibit selectivity and specificity, emphasising the importance of choosing the right sensing principles. Creating wound monitoring sensors involves precise fabrication on wearable bandages, blending sensing principles into the design with appropriate fabrication strategies. This demands accuracy in sensing various biomarkers, with devices that are skin-compliant, biodegradable and suitable for mass production. Breakthroughs in fabrication techniques and material integration are essential for progress. In the development of smart wound dressings, printed electronics stand out owing to its versatility and cost-effectiveness in making flexible, stretchable and biodegradable electronics. The subsequent discussion will explore why printed electronics is pivotal in this field, covering topics including flexible substrates, biocompatible inks, printing methods and the importance of sensing principles. Lastly, we'll examine flexible sensors, bioresorbable electronics and multisensory patches with drug delivery capabilities for smart bandages.

Why printed electronics?

Conventional electronics is not much challenged or there is only little chance to get replaced by the alternative stream of electronics manufacturing such as printed electronics (PE). This is mainly due to its ultimate competence to go for nanometre scale feature sizes, reliability and high integration density. However, the last decade witnessed a novel niche of applications that demand large volume electronics, mechanically flexible, and lightweight at lower cost, which negate integration density and response times[48]. Large-area energy harvesters, energy storage, wearables and health patches, robotics computing and intelligent packaging are some of the recently derived nonconventional end applications[49,50]. As Fig. 2a depicts, printed electronics is an alternative fabrication route characterised by straightforward processing steps, not being dependent on expensive cleanroom and is feasible within ambient air, low temperature and non-vacuum processing. This approach is compatible with solution processable functional materials to fabricate stretchable, flexible and large-area electronics in a limited number of steps in an additive fashion[51,52]. The additive method of manufacturing printed electronics helps to reduce the material wastages, carbon footprint and complex processing steps with respect to the subtractive fabrication of conventional electronics, and dodges the usage of many chemicals such as etching agents[53,54]. Printed electronics has gained ground as an environmentally sustainable process with its excellent capacity to print on several eco-friendly substrates like paper, textiles and biodegradable polymers. However, printed logic devices and sensors are identified with relatively low switching speed, longer response time, inferior integration density and stability. Nevertheless, this is expected to be improved with the development of long term stable functional materials, maturation of the processing technology and improved packaging[55].

Smart wound dressings are an epitome of the resonance between the technology of printed electronics and the end-application, where the processing prompted downsides such as a sensor's long response time, low integration density and moderate lifetime do not affect its functionality. In the contrary, the preferred features of wearables such as the skin-like mechanical nature, stretchability and the integration of multi-functionalities can be well accomplished via PE fabrication[56]. Moreover, the low feature size in the submicron regime has been reported with advanced printing techniques, which may fuel the high integration density of printed electronics in future[57,58]. Yet there are challenges in wearable electronics, that could be addressed through hybrid technologies of printed electronics and silicon technologies that ensure better performance at lower cost as shown in Fig. 2b.

Hybrid printed electronics for wearables

The evolution of printed electronics led to the gradual transformation of rigid and rectangular shapes of conventional electronics to more soft, stretchable and conformal form factors. Scientists are extensively working on better interfacing such electronics with the human body meanwhile enhancing its functionalities. Wearable electronic patches embody a multi-functional platform, incorporating sensors, circuits, displays, batteries, antennas and logic units for data acquisition, processing and control. Many of such components have been reliably fabricated using printable, flexible, and stretchable materials, which allow adaptability in wearable technology. However, the development of high-performance IC chip remains a challenge for wearables. It typically requires materials and fabrication techniques that are not fully aligned with the stretchable and flexible nature of electronics.

Here, we briefly discuss the potential strategies for embedding high-performance logic circuits into wearable platforms. The integrated chips (ICs) are an essential part of all kinds of electronic applications and wearable electronics is not different from them. ICs are sets of electronic circuits on a small flat semiconductor substrate where a huge number of tiny metal-oxide field effect transistors are built in. The silicon ICs are extraordinary for their performance and available in size below 0.25 mm^2 for wearables[59]. Such chips are not limited to single use, instead, they can be reused multiple times, ensuring other aspects like economic viability and recyclability.

The hybridisation of printed and conventional electronics would be essential for efficient, low-cost, sustainable, and wearable electronics. The hybrid platform of printed electronics and silicon technology provides astonishing opportunities to produce high performance, consistent and conformable wearable electronics[52]. However, it is challenging to obtain flexible and compact integrated circuits with conventional Complementary Metal-Oxide-Semiconductor electronics owing to their thick and brittle nature. To meet the requirements of wearable electronics, there are relatively new strategies such as chip thinning through grinding the chip back, enabling the conformability[60]. The flexural rigidity of a material exhibits an inversely cubic relationship with its thickness. Consequently, silicon nanomembranes with a thickness of 10 nm possess a flexural rigidity that is over fifteen orders of magnitude lower compared to bulk silicon wafers of 1 mm thickness[61]. Prof. John Rogers and his team extensively worked towards flexible and stretchable silicon that consists of submicrometric single-crystal elements patterned into shapes with microscale, periodic, wavelike geometries[62]. The aforementioned research team also worked towards replicating silicon thin sheets (down to 300 nm), so that multiple such thin silicon nano-membranes can be obtained from the same wafer. Their work also demonstrated the electronic applications of silicon through the fabrication of a flexible solar cell and flexible N-type metal–oxide–semiconductor transistor arrays[63]. In a different note, Myny et al. have extensively worked on thin film transistors for flexible logic circuits[64–67]. In a perspective article, they emphasised the opportunities of thin film transistor technologies as a potential road map to the development of flexible chip for wearables[68]. Later his team demonstrated the feasibility of a multi-project wafer concept for the production of flexible (thin film transistor) TFT platform (6502 microprocessors) through independent foundries[69] (Fig. 2c). There have been research efforts towards the investigation of inherently flexibles stretchable materials exploited to the development of logic circuits. As shown in Fig. 2d, elastic logic circuits made out of polymeric materials with transistor channels length of 2 μm have been demonstrated and a 42000 transistors were integrated in a 1 cm^2 area[70]. On an application scale, the chips integrated on yarns using the E-Thread® technology for wearable RFID based sensor applications has been reported[59,71]. Such threads can be eventually integrated in a wound dressing for wireless wound monitoring.

Printed electronics for wound care

Printed electronics is an additive fabrication method where functional materials, in the form of inks, are deposited on flexible or non-flexible substrates. Printed electronics for wearable medical application encompasses numerous facets, including substrates, inks, the printing process and biocompatibility. These aspects are elaborated upon in the ensuing sections.

Substrates

The role of substrates is imperative in fabricating flexible and conformable sensor systems with wearable form factors for health monitoring. Just more than flexibility or conformability, substrate properties such as biocompatibility and air permeability are vital[72]. The dressing's no adhesion to the wound, and capacity to retain and absorb fluid are enviable. The other characteristics, like substrate transpicuous, that allows the inspection of wound healing, soft and gel-like interfaces for user comfort, and stretchability to support free movement are also highly desirable. Compatibility in terms of surface energy and surface roughness are indispensable for functionality fabrication on such substrates[8,73,74]. Textile substrates can be modified to smart textiles for wound monitoring applications. The concept of 'smart textiles' is realised through weaving, knitting, or sewing of functional fibres and threads or by printing formulations on textiles[75,76]. Smart textiles are intelligent systems that can both perceive and communicate the environmental conditions or the wearer's medical status. Textile substrates have unique advantages due to their compatibility with the human body, breathability, exudate absorbance and user convenience. Cotton gauze and other polymeric fibre textiles are common examples of textiles for wound dressings where the exudate management and permeability for oxygen/vapour are the major focus areas. A skin-like fabric for fluid management is developed that enables one-way liquid transport through distributed channels acting like sweating glands. The water transmission rate is 15 times greater than the best commercial breathable fabrics and it is achieved by creating gradient wettability channels across a superhydrophobic substrate[77]. Interactive textiles for wearables have been reported in physiological health monitoring, for example knitted/sewed, printed/coated sensors for biopotential measurements[76,78], sweat[79,80], pH[81] and temperature[82] monitoring.

Like textiles, paper substrates are made up of fibres of bio-origin such as trees, sugar cane and rags and have colossal research and commercial interest. The mesh network of cellulose in paper gives it a unique set of mechanical properties for fabricating wearable electronics[82,83]. Owing to its exclusive and dominant attributes, including natural abundance, low cost, lightweight, matured manufacturing process, recyclability, biocompatibility and biodegradability[84–86] makes it an excellent wearable platform. The feasibility of paper based point of care approaches to rapidly perform an initial screening of disease testing outside the laboratory conditions is prospective for remote diagnosis with no time delay. The abundant hydroxyl groups of cellulose fibres are beneficial for microfluidics where liquid transportation is driven by capillary forces which helps the effective distribution management of the least quantity test fluid into multiple sensing assays[87,88]. Paper based modalities consisting of distributed multi-analytical sensing assays, together with wicking fluidic channels incorporated devices are extensively investigated for cheap, disposable and diagnostics point of care applications[86,89–91]. Similarly, research on smart bandages has also applied paper as a suitable substrate for fabricating sensors for wound monitoring due to its omnipresence, capillarity, breathability, exudate absorption, non-toxicity and skin compliance features[42,92,93]. A chromatography paper bandage for Human Neutrophil Elastase assay[94], a screen printed electroanalytical paper bandage to detect infection[95] and multi-sensor printed smart bandages for comprehensive wound monitoring are reported in previous studies[47,96].

Hydrogel-based wound dressings are attractive due to a couple of reasons, including its skin-like mechanical features, ability to keep the wound moist and maintaining adhesion-free coverage on the sensitive underlying tissue. Stretchability and soft interface are two important features for hydrogels and viscoelastic properties are exceedingly desired for wearables. They can potentially intervene in wound healing by turning themselves intelligent[97]. Hydrogels consist of 3D networks of hydrophilic polymers crosslinked through physical and chemical bonds and these insoluble hydrophilic structures possess a remarkable ability to bind large volumes of water and present an excellent microenvironment for wound healing[98]. Incorporating antibacterial agents, biomacromolecules, stimuli-responsive nanoparticles and its timely activations are described. Tuned electroactivity and electromechanical properties of certain hydrogels have been applied in different sensors for wound monitoring and wearables. Biopotential measurement sensor systems applied conductive hydrogel-based electrodes for enhanced contact between the sensor and the body[99,100]. Hydrogel is transfigured into a mechanically compliable pH sensor, a highly stretchable strain sensor and a temperature sensor and tested for health monitoring[101–103]. The degradation of hydrogel properties, lower environmental stability and mechanical instabilities are the major downside of hydrogel-based sensors[104,105]. Novel progress in self-healing hydrogels are promising for wound care because of their capabilities to repair the structural damages and regain the original functions, analogous to the healing of living tissues[106,107].

Polyurethane (PU) materials are frequently used for wound dressings in different forms like PU foam, foil bandages and bandage tapes. PU bandages propose distinct advantages in biocompatibility, transparency and permit transmission of water vapour, O_2 and CO_2 from the wound[108–110]. PU also supports autolytic debridement of eschar and acts as impermeable to bacteria. Moreover, they are stretchable and well conformable to different shapes which does not entail additional tapping. Opsite™, Tegaderm™, Bioocclusive™ are some of the PU bandages available in the commercial domain[111]. An electrospun nanofibrous PU film consisting of PU fibres of a few hundred nanometres in diameter offer an ultra-fine porous structure. This has shown an increased re-epithelialization rate and the dermis was well organised with reference to commercial PU foils[112,113]. PU substrates are well suited for printed sensor fabrication due to its compatible surface properties and printability. A dual parameter temperature-pressure sensor[114], highly sensitive and stretchable strain sensor[115,116] and motion detector[117,118] are some of the sensor devices for which PU material was applied as a wearable platform.

Silicone is a biocompatible, clear, inert and non-flammable elastomer that has a spectrum of applications in human-machine interfaces, personal healthcare and sports performance monitoring[119,120]. Silicone is a low Young's modulus polymer that can stretch like rubber and is mouldable to specific shapes[120]. It is possible to realise precision microfluidic channels in silicone substrates for drug delivery and elastic liquid metal conductive traces for embedded electronics[121]. Silicone exhibits a minimal mechanical footprint and hold skin-analogous viscoelastic behaviour. Silicone has also been involved in the development of smart wound dressing integrated sensors and microchannels for drug delivery and oxygen supply[47,122,123]. Recent research on the topic shows enhancement of breathability and moisture permeability in polydimethylsiloxane (PDMS) patches achieved via engineering the micropores and water absorption and antimicrobial properties promoted through a hybrid PDMS-hydrogel for wound dressings[124–126].

Finally, polymeric foils such as polyethylene terephthalate, polyethylene naphthalate, parylene C and polyamide are flexible, biocompatible and have the potential to be deployed in wearable sensor fabrications[72]. Parylene is produced via chemical vapour deposition of the dimer of p-xylylene and possesses excellent properties (being biocompatible, stress free, conformal, chemically inert and optical transparent) required for flexible electronics[127]. Polyimide is a class of polymer foils with a stiff aromatic backbone structure that provides excellent mechanical, chemical and thermal (up to 400 °C) stability. An important aspect of printed electronics is the flexibility, majorly due to the use of polymer foils that limit the processing temperature for the sintering of functional layers. However, polyimide serves the purpose to a great extent. The major drawback arises from the weak adhesion between PI and the ink due to the lack of functional groups on its surface[128].

Different materials play unique roles in both wound monitoring and printed electronics, each with distinct advantages. Textiles, for instance, have been natural fit for human wearables. They are breathable, absorbent and can be enhanced with smart features, making them ideal for wound care. Textile fibre/thread especially impact in surgical wound monitoring through smart sutures. Through spatially selective treatment of textiles, one could easily manipulate the absorption/roughness nature of textiles which could enhance the printability properties. Paper-based platforms provide degradable, cost-effective printable substrates, making them ideal for disposable wound care electronics. Meanwhile, hydrogel-based systems mimic skin like properties, offering a moist healing environment, particularly beneficial for burn wounds. These materials can also be customised for wound dressings through on demand printing, integrating healing agents for personalised care. Silicone-based systems stand out for their flexibility, transparency and stretchability, making them easy to mould into wound shapes. Their fluidic channels further enhance drug delivery efficiency. Polyurethane (PU) substrates also offer transparency and a printable surface, rendering them as another great option for smart wound care applications. Many studies have shown multilayer wound dressings, combining different materials to optimise benefits—for instance, integrating silicone with a hydrogel layer and Polyamide with silicone[122,129]. These platforms all support printed electronics at various scales, paving the way for innovative, adaptable wound-monitoring solutions.

Functional inks

Printing inks are composed of pigments (the functional materials such as conductive, semi-conductive or insulating materials), resins, solvents, fillers and additives in the liquid or semisolid state, with defined flow properties. The functional materials in the form of flakes, nanoparticles, nanowires, or polymers are dispersed/dissolved/stabilised in a vehicle medium[130–132]. The ink flow characteristics such as surface tension, viscosity, thixotropy and viscoelastic properties are specific to each printing technology[133,134]. The tailormade process-specific flow properties are critically important for effective ink transfer or drop ejection without print defects[135]. The process-specific ink properties and flow attributes are achieved through manipulations and optimisations with the help of solvents, additives and resins where the same functional pigments can be contained in different classes of printing inks (Fig. 3R1). For example, screen printing requires a high viscous thixotropic behaviour of inks (paste), whereas inkjet printing demands low viscous inks that need to meet the jetting window parameters[136,137]. Nanoparticle-based inks often encounter agglomeration or sedimentation[138]. The inks based on nanoparticles are stabilised in formulations by organic ligand shells, i.e. the nanoparticles are encapsulated with an organic material, known as a capping agent, to form a consistent and stable dispersion[139–141]. This capping agent needs to be removed after printing through curing or sintering to allow physical contact between nanoparticles, forming continuous connectivity.

Conductive inks. Conductive inks are mainly made up of silver, carbon, copper and conductive polymers. Among them, silver is ubiquitous in printed electronics due to its superior conductivity (~10^6 S/m), where the oxidative form maintains its conductivity and the material is available at moderate prices (in comparison to platinum or gold). High curing temperatures of the inks were a challenge in the inception of printed flexible electronics. The metal particles exhibit a drastic decrease in their bulk melting temperature when downsized to nanoscale and this property has been widely exploited for low curing temperature inks (fig. 3R2). The sintering temperature of the nanoparticle ink (T) is related to the bulk metal melting temperature (T_0), surface energy dependent parameter of the nanoparticle (σ) and radius of the nanoparticle (r) through

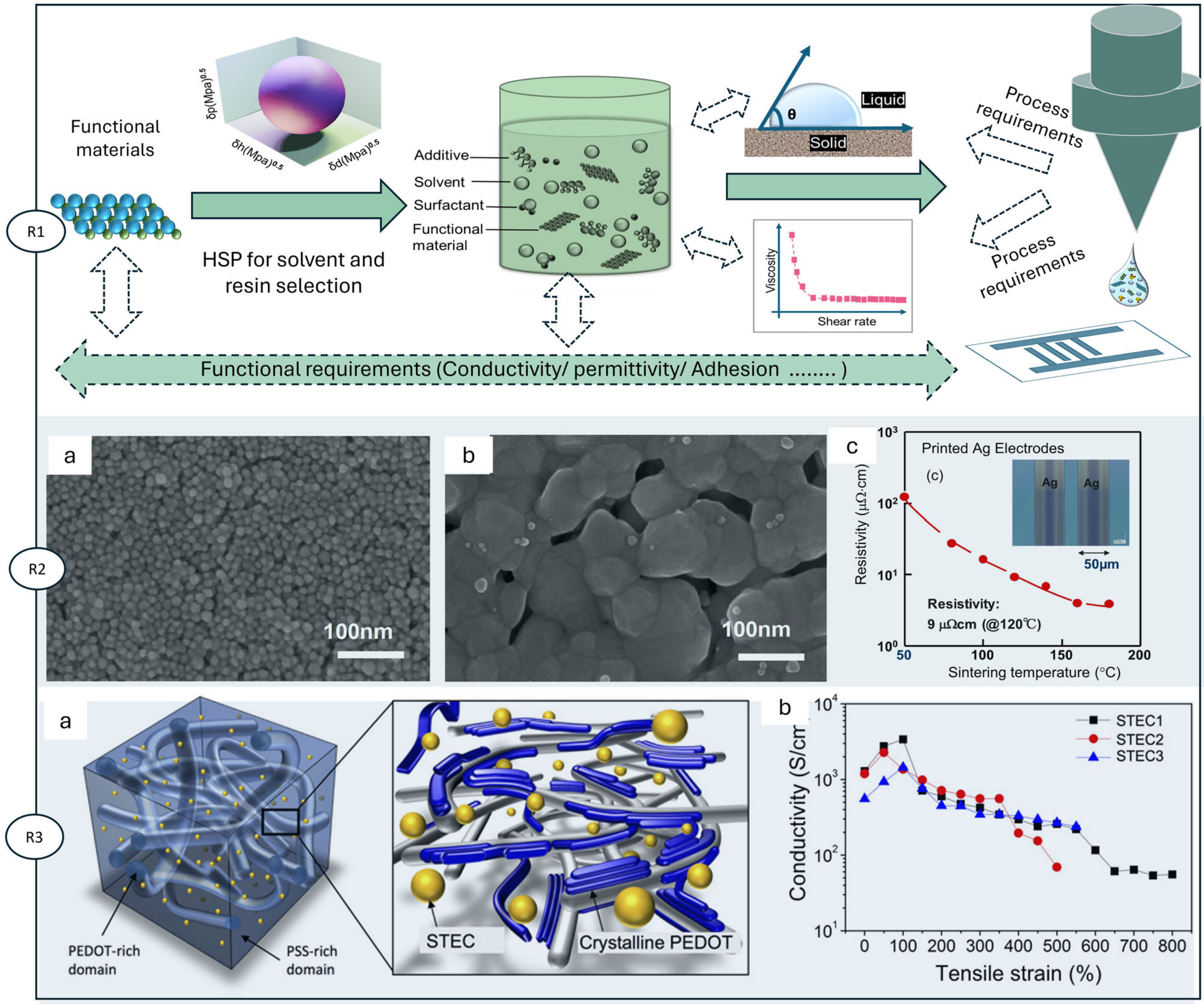


Fig. 3 | Functional ink formulation steps, composition and functional layer properties. R1 Schematic of functional nanomaterial-based ink formulation. The solubility/dispersibility of the functional materials needs suitable solvents and it can be identified and optimised with the aid of Hansen solubility parameter (HSP) model. The process specific properties are achieved with flow property modifications like surface tension - wettability characterisation and viscosity-shear thinning test. **R2** shows Low temperature nanoparticle sintering of silver nanoparticle inks and its impact of temperature on sintering[222]. Before sintering (**a**), after sintering at 100 °C (**b**) and resistivity of printed silver electrode over sintering temperature is shown (**c**). **R3** shows the improved stretchability and conductivity of PEDOT: PSS material. Morphology of a stretchable PEDOT film with ionic additives (**a**), the plot implies the improved stretchability and conductivity of it (**b**), image is reproduced from the ref. 175.

the relation,[142],

$$T = T_0(1 - \sigma/r) \quad (1)$$

Silver takes different forms in silver inks such as nanoparticles, silver flakes or silver nanowires of which the silver nanoparticles offer greater advantages to reduce the sintering temperature. Mo et al. studied the relationship of nanoparticle size, curing temperature and conductivity and have shown the ink curing below 100 °C with compromised conductivity[143]. Rosker et al. have shown that the aerosol jet printed silver can be obtained near bulk conductivity in low temperature conditions (80 °C). The work used reactive silver inks, i.e. a metal-organic decomposition (MOD) reaction converts a liquid precursor material into pure metal. The improved conductivity achieved through a path of densification and grain refinement (grains much larger than the mean free path) helped to approach the practical conductivity limits[144]. MOD inks benefit further in terms of less nozzle clogging and low temperature curing. Meantime, silver nanowire formulations offer good trade-off between sheet resistance (1–10 Ω) and transparency (70–90% at 550 nm) in low temperature (below 100 °C) processing[145,146]. Printed silver is biocompatible to the human body and it can be an antimicrobial agent, which opens a large number of opportunities for wearable electronics[147].

Copper based inks are very appealing and broadly investigated due to their good conductivity ~10^5 S/cm[148], low price and anti-microbial nature. Nevertheless, they have certain disadvantages like their instability against oxidation under ambient conditions to form insulating layers of copper oxide[149,150]. However recently published research outcomes show that the oxidation issues of copper inks while curing has been greatly reduced. The metal organic decomposition precursor inks provide solutions for printing copper inks on foils without significant oxidation and low temperature curing feasibility[151]. Rosen et al. reported a self-reducing copper precursor based on copper formate for the oxidation-stable, low temperature cured conductive structures of comparable bulk copper conductivity[152,153]. There are many approaches (capping agents for nanoparticles, formic acid environment for curing, photonic sintering etc) studied to reduce the copper oxidation[154].

Similarly, carbon-based inks composed of carbon black, graphite or carbon nanotubes (CNTs) are also applied in printed devices. Although its conductivity ~10^4 S/cm is one or two order less than silver, they are abundantly available, environmentally friendly, disposable and cheap[155,156]. Screen printed carbon-based electrodes are massively used for biosensors research due to their porous nature, ease of modification to design practical, low-cost and disposable devices[157]. More specifically, carbon-based materials can adsorb and store ions owing to its excellent pore structure and large specific surface area, contributing to outstanding electric double-layer capacitance performance. Carbon/carbon nanotube inks have been applied for wearable applications such as pH sensors, temperature sensors, glucose sensors and strain sensors[158–160]. Printed polymer-carbon nanotube composite inks fit to the stipulations of superior stretchability (above 50% strain) and a high gauge factor (above 10) for wearable strain sensors[161,162]. Previous studies show cell viability on carbon electrodes, proving its non-cytotoxicity and suitability of the carbon pastes as biocompatible electrode material[81,160,163,164].

Graphene is a nanomaterial of one atom thick layer of carbon atoms organised in a 2D hexagonal lattice with thermal, electrical, chemical, mechanical and optical properties that are desirable for many applications. Lin et al. made a serendipitous discovery that porous graphene can be induced on the polyimide foil with laser scribing at low power settings[165]. This laser source creates photochemical and photothermal changes on the substrate leading to the formation of 3D porous graphene. The reported LIG has a surface area of $428\,m^2\,g^{-1}$ and resistance of ~10 Ω/□ which are comparable to the graphene produced through conventional methods[166,167]. As a facile approach to get graphene on flexible substrates, it can be an alternative for low cost, eco-friendly and biocompatible conductor choice towards wearable printed electronics[165,168]. Likewise, MXenes, a versatile family of 2D transition metal carbides, nitrides and carbonitrides, have emerged as a interesting material since their discovery in 2011. Their remarkable mechanical and electrical properties, combined with printability, make them ideal for wearables and biosensing applications. Unique features such as high surface area, enhanced hydrophilicity, and rich surface chemistry enable functionalization and precise property tuning promotes MXene-based electronics (MXetronics)[169–171].

Poly(3,4-ethylenedioxythiophene) polystyrene sulphonate (PEDOT:PSS) is a promising conductive polymer for biocompatible electronics[172,173]. PEDOT:PSS, with its high transparency (~90%) and conductivity (~1700 S/cm), is ideal for diverse applications including solar cells, OLEDs, sensors and batteries[174]. Wang et al. reported enhanced conductivity, stretchability of PEDOT:PSS through ionic additives where it helps the modification of the layer morphology and simultaneously act as a dopant. As plotted in Fig. 3R3, through an interesting PEDOT:PSS formulation with ionic liquids, the transparent film has shown 3100 S/cm under 0% strain and improved conductivity of 4100 S/cm under 100% strain[175]. PEDOT:PSS can function as an electrode, charge-selective layer, stretchable conductor, biosensor, heating element, thermoelectric material and piezoresistive element in various devices[176–179].

Semiconductive inks. Semiconductors serve as the foundation for a wide array of modern electronic components, including transistors, diodes, solar cells and sensors. Semiconducting materials for printed electronics can be classified into three broad categories: metal oxides, organic materials and carbon-nanotube semiconductors.

Metal oxide (MOX) semiconductor materials are particularly prevalent in wearable electronics, especially for biochemical sensors, due to their high sensitivity to electrical resistance, large surface area, non-toxic nature and chemical stability when exposed to various analytes[180–182]. MOX semiconductors are characterised by valence compounds with strong ionic bonding. The metal nS orbitals exhibit high dispersion, while the oxide 2 P orbitals are localised, resulting in a smaller effective mass for electrons compared to holes, thereby enhancing electron transport properties[183]. These exceptional electronic properties make MOx materials suitable for pH, gas and humidity sensing applications[182,184–187], as well as for use in charge transport layers and transparent electrodes in solar cells and light-emitting diodes[188–190].

Organic semiconductors and polymers are great choices for wearable electronics due to their mechanical flexibility, affordability, solution processability and biocompatibility[191]. The topical studies shows diverse soluble organic semiconducting materials have been engineered, exhibiting outstanding field-effect mobilities surpassing $10\,cm^2V^{-1}\,s^{[-1}$[192]. Conjugated polymers, a major class of semiconductive polymers, consist of organic molecules with a backbone chain of alternating carbon-carbon single and double bonds. They possess HOMO—highest occupied molecular orbital and LUMO—lowest unoccupied molecular orbital. The energy gap between the HOMO and the LUMO determines the material's band gap energy, thereby impacts its electrical properties. These polymers exhibit both electronic and ionic conduction, with electronic conduction stemming from highly delocalised molecular orbitals and overlapping molecular orbitals[193,194]. The amalgamation of electronics with living tissue holds great potential for a variety of life-enhancing technologies. Due to their mixed electronic and ionic conductivity, these polymers have proven to be effective transducers of biological activity, with applications ranging from neural interfacing to biosensing[193,195,196]. Meng et al. introduced a multifunctional smart theragnostic bandage deploying conducting polymers capitalising on their unique intrinsic properties such as redox reversibility, electrical conductivity and volume enlargement. This innovative bandage seamlessly integrates multi-sensing elements for biosensing and on-demand antibiotic release[197].

Modifying the conductivity of conjugated conductive polymers for real-world applications typically involves doping with respective donor/acceptor materials[198]. Semiconductive polymers find widespread use in wearable electronics, particularly in photodiodes, energy harvesters, pH sensors, gas sensors, and temperature and strain sensing[198–200]. Poly(3-hexylthiophene) (P3HT), pentacene, polyaniline (PANI) and polypyrole are commonly used organic semiconducting polymers in printed electronics. P3HT is a well-studied and acclaimed for its solution processability, and stability, suitable for applications such as organic solar cells, flexible transistors, thermoelectric sensors, gas sensors and photodiodes[201–204]. P3HT's versatility as a biocompatible polymer allows it to fulfil various functions in flexible electronics, thanks to its ability to be synthesised with precise molecular weights, regioregularity and end-group functionalities[205,206]. Pentacene, a printable p-type organic semiconductor, enables the fabrication of high-performance thin-film transistors characterised by low operating voltages and high field-effect mobilities[207,208]. Solution-processed pentacene-based transistors have demonstrated comparable performance to those fabricated in cleanroom conditions, highlighting their biocompatibility by supporting cell growth[207]. PANI has been an intrinsically conductive polymer with good environmental stability and biocompatibility. PANI based sensors have demonstrated as excellent choice for electrochemical sensing especially for wound pH sensor applications. The sensing capabilities of PANI is due to its multiple redox states and reversible interconversions of emeraldine base to emeraldine salt[209]. Focusing on the sustainability of organic electronic materials, Park et al. demonstrated closed-loop recycling approaches to recover and reuse of organic electronic functional materials from wearable devices. Their approach employs environmentally friendly solvents, including water and anisole. The reliability of these recycled materials was assessed by fabricating sustainable wearable devices, enabling multiple reuse cycles of reconstructed organic electronic components without the need for additional replenishment[210].

Dielectric/insulating inks. Insulating and dielectric material properties are highly pertinent in thin-film micro- and nanodevices, whereas the miniaturisation/performance of electronics are limited by it. Dielectric inorganic thin films such as alumina (Al_2O_3) and silica (SiO_2) demand high temperature/vacuum processing to deposit on substrates which are not ideal for flexible substrates. Materials such as Cytop^{Tm}, polymethyl methacrylate (PMMA), polymer poly(vinylidene fluoride) (PVDF), silicone, PU and polyvinylpyrrolidone are applied as insulating coatings[211–214]. The above-mentioned polymer materials are biocompatible, function as ink matrix for other functional materials and are often used as a substrate/skin interface layer for wearable electronics[211,214,215]. In specific printed devices, specifically in the pursuit of enhancing device performance, increasing the relative permittivity of dielectric materials has become paramount. To achieve this, improving the dielectric strength of the polymer ink composition is essential. Incorporating high permittivity ceramic compounds like barium titanate, MOXs, or silicon dioxide into the polymer matrices alongside ink additives is a common approach for presenting dielectric inks[216–219]. In an interesting development, printed thin layers of hexagonal boron-nitride (h-BN)/PU layers are deposited on flexible substrates which possess high relative permittivity ($\varepsilon r = 7.57$) and transparency (≈78.0%)[220]. Carey et al. demonstrated fully inkjet-printed 2D-material active heterostructures with graphene and h-BN inks, to fabricate flexible and washable FETs on textile, achieving a field-effect mobility of ~$91\,cm^2\,V^{-1}\,s^{-1}$, at low voltage (<5 V)[221]. Electrical requirements for the dielectric materials are high breakdown strength and low leakage current in the thin layers and optimal thickness is a trade-off to reduce the operating voltage and to avoid electrical shorting[222].

Bio-sourced inks. Nature-derived functional materials possess blends of several properties including bioactivity, optical, electromechanical along with skin-like features, either by themselves, or as a composite. Such nature-derived or -inspired materials-based inks and functional formulations are increasingly relevant for the wearables owing to their biocompatibility, biodegradability, abundance, economic viability besides due to the e-waste posed threats. A natural polymer from insects, silk fibroin has been deployed for bacterial detection in wound dressing and tissue engineering scaffolds[223,224]. Silk polymorphism and biomaterial composites in silk fibroin inks can adjust its biochemical properties. Incorporating biomolecules like growth factors and enzymes into silk inks enables the study of cell-material interactions. A demonstration involved printing functional polydiacetylene –antibody–silk inks on laboratory gloves, showing colour change in the pattern upon exposed to high E. coli levels[223]. In a recent work graphene incorporated with silk fibroin and this composite ink has been printed to self-healing electronic tattoos that can be seamlessly mounted onto the skin. These tattoos demonstrate sensitivity to fluctuations in strain, temperature and humidity[225]. Another distinct work described about an electronic-protonic material which is rich in eumelanin released by cuttle fish, utilised as the precursor to synthesise carbon quantum dots as a fluorescent sensor for highly sensitive and selective detection of organic compounds[226]. In a recent study, egg albumin has been formulated to fabricate the active layer for biocompatible humidity sensors which has recording in the range between 0% and 100% RH[227]. Bioresorbable devices are of great interest and beta carotene is a red-orange pigment found in plants and fruits, that is used as inkjet printable semiconducting layer ($6 \times 10^{-4}\,cm^2\,V^{-1}$) for the transistor fabrication[228].

Keratin is a structural protein that is naturally present in different forms (hair, nail). It has been used as an active material for humidity sensing, transient resistive switching, random access memory device films and diodes. Resins are indispensable part of printing inks and shellac is natural resin that acts as a renewable, biodegradable and water insoluble. Shellac can be a dispersing agent for conductive particles and can be a modifier for functional surfaces[229,230]. As a noteworthy approach on sustainability, wood lignin has been converted to graphene and it opens the door to many interesting sustainable electronics applications[167,231]. In their research on the lignin converted graphene, Edberg et al. show forest wood based ink screen printed onto the desired substrates and its conversion to conductive graphene by laser induction[232].

The most abundant organic polymer, cellulose (and its derivatives) are incorporated with other functional materials, through doping, blending or coating. It exhibits the potential for new composite materials of printed electronics[233]. Nanocellulose, mainly cellulose nanofibrils and the cellulose nanocrystals are especially interesting in functional bioink formulations due to their similarity to extracellular matrices and their excellent biocompatibility. Moreover, their high surface area, dispersibility, the feasibility of forming stable metallic particle suspensions and the ability to form self-standing films are beneficial for functional applications[234,235]. Collagen and gelatine are other common polymers, often combined with nanocellulose to form more biocompatible 3D culture platforms[235].

Chitosan is an another bioactive linear polysaccharide polymer derived from the shells of crabs, lobsters and shrimps. It has been significantly applied for wound dressings and other biomedical applications due to their antibacterial activity, nontoxicity and ease of modification[236]. Chitosan is a biodegradable (degraded by several enzymes, mainly by lysozyme, a generalised protease found in all mammalian tissues[237]) FDA approved polymer material for drug delivery, wound dressing materials and biosensor applications[238]. It has been specially interesting for wound healing due to its haemostatic properties[239]. Chitosan has been formulated to printable form and used in sensing (humidity, gas and biorecognition) applications[236,240,241]. Long et al. developed 3D printed Chitosan-Pectin hydrogel incorporating the local anaesthetic drug lidocaine hydrochloride as a potential wound dressing[242]. Similarly, Heidenreich et al. formulated and optimised a bio-ink consisting of collagen and chitosan blends to print tissue scaffolds[243].

Liu et al. worked towards a bio-sourced, biodegradable, biocompatible and elastic polymer developed for medical applications. Their phosphor-ester cross-linked vegetable oils were prepared by cross-linking reactions between phosphorylated castor oil and epoxidized vegetable oils, epoxidized soybean oil and epoxidized linseed oil, without any solvent or initiator, at 37 °C. The implantation test for the elastomer polymers in animal wounds shows that the elastomer can be absorbed completely within 3 months[244].

Biocompatible printed electronics

The biocompatibility of printed wearable electronics depends on their interaction with the human body from both physical and biological perspectives[245,246]. The electronics development for wounds must be analogous to human tissues and non-toxic at the cellular level interactions. Physical incompatibility can lead to tissue damages and device failure and the imperfect contact with the tissues obstructs the monitoring functions. These devices are requisite to conform to the body's curvilinear surfaces, requiring materials that mimic viscoelastic skin behaviour. Achieving physical compatibility involves the selection and manipulation of materials that exhibit intrinsic or extrinsic flexibility and stretchability. Materials like conductive polymers are frequently employed in sensors and devices due to their intrinsic properties. Despite their exceptional flexibility, such materials exhibit relatively weak electrical properties compared to traditional electrode materials. Incorporating functional materials such as silver nanoparticles, CNTs and graphene into a suitable matrix (e.g. PU or Silicones) can tailor the Young's modulus of composites, enhancing stretchability and flexibility with excellent electrical properties. Hazendonk et al. have shown a printable composite ink based on TPU and graphene for screen printing of nontoxic stretchable conformable conductors applicable for different wearable applications[247]. Similarly, hydrogels modified at the molecular level for bioelectronic properties offer another example of achieving desired physical characteristics along with the electronic functionalities[248,249]. In a different approach, attributes of conventional materials have been enhanced using nanometre-scale thickness deposition and specific geometric patterns, like serpentine or wave structures, to improve flexibility and stretchability[250]. Chang et al. shown the fractal designs combined with carbon nanotube-silicone ink to print electrodes with tensile insensitivity and softness for wearable sensors[251].

The biological incompatibility of electronics can result in skin contamination, cytotoxicity, irritation and infection. The biological acceptability of wearable electronics can be achieved from the perspectives of materials, design and packaging. Bhushan et al. studied the silver and carbon printed layer biocompatibility evaluating ionic leaching induced cellular inflammation. In case of carbon leaching, the impact was observed to be minimal causing negligible cytotoxic response meanwhile the silver ions

Table 1 | Printing process and their comparison[52,56,80,133,274,285,435–437]

Printing techniques	Printing form	Printing speed (m/min)	Linewidth (µm)	Thickness (µm)	Ink viscosity (cP)	Surface tension (mN/M)	Printing on 3D shapes
Offset lithography	Planographic	200–500	10–50	0.5–2	20,000–100,000	--	No
Flexography	Raised	8–180	20–100	0.1–2.5	50–500	13.9–23	yes
Gravure	Sunken	5–600	2–50	0.01–5	50–200	41–44	no
Screen	Stencil	0.6–100	19–50	1–20	500–50,000	38–47	Yes
Inkjet	Digital	0.02–5	20	0.5–15	20	15–25	No
Slot die coating	Slot die lip gap	1–600		0.005–100	1–10,000	25–50	No
Spray coating	Sheath gas	6–60	500	0.005–10	1–30	--	Yes
Reverse offset	Printing plate	0.0001–0.1	0.6	0.05–1	40–100	--	No
Aerosol jet	Digital		5	0.1–5	1–2500	<30	Yes
Electrohydrodynamic	Digital	0.06	(<1 µm)	0.001–0.1	1–10,000		yes

caused substantial decrease in cell viability[252]. Selecting non-allergenic, non-inflammatory and non-toxic printable materials, such as CNTs[253], gold[254], silicone, P3HT[255], magnesium[256], PANI[81], PEDOT:PSS[257], PVDF[258] and many other of this material class ensures biological compatibility at the wound-electronics interface. The design of wearable multilayer functionalities can be innovative for achieving biocompatibility. Dai et al. report on body neuromorphic computing, based on electrochemical sensors, utilising a silicon layer as both the bottom and top layer ensuring skin compatibility[259]. The upper layers, which do not require direct contact, can thus be made from a broader range of materials and such smart design approach helps overcome biocompatibility challenges. In an important research work on wound monitoring patch development, a semipermeable porous PU membrane has been utilised as analyte diffusion limiting layer interface between the wound and sensor ensuring the biocompatibility of the patch[260]. Alternatively, compatible thin layer packaging/encapsulation of wearable electronics can provide a way out to biocompatibility issues. By ensuring that the packaging itself is biocompatible (like a silicone coating), the risks of adverse reactions can be significantly reduced, enhancing the safety and effectiveness of these innovative devices[261,262].

Printing systems

Printing processes exert a profound influence on device performance, with morphology, thickness and resolution being the pivotal factors. The chosen printing technique profiles the morphology and uniformity of the layer, impacting crucial aspects such as semiconductor mobility and intermolecular interactions (in organic solar cells, organic thin-film transistors (OTFTS) etc[263–265]. Researchers are actively exploring novel printing/coating methods to achieve crystalline semiconducting films, thereby enhancing mobility and better charge transport[265–268]. Diao et al. assessed diverse solution-based thin film deposition methods and their impact on device performance. Utilising a micropillar-patterned blade, they sheared a solution of TIPS-pentacene, yielding aligned single-crystalline domains. TFTs fabricated through this method showcased fourfold greater mobility than spin-coated counterparts[267]. The thickness and uniformity of active or charge injection/transport layers have detrimental impact on devices performance matrices[269,270]. Kumar et al. reported a novel hybrid printing process for the deposition of a uniform layer deposition of functional material. The process comprise of finite micro-droplet generation of ultrasonic spray coating combined with the screen printing mesh enabling a nature inspired deposition technique for ultra-thin functional layers. The innovative process enables even 5 nm thin uniform layers[271]. For instance, improved semiconductor layer morphology bolsters charge carrier mobility in OTFTs, while high-resolution printing enables shorter channel lengths, leading to better frequency response and higher integration density. Furthermore, the uniformity and well-defined thickness of the gate dielectric layer facilitate low-voltage operation. The impact of printing processes extends beyond morphology and thickness, encompassing crucial aspects like integration density, device miniaturisation, imperceptibility and production volume. Different printing process characteristics are compared in Table 1.

High volume printing process. High-volume printing methods like screen and gravure printing process are the key for scale up of printed electronics. Leveraging these established processes in print industry could help to smooth out the transition from lab to production of smart wound dressing. With rising demand for affordable health monitoring patches, these techniques play a major role to efficiently meet large-scale production needs.

Screen Printing: Printed electronics is in the lab to fab transition era where screen printing is the workhorse technology. Screen printing (Fig. 4a) is a facile printing technique which does not even need any expensive machineries (obviously for limited volume) and printing can be fulfilled in complete manual processing without significant mismatch in print quality[272]. The pioneer printing scientist Steven Abbott suggests a simple yet golden rule for quality screen printing, that is 'Let the mesh do the metering, let the stencil do the shaping and let the ink do its thing'[272]. This printing technique is an ancient method categorised as a thick film material deposition method using high viscous (>500 cp) and thixotropic inks/pastes[133,273]. Although the technique was never considered as a fine line printing technique, recent studies of the screen printing has been shown potential to be classified as high resolution printing (120–100 µm in 2005 to 19 µm in 2020)[274]. Screen printing is especially interesting due to its simple processing, roll-to-roll feasibility, low cost and compatibility for various substrates including rough textiles as it can deposit thick films[275–277].

Gravure printing: Gravure printing (Fig. 4b) is an intaglio process where the image area is sunken, and the dots are etched on the cylinder surface. In this method, the image-carrying cylinder is partially submerged in a low viscous ink, and a doctor blade system swipes away the ink from the cylinder surface, leaving the image area consisting of cells to carry the ink and transfer it to the substrate at the printing nip. The ink transfer is taking place due to the capillary forces acting on the ink in the cells at the printing nip[278,279]. In optimised production conditions, researchers from UC Berkeley reported a fully gravure printed polymer transistor on foil with a feature size of ~5 µm[280]. Gravure printing processes offer design feasibility for a seamless repeating pattern and excellent layer uniformity over long run roll-to-roll printing[279,281], emphasising its importance for commercial scale production of wearable electronics. Researchers demonstrated gravure-printed biosensors with electrochemical stability and uniform redox kinetics on 150-metre-long flexible rolls for wearableapplications[282].

Large area, uniform thin coating. Spin coating is widely used for depositing uniform thin layers in small-area electronics due to its simple operation and ability to control thickness. However, it is unsuitable for large-scale production due to material waste and limited scalability. Large

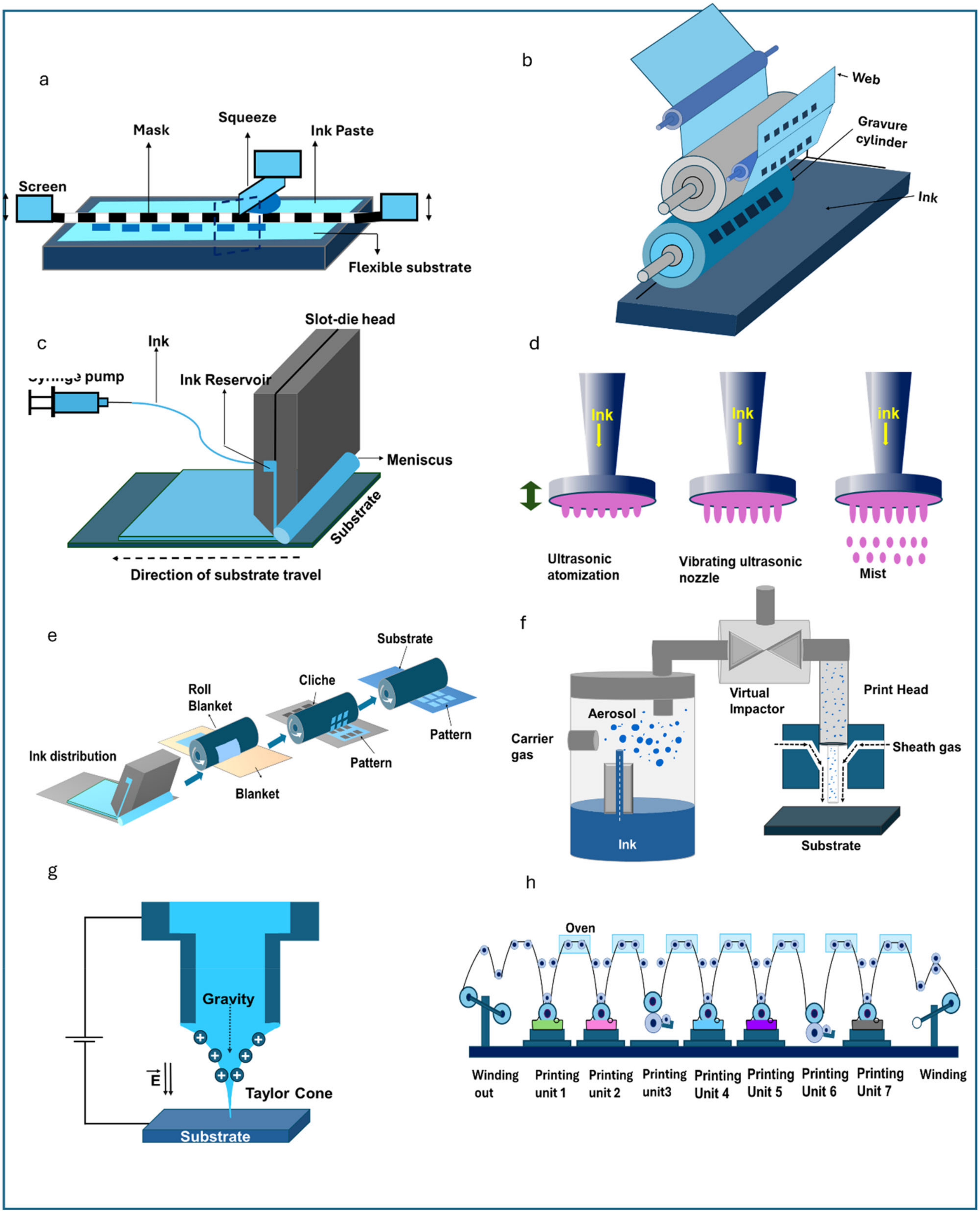


Fig. 4 | Schematic illustrations of printing process. Large volume conventional printing techniques for printed electronics, screen printing (**a**), gravure printing (**b**). Uniform thin layer coating techniques for printed electronics, slot die coating (**c**) and spray coating (**d**). Low feature size printing methods, reverse offset printing (**e**), aerosol jet printing (**f**), and EHD printing (**g**). The illustration of a roll-to-roll hybrid printing line where the multiple printing stations (with interchangeable printing and coating technologies) can be arranged according to the device layer requirements (**h**).

area coating techniques like slot-die and ultrasonic spray coating address the limitations of spin coating, paving the way for cost-effective, large-scale thin-layer deposition.

Slot-die coating: Slot-die coating is a forefront technology for uniform film deposition of liquid chemistries at low operational cost and minimised material wastage. The coating system consists of a pre-metered coating solution introduced through the dies gap to the substrate (Fig. 4c). The reservoir cavity in the dies helps to achieve uniform distribution of the solution throughout the web width. The solution in the coating gap (the gap between dies and substrate), bounded due to the upstream and downstream meniscus, leads to the formation of the coating bead. The stability of the coating bead is highly desirable for the steady deposition of the material. Ideally, a stable coating window can be mapped for slot-die coating where a consistent and uniform defect less coating is possible.

Ultrasonic spray coating: Ultrasonic Spray coating (Fig. 4d) is important due to its ability to functionalise any surface geometry with extremely thin, uniform, conformal coating. The coating head has a piezoelectric transducer which creates a mechanical vibration on the coating head at a certain ultrasonic frequency leading to atomisation of the ink resulting in micrometre-sized droplets. The amplitude of the standing waves reach a critical point where the capillary waves of the atomised fluid break up, attributed to Faraday instabilities[283], leading to the formation of a mist of tiny droplets. The uniform size of the tiny drops, multi-pass coating capabilities, low clogging possibilities enable a lower roughness profile of thin and thick layer deposition of functional materials over extended run time. Extremely thin and uniform functional layer deposition down to 5–20 nm is compatible with USSC[271,284,285].

Ultrahigh resolution printing. High-resolution printing with line widths of 10 µm or less enables the miniaturisation and high integration of electronic components, paving the way for imperceptible and densely integrated complex functions such as logic circuit elements[192,286]. Conventional printing methods typically produce electrodes with widths of 100–200 µm and rounded shapes at ends and corners, leading to larger channel widths of 1000 µm or more in a thin film printed transistor devices. Therefore, achieving electrode miniaturisation and precise shaping is essential for reducing the size of TFT devices and improving key performance metrics such as cutoff frequency[222]. Various high-resolution printing processes are emerging, including reverse offset, electrohydrodynamic printing and aerosol jet printing.

Reverse offset printing: Reverse offset printing (Fig. 4e) has gained significant attention for electronic component fabrication. It offers sub-micrometric printing resolution, precise overlay and sharp-edged features comparable to photolithography[222,286,287]. The process involves applying a thin layer of ink to a blanket and using an intaglio-patterned cliche for patterning. The process window is influenced by a complex set of factors, with a particular focus on the strength of adhesion between the blanket and the ink, the adhesion strength of the ink cliché and the cohesion strength of the ink. As the linewidth decreases the printing window also diminishes[288].

Electrohydrodynamic printing: Electrohydrodynamic printing (Fig. 4f) or superfine inkjet printing, has also attracted the research interest due to its ability to print fine structures in the few 100 s of nanometre range with high aspect ratios[289,290]. It operates in a contactless mode, during this electric influence, ions in a polarisable liquid assemble at the liquid's surface, leading to the formation of a conical Taylor cone at the nozzle tip due to ion repulsion. When the electric potential surpasses a specific threshold, the charge repulsion at the cone's apex overcomes surface tension, causing the emission of a fluid droplet toward the grounded substrate and form the print[291,292].

Aerosol jet printing: Aerosol jet printing (Fig. 4g) is a non-contact direct-writing technology that deposits various functional materials on 2D and 3D surfaces. This versatile digital printing technique offers 5-axis printing capabilities, handles a wide range of ink viscosities and achieves feature sizes as small as 10 µm. The system atomises inks using ultrasonic (for low viscous) or pneumatic (for high viscosity) methods and in the nozzle, a sheath gas stream is wrapped around the atomiser flow, which results in a focusing effect and avoids the potential clogs in the print head[293]. These advanced printing techniques hold great potential for the miniaturisation and high-density integration of electronic components. Compatibility with 3D shapes and consistent printability for long runs makes the process impactful for many applications like chip interconnections.

Despite remarkable high-resolution patterning capabilities, particularly below 10 µm, the aforementioned printing techniques necessitate a cleanroom environment and optimal ink characteristics. The pivotal question that emerges when transitioning high-resolution printing into manufacturing is yield, consistency and variability. As feature size diminishes, the printing window constricts, rendering the process increasingly susceptible to environmental factors. The feature sizes and uniformity of thin layer functional materials are keeping its challenge especially in long run jobs. Though Kim et al. presented a modified flexographic printing method using nanoporous polymer-coated aligned CNTs as the image carrier, achieving a combination of low feature size (3 µm) and high throughput (12 m/min). The printed structures exhibit highly uniform nanoscale thickness (5–50 nm) and high fidelity (edge roughness ~0.2 µm)[294]. Moreover, the intervention of machine learning model assisted data-driven optimisation and the adoption of hybrid printing and coating (Fig. 4h) based fabrication would enhance print quality and consistency in long run jobs. A recent research introduced micro factory, a self-driving digital twin that transforms roll-to-roll printed electronics through high-throughput, closed-loop optimisation. The developed platform combines printing-inspired automation with machine learning models to improve device up scalability and performance[295].

Sensing and actuation

The sensing principles are the fundamental of health monitoring, translating vital biomarkers into sensor signals. They transduce the biomarkers into meaningful sensor output, shaping sensor characteristics. Unlike conventional sensors in a controlled environment, wound monitoring sensors are in a heterogeneous environment, demanding enhanced sensitivity and selectivity towards biomarkers, making the choice of sensing principle pertinent. Moreover, the sensing principle must be meticulously chosen for reliable, continuous operation and reproducible measurements. Wearable sensor platforms, in close proximity to the body, necessitate stretchable, flexible operation and resistance to cross-sensitivity issues. In the context of wound dressings, they need to be fabricated using specific approaches (like printed electronics), and the sensing principles have to align with the process compatibility and measurement methods.

Wearable sensors employ various operating principles to monitor physiological parameters relevant to wound healing (Table 2). Geometric sensing principles utilise changes in the geometry of resistive or capacitive sensor elements to measure strain, pressure and forces. High gauge factors, stretchability and low Young's modulus are crucial for strain/pressure sensor development, achieved by embedding conductive nanoparticles or nanowires in low Young's modulus polymer matrices[114]. Reported wearable pressure/strain sensors often utilise capacitive/resistive sensing mechanisms that track dimensional variations induced by pressure/strain on the dielectric/conductive medium. Porous or fibrous scaffolds of deformable dielectric media are effective for highly sensitive wearable capacitive pressure sensors[296,297]. Piezoelectric sensing principles also have a great interest in mechanical force measurements. The piezo effect is reversible i.e. the application of electricity leads to mechanical vibration on the piezo element or vice versa. Excellent frequency response, feasibility for many shapes, sensitivity and stability promote the piezo materials to be used in pressure orstrain sensing and for actuation.

Bioimpedance sensing exploits the electrical properties of biological tissues, which depend on their composition. A simple bioimpedance measurement over time across the wound can give a clear indication about wound size and healing progress, especially tissue regeneration and remodelling[298]. Electrical impedance tomography (EIT) enables drawing the

Table 2 | Sensing principles for sensors and actuators

Principles	Sensors/Actuators	Sensing output	ref	Significance in smart wound dressing
Geometric	Strain, pressure	Resistance, Capacitance	112,114,438,439	Early detection of pressure ulcer, Control and monitor of bandage pressure/strain to optimise the wound condition
Bioimpedance	Body analysis, Imaging, Moisture measurement	Impedance, Capacitance	215,299,440	EIT imaging of wound, Moisture/pathogen detection
Biopotential	Body vitals (ECG, EEG EMG, etc), body analysis, EIT	Voltage	316,441	Body vitals monitoring
Piezo electricity	Pressure, strain, actuation	Voltage	409,442,443	To optimise wound healing condition and Sensing application
Thermoelectricity	Temperature, strain	Voltage, Current	300,327	Temperature monitoring for infection in wound. Simultaneously the same device can be used a strain/pressure sensor
Electrochemistry	Pathogen, pH, gas, biomolecules, inflammatory mediators, growth factors	Impedance, voltage, current, calorimetry	260,365,417,444,445	The infection detection helps to take timely treatments against bacterial colonisation. This also enables the cellular level monitoring of wound dynamics
Photovoltaics, electroluminescence	PPG, SPO2, pH, Ligh actuation	Current, Light	403,446–448	Enabling the Oxygen measurements in blood for better healing. The light is an actuator for the new cells growth and inhibiting the bacterial growth. Light is also act as an actuator for drug delivery
Temperature coefficient of resistance	Temperature, Moisture, cell growth monitoring	Resistance	301,302,325,449,450	Thermal based systems can be used to monitor temperature and moisture. It can also be applied for cells growth monitoring.

conductivity map of the specimen, revealing physiology, composition and damages, applicable for wound imaging[299].

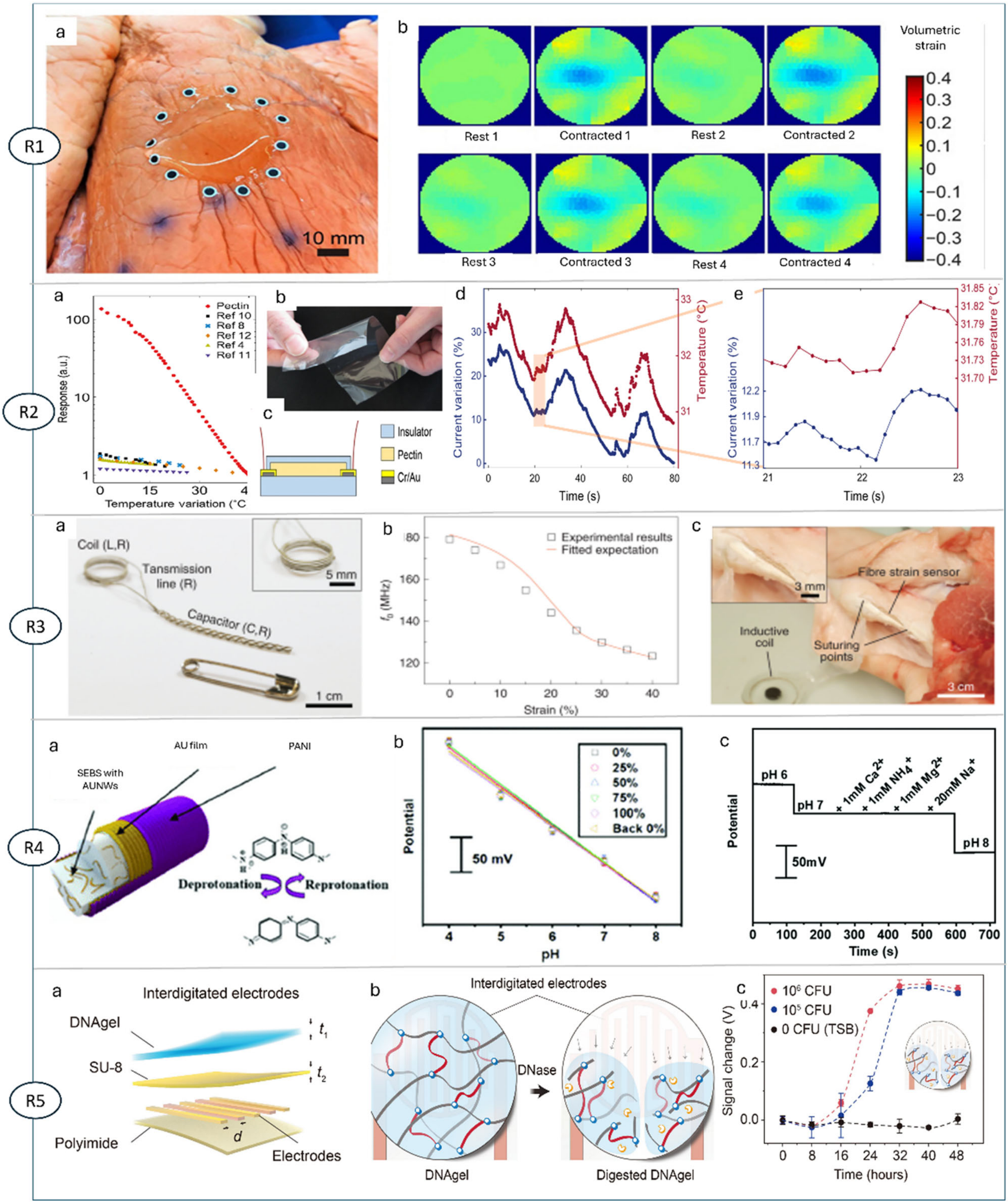
The Seebeck effect, a thermodynamic phenomenon based on coupling of electrochemical potential and temperature, is utilised in self-powered strain sensors and temperature sensors[114,300]. Likewise, the temperature coefficient of resistance (TCR) is an intrinsic material property that represents its responsiveness to temperature variations, quantifying the relative change in resistance with respect to temperature alteration. TCR-based sensors, like the Transient Heat Plane Source (TPS), employ a thin film heater coil as a sensing element. By measuring changes in thermal conductivity or effusivity, TPS monitors specific quantities of interest, applying a defined heating pulse to gauge alterations against standard conditions. Applications span over various fields, including temperature sensing, textile moisture content measurement[79,301] (for wound monitoring) and cell growth monitoring[302,303].

Electrochemical sensors work on the principle of redox reactions and are suitable for biofluid analysis. Electrochemical sensors exploit specifically modified working electrodes to catalysis particular reactions that produce or consume electrons[304,305]. The modification can be an ion-selective coating on the electrode or immobilising a molecule such as bioreceptors or a mediator on it. Based on the electrochemical reaction, there are different measurement methods such as potentiometry, voltammetry and electrochemical impedance spectroscopy. Electroluminescent and photovoltaic principles are making use in light-emitting diodes (LEDs) and photodiodes respectively and they share a complementary device architecture that are widely used for sensing and actuation. In photodiodes (solar cells), incident light generates electron-hole pairs within the active layer, and the electric field within the device separates these pairs, leading to a flow of current. Conversely, in LEDs, when the electric charge is applied, electrons and holes recombine within the active layer, releasing energy in the form of photons, which results in light emission. Light-emitting actuators have been effective in wound healing rate through light therapy. Photoplethysmogram and peripheral oxygen saturation are important biomarkers for wound monitoring obtained from blood flow, indicating the patient's vital status and physical condition is achieved with lighting and photodetectors.

The following section predominantly investigates various individual sensors, actuators, bioresorbable devices and patches that integrate sensor fusion and drug delivery for wound monitoring. The review focuses on sensors and functionalities developed using printed electronics. However, it also includes studies employing hybrid and microfabrication techniques. Non-printed devices are discussed due to their unique approaches in the particular work and potential for future translation into printed ones, ensuring a comprehensive perspective. Conventional microfabrication methods serve as a benchmark for printed electronics, particularly in terms of device functionality, wearable design, cost-effectiveness and sustainability.

Flexible sensors and actuators for wound monitoring

Advanced materials and technological progress empower the concept of smart health patches. This enables a natural interface between electronics and the human body to function as a wearable health monitoring device that fuses multiple sensors and actuators into conformable substrates. This skin-compliant health monitoring patch concept has provided new insights for the health check, diagnosis and real-time treatment as such one can collect physiological health status from the human body and interpret in time for a dynamic intervention[122,306]. In recent times, there is an immense demand for medical big data and personalised medical care and as a result, the demand for smart wearable health patches of wound monitoring is also rising[307,308]. The smart wound sensor patch promises are exceedingly desirable, cogent and compelling prospect for wound care, nonetheless there are mainly a few challenges to address. The technical challenge is to produce miniature, flexible, precise sensors and its communication for a lower cost. The wound-specific challenges are that the developed sensor solution should be multi-parametric, biocompatible, robust, and making non-invasive measurements. The practical challenges associated with these clinical products are



that they need to be easy to use, disposable, and interchangeable with types of dressings[12,309,310]. The sensor system tracks mainly the physical, biochemical and biological markers of the wound. We introduce the single sensor and actuator concepts for the wound monitoring followed by multiparametric systems.

EIT imaging sensor. Assessing wound parameters like colour, size and depth traditionally requires visually inspecting the wound by removing the dressing[311,312]. However, this process can be invasive and disruptive to the healing process. An innovative approach involves wound imaging and monitoring without dressing removal through techniques like EIT. EIT allows for integrating sensing electrodes into wearable foils or textiles, enabling real-time continuous wound monitoring[299,313]. The authors of the article have successfully demonstrated wearable EIT imaging sensors printed on textile substrates, capable of reproducing cross sectional image of bones and tissues in the human forearm[314]. Notably, a

Fig. 5 | Concepts in individual sensors applicable for wound monitoring. R1 implies the realtime printable EIT sensors on living tissues for health monitoring, adapted with permission from ref. 318. Photograph of in situ 3D printed hydrogel ink on a porcine lung (**a**). The EIT based in situ spatiotemporal mapping results of on a porcine lung undergoing cyclic contraction (**b**). **R2** depicts the biomimetic temperature-sensing layer for artificial skins based on pectin, adapted with permission from ref. 331. The pectin-based skin shows high temperature sensitivity and the plot shows its response comparison to other artificial skin research works (**a**). pectin film (**b**) and the sensor architecture (**c**), The electrical current variations (blue dots) plotted against temperature and compared with thermal camera based measurements (red) (**d**), Ambient temperature variations are shown in the zoom in image. **R3** is the stretchable and suturable fibre sensors for wireless monitoring of connective tissue, adapted with permission from ref. 337. Photograph of a wireless fibre strain-sensing system consisting of one conductive fibre where the inset shows the coil for the passive wireless readout. **a** The strain sensing based on the resonant frequency of the wireless system compared with the sensor simulation is plotted (**b**), picture of the wireless sensor sutured onto the Achilles tendon of a pig to monitor the strain (**c**). **R4** shows the development of a stretchable fibre turned in to a potentiometric pH senso, adapted with permission from ref. 364. Depiction of different layers deposited on fibre for the fabrication of a pH sensor (**a**), pH sensor response to pH at different stretching levels (**b**) and the plot shows the sensor selectivity to pH from different ions (**c**). **R5** demonstrate a printable sensor technology based on a DNA hydrogel, named wireless infection detection on wounds adapted with permission from ref. 370. The sensor built up in (**a**), the sensing mechanism shown, DNAgel degradation upon exposure to DNase, resulting in a change in the capacitance of the sensor (**b**), Signal change when sensor is exposed to S. aureus culture.

Berkeley university research team used an inkjet-printed foil sensor and EIT to visualise pressure ulcer formation in vivo on mice, showcasing the technology's potential. The sensor was enabled to foresee the early formation of pressure ulcer by continuous tissue imaging based on bioimpedance mapping[215]. However, such sensors use electrode gels for the interface with the human body. They are therefore not suitable for long time continuous measurements as the electrodes can dry out, thus increasing the skin-electrode impedance, significantly decreasing the signal to noise ratio.

Owing to intrinsic dissimilarities between soft, wet and living human tissues and rigid, dry, and synthetic electronic systems, the development of more compatible, effective and stable interfaces between these two different domains are of a high research interest. Wearable electrodes offer a promising alternative, serving as biopotential electrodes that interface the body with hardware, coupling ionic and electronic conduction. Developing a such soft, ion-conductive interface between the electrode and skin is essential for continuous health monitoring. Innovative advancements like soft conductive dry electrodes (conductive nanomaterials embedded in an elastomeric matrix), long term stable electrode gels/hydrogels are highly promising for such body sensor interfaces[315]. Wang et al. demonstrated eutectogel bioelectrodes composed of deep eutectic solvents, waterborne PU and tannic acid, which show excellent long-term measurement capabilities and biocompatibility with 178% elongation at break and 0.22 mS/cm ionic conductivity successfully applied for wearable biopotential signal monitoring[316]. In an effort to enable a soft, robust ionic interface between stretchable bioelectronic platform and body, the synthesis of a highly conductive polyacrylamide hydrogel achieved by incorporating PANI-decorated PEDOT:PSS is reported. The new type of hydrogel (conductivity 24 S/cm, interface strength 296.7 J/m^2 has been reliably integrated on sensing platform through polymerisation, covalent bonding and hydrogen bonding[317]. An innovative concept of real-time sensor development on living tissues and its dynamics monitoring are enabled through EIT imaging (Fig. 5R1). Applying an adaptive 3D printing system, a hydrogel-based sensor was directly printed onto a respiring porcine lung enabling the continuous spatial mapping of the deformation in real-time, demonstrating the potential for 'lab-on-tissue'[318]. Draeger, a medical technology firm in Lübeck, Germany introduced EIT based non-invasive imaging systems (PulmoVista 500), visualising respiratory functions, providing continuous, regional information about the distribution of ventilation in the lungs[319]. Such advancements could pave the way to wearable sensors for non-invasive, continuous wound imaging and diagnosis.

Temperature sensor. The temperature of the wound is an essential physical wound marker. Initial wound temperature monitoring studies on patient's wound are performed via thermographic heat mapping systems and commercially available temperature sensor integration into the wound bed. However, those thermal imaging systems are expensive, bulky and conventional sensors lack wearability features[320,321]. During the past years, many of the sophisticated wearable temperature sensor designs and fabrication strategies were reported such as flexible[322], strain insensitive[323,324] and highly stretchable[325,326] devices exploiting the material's TCR or Seebeck coefficient for sensing function. In an interesting development, a skin-inspired biocompatible flexible and stretchable temperature sensor is fabricated, possessing the excellent permeability of air and high quality water-proof properties[325]. The stretchability of the TCR based sensor was mainly achieved by choosing specific patterns of the sensing element such as serpentines, embedded in a stretchable medium. Such a tortuous sensor pattern is then able to cope with the tensile stretching by partially unfolding itself avoiding the resistance shift from it. On a different note, a thermoelectric temperature sensor can act insensitive to strain as it measures the open circuit voltage and the stretching induced resistance change will not impact the sensor reading. Jose et al. have reported a highly stretchable sensor that can act as temperature and strain sensing element where they can perform simultaneous sensing without any interference from each other. The sensor element consists of a thermoelectric element that can measure temperature difference between the normal skin and wound as thermoelectric voltage meanwhile the stretching induced resistance change of the sensor act as a strain sensor[327]. Later, another research work reconfirmed the applicability of such temperature-strain sensor[328]. He et al. described a flexible and stretchable colorimetric thermochromic temperature sensible membrane based on phase changing of materials. The colour of these obtained flexible thermochromic membranes could change in response to variation of external ambient temperature and thus presenting perfect thermochromic performance[329].

In order to comply more with the skin, Liu et al. reports a thermally responsive hydrogel inspired by the temperature reversibility of hydrogen bonding for wearable applications. The skin-friendly conductive hydrogel with multiple hydrogen bonds was designed by using biocompatible poly(vinyl alcohol) (PVA), phytic acid (PA) and gelatin[330]. Often the TCR based temperature sensors and thermoelectric sensors have shown limited sensitivity (Like few ohms change/microvolts change). A nature-inspired material design was applied for an exquisite temperature sensitivity (sensitivity of 10 mK in a range of 45 K) and performs temperature mapping mimicking the animal skin. In this remarkable research, Giacomo et al. make use of pectin, a component of all plant cell walls made up of polysaccharides, where the pectin film mimics the sensing mechanism of pit membranes and parallel their record performances. As the temperature rises, cross-linking between pectin molecules decrease exponentially and leads to increased ionic conduction offering excellent sensitivity (Fig. 5R2)[331].

Strain sensor. Manipulating the mechanical forces (strain/pressure) at the wound site to unambiguous ranges has an impact on the pace of wound healing and the quality of new tissues formed[332]. A printed strain sensor on a wound dressing is reported for optimising the wound healing conditions. It consists of a flexible antenna coil whose resonant frequency linearly changes in response to the applied strain, and is described for metered forces on the wound[333]. Likewise, Agarwala et al. demonstrated 3D aerosol jet printed silver patterns for strain sensing to use in bandages[334]. Yang et al. presented a breathable skin-like printed capacitive pressure sensor made up of a nano fibrous membrane showing excellent sensitivity (4.2 k Pa^{-1}), ultra-low detection limit (1.6 Pa) and good breathability (Gurley value 17.3 s/100 mL)[335]. There are strain sensors reported that are not limited to lab testing but even applied on living beings. In a notable approach, Liu et al. have developed a corn protein fibre based smart scaffold that supports cell growth and accelerates wound closure and simultaneously track motions on the skin. This smart scaffold has been tested on rats in vivo and this sensor system can trigger alarms once excessive wound deformations occur[336]. Lee et al. demonstrated a wireless and suturable fibre strain-sensing system that can be used for the real-time monitoring of physiological strains (Fig. 5R3). It consists of a fibre-based passive RLC circuit including a suturable capacitive fibre strain sensor that allows wireless measurement of tiny tensile strains (lower limit of strain detection of 0.001%) generated on a connective tissue. To demonstrate the in-vivo application, the fibre strain sensor was directly sutured onto the Achilles tendon of a posterior leg in an anaesthetised minipig. The resonant frequency of the implanted wireless system was continuously measured with a network analyser during bending and stretching of the leg[337].

Moisture sensor. Wound dressings that regulate the moist balance help to improve tissue regeneration, reduce pain and diminish the infection chances and a regular moisture balance check is covetable[24,338,339]. Moisture content quantification of wearables through sensors have been studied as well, where different sensing principles such as impedance, capacitance, thermal, resistive and visual methods are applied[7,301,340,341]. The drop in impedance or rise in capacitance can be the sign of moisture presence and changes in the heating patterns of the heaters are reported as indicators for such moisture sensors[79,342]. McColl et al. investigated in vitro studies of moisture levels in wound dressings where the impedance principle is applied for real-time measurements using a printed sensor[343]. In addition to moisture balance sensing, wound exudate detection would assist the optimal wound dressing removal and redressing and avoid the chances of maceration. The moisture sensors in wound dressings also help to remove the wound dressing on time avoiding unnecessary wound dressing removal. This would help the unnecessary disturbance in the new cells' growth and is economically impacting with less material usage and manpower[7]. Reeder et al. presented epidermal, microfluidic and electronic systems that adhere to the skin to enable the capture, quantification and analysis of the body fluids[344]. Incorporation of such a wearable microfluidic collecting device and measuring pH and biochemical composition of wound exudate can also be incorporated in future bandages. Though the biosensors are exemplary for health monitoring, undefined biofluid volumes present challenges in composition analysis due to cross-sensitivity and error prone measurements[345–347]. Besides, previous research indicate that wound exudate volume is a significant marker of wound chronicity[348–351]. Jose et al. contributed on this topic through a sensing system model decoupling these parameters using a printed textile sensor platform that combines a thermal and impedance measurements. The thermal sensor (based on thermal effusivity) measures the biofluid volume and the impedance sensor is used to monitor the compositional variations of the fluid with no cross sensitivity[62].

pH sensor. Studies have consistently highlighted the dynamic pH range of wounds, spanning from 4.5 to 8.5, influenced by various factors such as wound type, healing stage and infection status[352,353]. A couple of research investigations are published on wound pH sensor development on the ground of its relevance and the list consists of optical, electrochemical, ion-selective transistors and colorimetric sensors[81,176,354,355]. MOXs have emerged as frontrunners in pH sensor material exploration, with significant attention due to excellent sensitivity and reliability[185]. Flexible potentiometric sensors made with Iridium oxide (IrO2) have demonstrated a pH sensitivity of 72.5 mV/pH (pH 3–11) when electrodeposited and 51 mV/pH (pH 2–12) when dip-coated[185]. Other MOXs like TiO2, RuOx, WO3 and ZnO have also shown promise for pH sensing. However, there are restrictions especially in terms of limited solution processability and compromised flexibility properties. PANI, graphene/graphene-oxide and PEDOT:PSS are predominantly used for the sensing functional layer of the wearable pH sensors[355,356]. Rahimi et al. extensively worked on printed potentiometric PANI pH sensors for wound monitoring. They initially developed a printed flexible pH sensor and later focused on a stretchable pH sensor and transparent pH sensor for in-situ measurements in the pH range 4–10[357,358]. They have also reported a wound monitoring pH sensor array which is biocompatible, non-cytotoxic and shows sensitivity in the range of 55 mV/pH[81].

Various works were published on wearable printed pH sensors for wound dressings and textiles that exhibited a coverage of the entire wound pH range and a superior sensitivity[359–361]. The previously reported work from Jose et al. describes the development of a pH sensor on textiles. It has shown the development of screen printed electrodes on textiles which are functionalised with spray coated PANI for wearable pH sensor[209]. Fibre based pH sensors are also exceedingly interesting for wearables as they are extremely miniaturised, compatible with the skin and can be integrated into conventional bandages[362,363]. A gold fibre based stretchable potentiometric conductive polymer-based pH sensor has shown excellent sensing characteristics as in Fig. 5R4, however its fabrication follows a relatively complex microfabrication process[364]. Solution processed PANI based potentiometric pH sensors are more often studied for wound monitoring due their high sensitivity, biocompatibility, feasibility to miniaturise and low cost producibility[81,209,365,366].

Bacterial detection sensor. The detection of infection in real-time and in-time treatment is a substantial aspect to be considered and S. aureus, P. aeruginosa and E. coli are commonly encountered bacterial species in such wound infections. Bacterial colonies in wounds form biofilms that diminish the efficiency of destroying bacteria through external factors, e.g. immune cells and antibiotics. They are a key factor in the leading of wounds to be chronic, and imperative to detect the bacterial species in the early stage of colonisation. Bacteria detection methods include direct techniques, which involve conjugating bacteria with enzyme-tagged antibodies and indirect approaches, which rely on detecting byproducts or metabolites indicating their presence or absence. Analysis may employ methods like colorimetry, electrochemistry, or luminescence[367]. Interdigitated electrode-based impedance spectroscopic measurements show the feasibility of detecting bacterial colonies in wound fluids where the normalised impedance spectrum presents detection peaks at certain frequency ranges[368,369]. Xiong et al. reported significant work on wound infection using a bacteria-responsive DNA hydrogel (Fig. 5R5). The engineered DNA hydrogels degrade selectively to deoxyribonucleases associated with pathogenic bacteria. The sensor architecture consist of IDEs with DNA hydrogel, responds to bacterial presence through dielectric changes, which can be wirelessly detected using near-field communication[370]. Qiao et al. developed a hydrogel based wound dressing that can detect and monitor bacterial infection and it applies the pH-responsive fluorescence resonance energy transfer (FRET) mechanism for detection[371]. A microbiologically responsive wound dressing composed of biocompatible UV photo-cross-linkable methacrylate gelatine encapsulating both antimicrobial and fluorescent vesicles was investigated for bacterial detection. In vivo and vitro studies of this work demonstrate a visual warning of infection through simple colour changes in a bacterial environment and the proposed wound dressing was able to kill/inhibit the growth of S. aureus and P. aeruginosa[372]. In a different approach, Sharp et al. developed a sensor to detect and quantify pyocyanin, a cytotoxic pigment secreted by the P. aeruginosa bacterial strain and the measurements were carried out using carbon fibre electrodes applying the voltammetry technique[373]. Exploring a different dimension, researchers successfully developed a method to create multifunctional hydrogel by incorporating living cells into them and print them to 3D shape[374]. González et al. demonstrated printing of such hydrogels in

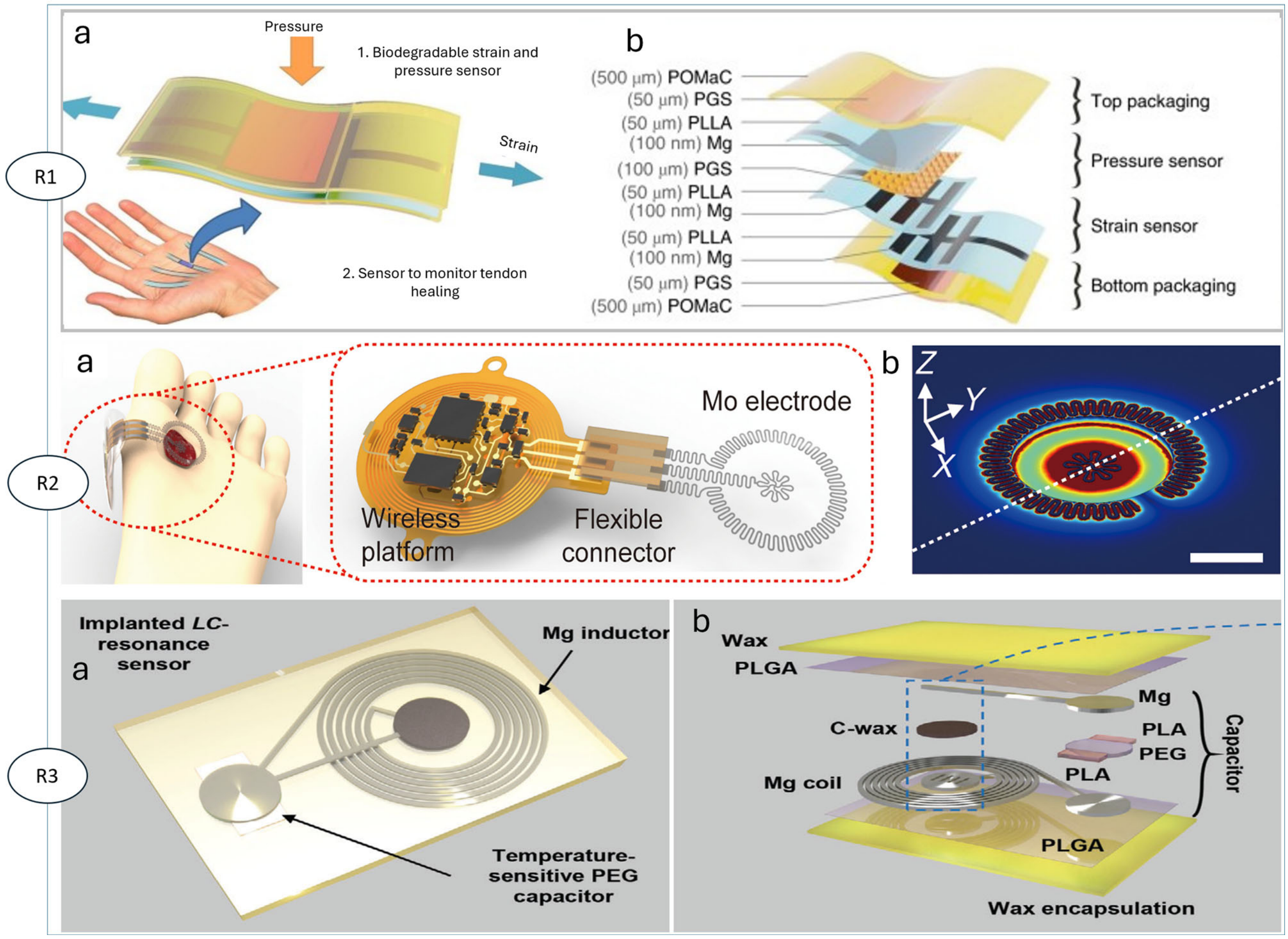


Fig. 6 | Different approaches for bioresorbable electronics. In **R1**, A fully biodegradable strain and pressure sensor is depicted (**a**). The biodegradable elastomer PGS is applied for dielectric layer for the capacitor constituting the pressure sensor. The biodegradable elastomer Poly(octamethylene maleate (anhydride) citrate (POMaC) is used for the strain sensor and packaging where Magnesium is applied as the conductive material for the electrodes (**b**), adapted with permission from ref. 394. **R2** represents a bioresorbable wireless, battery-free system for electrotherapy consist of a releasable flexible connector, wireless platform and a stimulator that consist of a concentric pair of Mo electrodes with filamentary serpentine layouts for stretchability (**a**). FEA results of the electric field between the positive (+) and negative (−) electrodes (**b**), adapted with permission from ref. 396. In **R3**, A bioresorbable wireless capacitive temperature sensor (**a**), where the sensing layer made up of PEG as the dielectric sensing layer with high sensitivity to body temperature range (**b**), adapted with permission from ref. 397.

wound shape, embedding bacteria capable of sensing or eliminating Staphylococcus aureus, a bacteria responsible for infections[375]. Bacterially infected wounds form a biofilm at the wound site and cause bacterial cell accumulation, embodiment and growth. It is well established that biofilms promote the chronicity of wounds and are also challenging for bacterial sensing.

Wound exudate-based biomolecule sensors. The intricate milieu of wound fluid carries a myriad of components, among which cytokines, growth promoters, microbes and metallo-matrix proteases (MMPs) stand out as significant markers for wound monitoring. Research indicates that heightened levels of active MMPs can impede wound healing progress. Various methodologies have been employed for MMP detection, including enzymic activity assays, immunoassays utilising fluorescently labelled antibodies, and electrochemical immunoassays.[376,377]. Krismastuti et al. worked on biomarker sensing in wound fluids, especially with ultra-low level MMP detection[378,379]. They designed and fabricated porous silicon resonant microcavity (pSiRM) and functionalized the sensor using a fluorogenic MMP peptide substrate featuring both a fluorophore and a quencher. This peptide-functionalized pSiRM was then employed as a fluorescence-based optical biosensor for MMPs[380]. In a similar approach, A FRET biosensor was designed and fabricated with a fluorescein isothiocyanate-labelled peptide (Pep-FITC) through covalent binding to graphene-oxide for MMP-2 detection. The proposed sensor work presents detectability at low concentration levels and high selectivity among other MMPs[381]. Hugo et al. developed a graphene based transistor on wound dressings for real-time MMP detection in wound exudate. It consists of an aptamer-based field effect transistor using graphene as sensing channel material which was applied for the selective and sensitive monitoring of MMP-9[382]. Similarly, biosensors for cytokines such as IL-1, IL-6 and tumour necrosis factor-alpha (TNF-alpha) related to wound healing and its influence were reported. Cytokines plays a major role in the delayed healing as its raised level can promote and prolong the chronic inflammatory response[383]. Battaglia et al. prepared fibre-optic surface plasmon resonance (SPR) sensors for cytokines in biological fluids and it has shown a detection limit below 1 ng/mL[384]. High-volume production-compatible roll-to-roll printing of biosensors is appealing for wound monitoring. The study of Noushin et al. showcased roll-to-roll printed IL-6, IL-10 responsive biosensors utilising gold nanoparticles and CNTs as sensing layer. These biosensors exhibit good selectivity, linearity and possess a very low detection limit[385].

Bioresorbable sensors

The concept of 'bioresorbable electronics' was introduced in 2009 as a remedy to the biocompatibility issue of implantable medical devices[386]. Bioresorbable sensors offer a promising avenue in post-surgery wound monitoring, circumventing the limitations of non-biodegradable alternatives. The environmental benefits of biodegradable sensors are evident, yet bioresorbable ones go a step further, eliminating the need for surgical extraction, promoting healing, enhancing patient comfort and curtailing costs and risks linked with removal surgery[387]. The synergistic fusion of materials and technology within wearable devices unveils a category of biomedical innovations that can seamlessly integrate optical or electronic functionality while harmlessly dissipating upon the completion of their intended purpose. However, the fabrication of high-performance, fully biodegradable wireless sensors presents an array of formidable challenges and stringent demands. To clinically implement the current technologies, multiple factors are in consideration, mainly associated with the material performance, the control over its degradation kinetics and biocompatibility of the device. Magnesium, zinc, tungsten and molybdenum (dissolution rate 70, 7, 1.7 and 0.3 nm/hr respectively) are forefront bio-metals in the development of bioresorbable conductive materials[256,388]. Yin et al. systematically studied these metals for transient electronics based on their reactive dissolution and their impact on conductivity, morphology and chemical composition in simulated body fluids[389]. Functional semiconductor materials like silicon (thin film), ZnO and indigo etc. are applied in several works to obtain the desired functionality and bioresorbability[256,386,390]. Other than silicon wafer etching, crystalline silicon films can be derived from the solution processed liquid silane precursor and it is shown to be printable for different device applications like transistors[268,391]. Different works utilised magnesium oxide (MgO) as a dielectric material for transient electronics[392]. Further the substrates and insulating materials complement by a range of options including nature-derived polymers such as cellulose, silk and chitosan, as well as synthetic polymers like PLGA, PLA, PVA, PVG and PEG[390,392].

In advancing on bioresorbable devices, Tao et al. presented a fully degradable, wireless-controlled therapeutic device capable of infection detection and drug delivery. Tested in vivo against Staphylococcus aureus infection, the implantable disappears upon completing its function. The device constitutes of silk as the substrate, magnesium for conductive elements such as the heater and RF coil and magnesium oxide as the dielectric. The silk overcoat pocket primarily influences the dissolution of elements, defining the dissolution rate[393]. An accurate measurement of mechanical forces and tissue contractions of post-surgery wounds is crucial for effective monitoring and recovery assessment. Boutry et al. pursued bioresorbable stretchable pressure and strain sensors for orthopaedic applications, designed to monitor mechanical forces on tendons post-surgical repair (Fig. 6R1). It can detect minuscule strain levels as low as 0.4% and accurately gauge pressures as minute as that exerted by a single grain of salt (12 Pa), all while ensuring that the two distinct sensing components operate without any mutual interference[394]. In a notable advancement, Boutry et al. introduced a robust, conformable arterial pulse sensor with excellent response features for post-surgical blood flow measurements. The pressure sensor, entirely made of biodegradable materials, operates wirelessly through inductive coupling. It utilises magnesium as the conductive part and poly(glycerol sebacate) (PGS) as the dielectric for pressure sensing. The device's encapsulation includes a soft elastomeric POMaC layer in contact with the artery and a stiffer PHB/PHV layer in contact with surrounding muscles. This design of the device architecture enhances sensitivity to artery expansion over respiratory motion[395]. Song et al. developed a miniaturised wireless, battery-free bioresorbable electrotherapy system and impedance sensor for wound monitoring (fig. 6R2). This study was based on a splinted diabetic mouse wound model and confirm the efficacy for accelerated wound closure by guiding epithelial migration, modulating inflammation and promoting vasculogenesis enabled by electrical stimulation. Meanwhile the impedance measurements can assist the wound healing monitoring. As a bioresorbable electrode material, this investigation used molybdenum and studied the degradation in a biological environment[396].

For surgical wounds, a wireless, LC resonance-based passive sensor device is reported, which is fabricated entirely with bioresorbable materials. This sensor was applied for monitoring local body temperatures over clinically relevant timeframes (Fig. 6R3). The element consists of a capacitor made up of PEG (exhibits a strong temperature dependent dielectric constant near body temperatures) and magnesium for electrodes and the inductive coil. The device in vivo demonstrations indicate its stable operation as subcutaneous and intracranial implants in rat models for up to 4 days[397]. The current challenge lies in the vulnerability of metallic electrodes to electrolytic biofluids, impeding the development of viable transient electrochemical sensors. Nevertheless, there are chemical bioresorbable sensors reported for wound monitoring. This study utilises a bioresorbable electrochemical material and unique architecture comprising thin films of tungsten (W) plus tungsten oxide (WOx), taking advantage of their high catalytic activity and uniquely gradual dissolution. The work demonstrates pH and dissolved oxygen concentration in a buffer media[398]. Taking a different approach for pH sensing, the nanostructured sensor presents a linear fluorescence response within the physiological pH range from 4 to 7.5. The sensing mechanism enabled by the swelling/shrinking of its polymeric multilayers (polymers labelled with Rhodamine-fluorophore) was deposited on silica membrane. Implantation studies in mice demonstrated real-time monitoring of subcutaneous pH through the skin and the sensor fully degraded within a week post-implantation[399].

Bioresorbable transient electronics offer promises in wound monitoring, especially in surgery wounds, yet demands substantial research for practical application[256]. Creating a fully biodegradable wireless sensor system poses a major challenge due to scarce of high-performance materials like semiconductors. The material availability and cytotoxicity concerns hinder progress, along with the need to assess degradation by-products' toxicity. Likewise, conventional microfabrication approaches constrain the range of biodegradable functional materials that can be used because of incompatibility with the device fabrication, necessitating considerations around thermal stability, solvent compatibility and mechanical flexibility[400]. Moreover, the degradation rate typically decreases with increased crystallinity and cross-linking density and this also implies to the relevance of solution-processed electronics fabrication routes in bioresorbable electronics. In this case, solution processing-based fabrication and transfer printing methods could replace traditional methods. Functional limitations arise from other sources too, like energy storge devices, confining bioresorbable devices to passive roles. On the other hand, the alternative strategies with wireless power and data transmission increases the complexity of the system and increases the overall size of the devices but eliminates the risks of infection at the skin and wire interface. Nonetheless, recent strides in biodegradable batteries offer high hopes for transient electronics. Tsang et al. proposed a bioresorbable battery using body-friendly electrolytes (MgCl2, NaCl, PBS) and a PCL polymer separator. Magnesium and iron served as anode and cathode, respectively. The battery, encapsulated in PCL, exhibited voltages of around 0.5 V, 0.4 V and 0.7 V with lifetimes of 49 h, 90 h and 99 h in various body fluids[401]. Overcoming the hurdles of bioresorbable electronics could revolutionise wound care and surgical monitoring, but extensive investigation and innovations are imperative.

Actuator modulation for wound healing

Actuators in wound dressings serve dual purposes; accelerating wound healing and facilitating drug delivery. For enhancing the healing process, actuators can intervene through mechanical, electrical, thermal or optical stimulation to the wound site. Light therapy leveraging optical actuators have shown promising to promote wound healing[402]. Studies imply that the use of light sources of specific wavelengths increase the growth factor production, cell proliferation, cell mobility, adhesion and ECM deposition during the proliferation and remodelling phase of wound healing. An innovative wound patch utilising light therapy has been described in the literature. It features hyaluronic acid-based gelatine nanofiber membranes combined with Light emitting diode (LED) arrays to create a skin-attachable patch for wound healing. This wound dressing mimics the extracellular matrix's structure and adheres tightly to the skin, enabling effective coupling with the flexible LED patch[402]. The flexibility and possibility to fabricate onto different surfaces including commercial wound dressing films make printable OLEDs highly promising as wearable light actuator for medical applications. In an EU-funded project MEDILIGHT, the research consortium has harnessed the red and blue light in a disposable wound dressing to enhance the healing and inhibit the infections of chronic wounds[403].

Research data reveals a clear link between wound bed scores and temperature, with multiple studies indicating that maintaining a critical temperature is necessary for normal cellular activity. It implies that at 33 °C, neutrophil fibroblast and epithelial cell activity declines significantly. Moreover, when the wound bed temperature drops below the core body temperature, it's suggested that healing is hindered due to decreased collagen deposition and a reduction in late-phase inflammatory cells and fibroblasts[23,404]. It is critically important to support thermal management for the wounds and actuators enable localised warming which could help to maintain a thermostatic state for healing[405]. Patterned flexible Joule heating elements are incorporated in wound dressings for optimising thermal management of wounds and to activate drug delivery[406]. Printed flexible heaters have already taken a huge interest in wearable electronics due to their form factor and cost benefits[407].

Electrical stimulation boosts wound healing by improving blood flow, easing pain, curbing bacterial growth and promoting cell migration and new blood vessel formation. In a study on 40 stubborn wounds, 78% showed remarkable progress in swelling, inflammation, depth and size after this therapy, indicating healing processes were triggered. All patients with painful wounds experienced pain relief within 2 weeks[408]. However, such systems are not common due to the safety concerns. Thus, introduction of wearable and self-powered electrical stimulation devices for wound care in clinical applications would be convenient. In a topical study it is demonstrated that low-intensity, low-frequency ultrasound activation of a piezoelectric scaffold provides the electrical stimulation. It facilitates proliferation of fibroblast and epithelial cells, enhances the expression of genes crucial for wound healing (collagen I, III and fibronectin). Simultaneously it suppresses the growth of S. aureus and P. aeruginosa bacteria. The mentioned piezoelectric scaffolds are made up of a biodegradable self-charged Poly-L-lactic acid (PLLA) nanofiber matrix[409]. Drug delivery is another key application where actuators play a pivotal role for wound dressings. Actuators can be triggered to release medication in a controlled, targeted manner directly into the wound bed. This approach minimises systemic side effects while ensuring adequate local therapeutic levels.

Multi-sensing modalities for wound monitoring

While single sensor wound monitoring provides useful data on individual parameters, wound healing involves complex, interconnected biological and physiological processes. Measuring just one parameter fails to capture this intricate nature, especially for chronic wounds complicated by factors like infection and diabetes. A different approach of simultaneously tracking several biomarkers and vital signs is critically needed. This provides a richer, holistic view of the wound environment and healing status in real-time. Key parameters to collectively monitor include bioburden, oxygenation, inflammation markers, moisture, pH, temperature and metabolic activity. Integrating data across these interconnected streams unlocks new clinical insights for personalised interventions and optimised healing for even the most complex chronic wounds. Such wound monitoring multi-sensing patches need to be designed with system level design of different components and an example case is shown in Fig. 7a. A generic wireless smart patch design comprises three sections: a bio-interface module with soft sensors and stimulation electrodes for wound interfacing, an NFC electronics module for power management, data conversion and computation, and a communication section enabling external device connectivity via the NFC module[410,411].

Further, a paper based low cost, breathable, single-use bandage with printed sensors is reported for chronic wound management. This bandage can measure the pH and uric acid concentration of the wound and also it facilitates early detection of pressure ulcers through EIT based imaging[42]. M.F. Farooqui & A. Shamim briefed a low-cost printed paper based smart bandage that has an intelligent design consisting of a disposable sensing part and a reusable readout part. The sensing part consisted of a capacitive principle bleeding and pressure sensor and resistive principle pH sensor. The pH sensor was intended to be used for single time exposure to wound fluid and the dryness of the bandage can have an adverse impact on pH readings[96]. End-user cumbersome is posed due to bulky batteries, antennas and discrete components in wearable electronics. Zheng et al. presented an electronics-free, AI-enabled paper based multi-sensing platform (PETAL) for comprehensive wound assessment through deep learning algorithms (Fig. 7b). This biocompatible sensing patch leverages a panel of colorimetric sensors to holistically monitor five clinically relevant markers/parameters related to inflammation, infection and wound conditions. The PETAL platform can classify healing vs. non-healing status of burn wounds with 97% accuracy by analysing exudates in ex situ. By performing in-situ multiplexed wound sensing, the PETAL sensor paves the way for integration into active wound dressings[412].

Another significant approach utilising the latest fabrication routes, is the laser-induced graphene-MXene hybrid scaffold-based sensor for a silicone based stretchable bandage. The bandage consists of a pH, temperature and uric acid sensor and the sensor shows excellent sensitivity, repeatability and selectivity. However, the stretching induced resistance change from skin movements may adversely affect temperature readings[413]. In a recent advanced study towards wound monitoring (in Fig. 7c), a multi-sensing platform called VeCare is introduced, in which a microfluidic bioanalytical device for temperature, pH, tumour necrosis factor–α, IL-6, transforming growth factor–β1 and bacterial detection is proposed. The flexible sensor system used a microfluidic exudate collector and multiplex sensor measurement. The investigation extended to testing on mice wounds and patients with venous leg ulcers and presented promising results for in-time personalised wound management[414]. Guo et al. presented a pro-healing Zwitterionic skin like hydrogel sensor that consist of multi-sensing sandwich layer architecture for real time monitoring of infection, strain and glucose. The sensor architecture is shown in Fig. 7d and the monitoring and distinction of temperature, mechanical and glucose information without signal interference is achieved[415]. As a more biomimetic approach, Brown et al. presented a smart stretchable, breathable sensing-actuating wound dressing platform on a micro-fibrous silicone substrate. It emulates the native epidermal mechanics and physical extracellular matrix architecture for intimate bio-integration. The multiple biosensor array can continuously examine inflammatory biomarkers such as lactate, glucose, pH, oxygen and wound temperature that correlates to the wound healing status. Additionally, a heating element was incorporated to maintain the optimal thermal conditions at the wound bed[416]. Kalidasan et al. presented a bioelectronic platform using surgical threads to monitor post-operative wounds (Fig. 7e). These multifilament sutures, coated with a conductive polymer and equipped with capacitive sensors on pledges, enable the monitoring of physicochemical conditions in deep surgical sites. Their study showcases the application of these surgical sutures in living pigs, demonstrating their ability to monitor wound integrity, gastric leakage and tissue micromotions[417].

Wound monitoring and on-demand therapy

Advanced wound care with sensor-responsive wound dressings is agile for future wound care. Wound dressings are used to be passive and it is challenging to control the release profile of drugs based on the demand of the wound[418]. However, recently, drug delivery dressings activated by exogenous stimuli (i.e. light, heat and electricity) enable the precise moderation of drug release. Sensor-based data-driven drug delivery is an area of intelligent wound dressings which is progressing[419]. Smart wound monitoring and responsive systems are built upon different substrate platforms. Such systems can be on paper, foils, textiles, hydrogels or surgical sutures. Each of such platforms has impact and significance on different aspects of wound dressings. It spans from compatibility for fabrication, device performance, wearability, wound compatibility, breathability, biodegradability and drug delivery. Ochoa et al. built a low-cost alternative paper-based platform for oxygenation. The fabrication of the flexible oxygenation patch makes use of

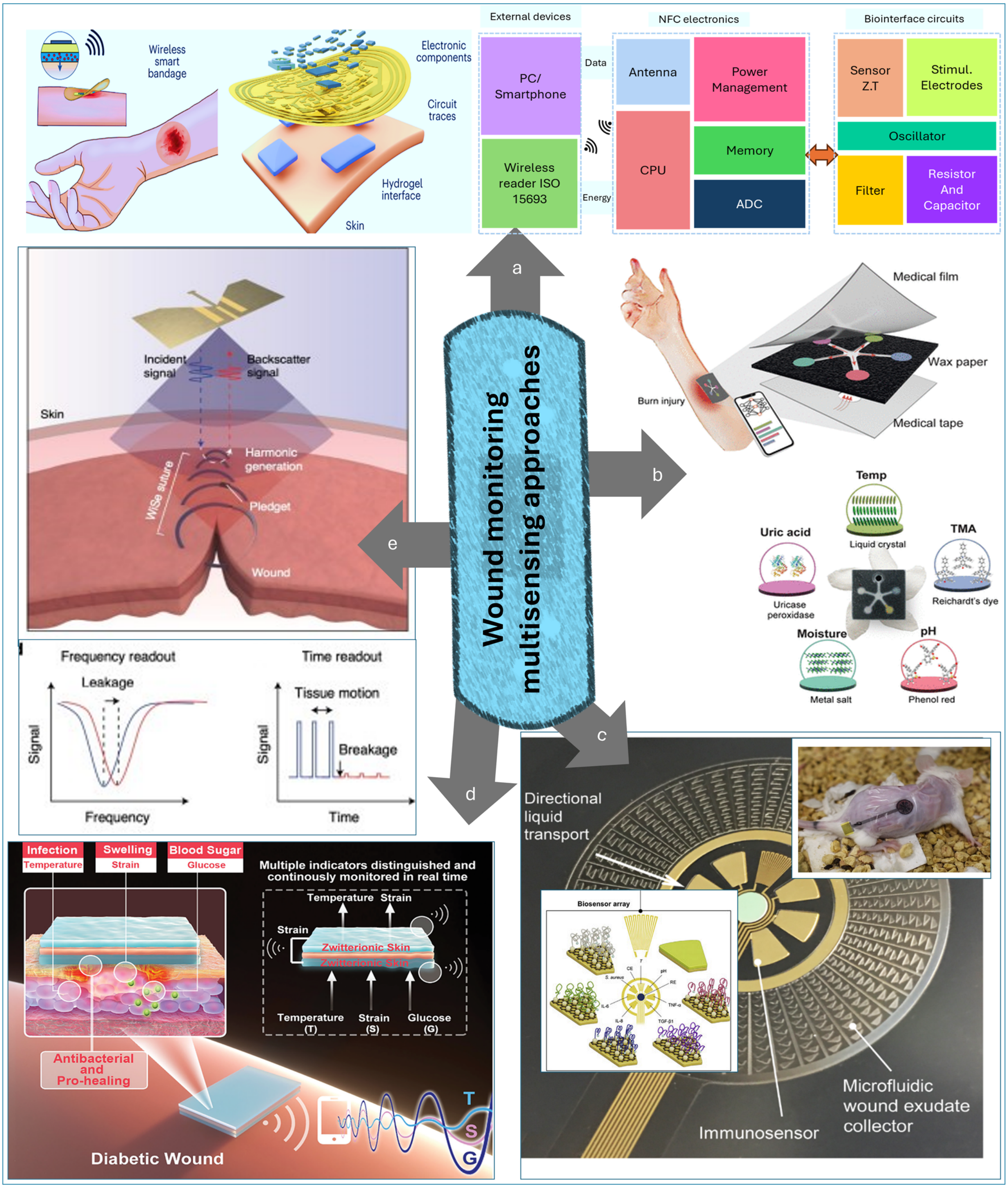


Fig. 7 | Different approaches in multisensory wound bandage concepts. A flexible wound monitoring platform and its system design in schematics is shown in (**a**), *reused with permission from ref.* [410]. A passive calorimetric sensing platform called petal is Illustrated where the wound healing biomarkers such as pH, temperature, moisture, uric acid, etc, can be extracted with PETAL sensor and a smart phone (**b**), *adapted with permission from ref.* [412]. A multiplexed multi biosensing platform named VeCare, that allows normal skin function by letting oxygen in and moisture vapour out. This microfluidic bioanalytical device can monitor temperature, pH, tumour necrosis factor–α, interleukin-6, transforming growth factor–β1 and bacterial presence (**c**), *adapted with permission from ref.* [414]. A skin like sandwiched zwitterionic hydrogel-based monitoring system where it can monitor temperature, strain and glucose level at the wound site. The capacitance of the upper/lower layers is insensitive to glucose–temperature and variable to measure strain, where the resistance of the upper hydrogel to measure strain and temperature and it is insensitive to glucose; and the resistance of the lower hydrogel can detect three indicators (**d**), *adapted with permission from ref.* [415]. Suture based wound monitoring where the capacitive sensor (based on threads coated with PEDOT: PSS) transduces changes in the physiological environment. The biological events such as infection or leakage transduce into shifts in the resonant dip of the harmonicignal spectrum measured by the wireless system (**e**), *adapted with permission from ref.* [417].

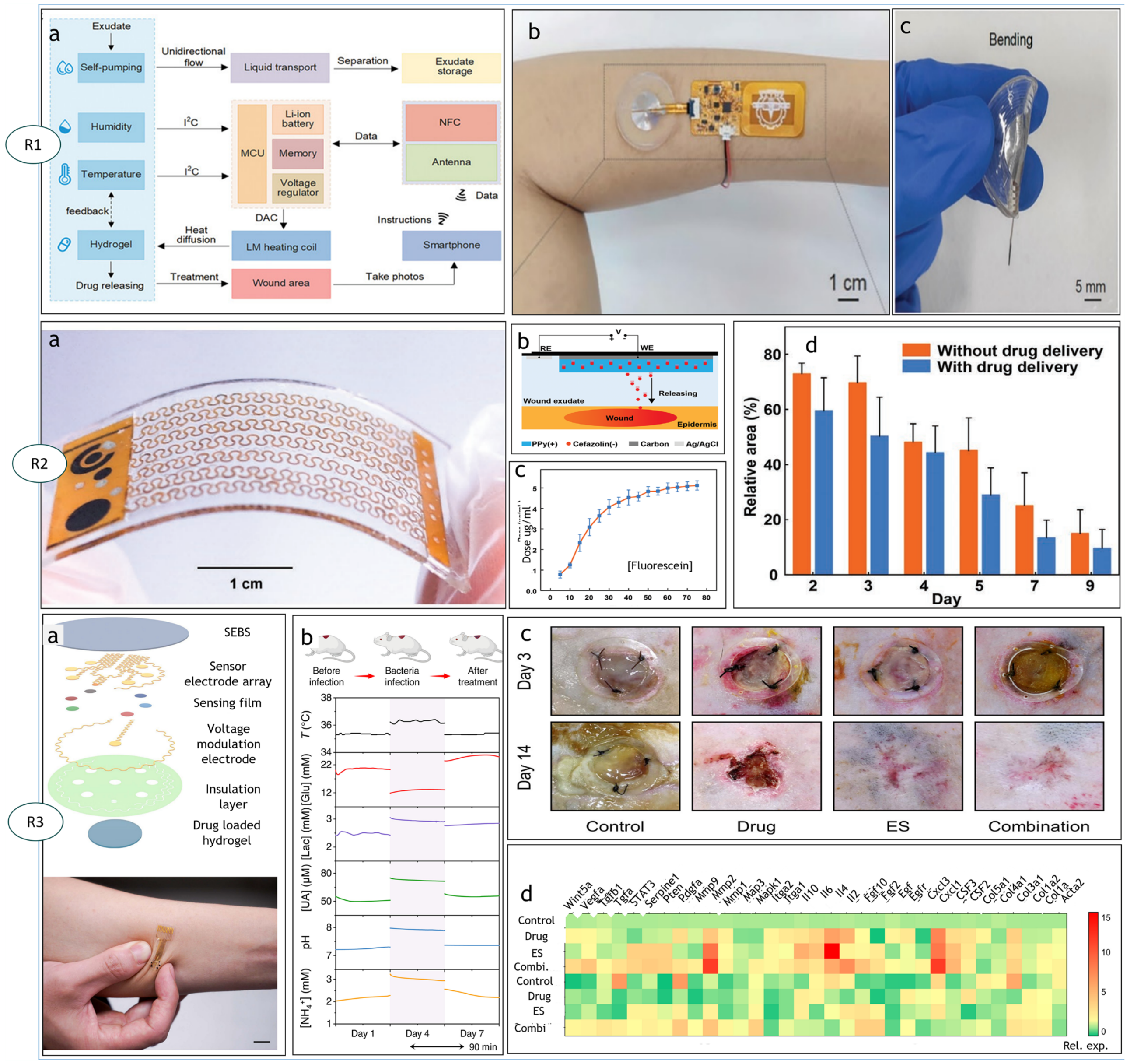


Fig. 8 | Wound patches with sensing and together with demand drug delivery approaches for smart wound care. **R1** shows the system-level block diagram of the design of a smart wound monitoring system (**a**) and the photograph for the application of the system on the human body (**b**). The systems can be bendable as shown in (**c**), *adapted with permission from ref.* [406]. **R2** discuss a work on wound dressing with temperature and pH monitoring for diagnostics where the drug delivery is controlled with electric potential, *adapted with permission from reference*. The image of the smart bandage (**a**), the schematic of the drug release with electric actuation and the plot for controlled drug release is shown in (**b**, **c**). The impact of on demand drug delivery on enhancement of wound healing monitored on the basis of wound area reduction with time (**d**)[129]. **R3** depicts Sani et al. work on the multi model biosensing and -therapy is integrated in a stretchable wound patch, *adapted with permission from reference*. The image shows the patch layer stack and when it is applied on skin (**a**), preclinical in vivo measurements of wound biomarkers demonstrate the comprehensive wound status analysis (**b**). Images of Wound for day 3 and 14 are compared for on demand drug treated, electro stimulation therapy applied and combination of both applied in comparison to control wound (**c**) and its qRT-PCR analysis for the library of wound biomarkers are given (**d**)[260].

paper as a substrate and applied scalable printing technology for material deposition. The platform can sense oxygen in a range of 5–26 ppm and generates a double or triple oxygen level in the wound bed[47]. An imperative feature of paper-based sensors is that they can spontaneously transport body fluids via capillary flow and are spared from the use of pumps[367]. The paper-based systems have the benefit of cost, biodegradability and decent surface qualities for printing functionalities, where on the other side silicone foils are better in terms of device fabrication, barrier properties, soft interfaces and skin-like features. In an approach to achieve effective synergy between wound exudate management and on-demand wound therapy, Ge et al. presented a wireless closed loop wound dressing based on multilayer silicone (Fig. 8R1). The wound exudate is spontaneously pumped into the microfluidic channel for storage to reduce the likelihood of bacterial infection and the potential damages in the surrounding skin. Meanwhile the status of the wound can be monitored through the temperature and humidity sensor and this feedback from the wound site activates the heating actuation to execute the on-demand drug release[406]. A recent research publication briefed the battery-less (through inductive coupling with smart phone) smart bandage of a double layer structure for closed-loop monitoring and treatment system. The top layer is a reusable part for NFC-enabled flexible circuit board enabling wireless power harvest, drug delivery control, signal process, temperature sensor and wireless data transmission. The bottom layer of the wound dressing is a disposable part comprise of a uric acid sensor, a pH sensor and a drug delivery electrode with antibiotics coated membrane. Screen printing process has been used for the electrode and interconnection deposition. This Silicone based smart wound dressing is demonstrated and applied for in-situ animal wound healing studies (Fig. 8R2)[129].

Hydrogel- based wound monitoring and drug delivery systems are growing rapidly due to its intricate features such as skin-like properties, biodegradability, self-healing capacity, no-adhesion on tissues and offering moist environment for wound heling. Such hydrogels can crosslink the drugs of interest and are able to deliver on demand with the activation via light or heat. Moreover, the hydrogels can be printed such that it can be fabricated with 3D printing technologies. Topical research on a hydrogel based visual/colorimetric sensing and drug delivery system shows promising outcomes for wound management. The hydrogel patch presents a dual-sensing approach, whereas the visible colour changes of cabbage juice and brilliant yellow loaded beads are used for measuring the bacterial density variation and pH changes. Moreover, they can deliver antibiotics to the wound site in order to eradicate bacterial colonies[420].

Pang et al. reported an innovative wound dressing that was designed and fabricated for detecting infection from temperature measurements and activating antibacterial hydrogel through LED lighting. The smart bandage is structured so that a flexible electronics layer including a temperature sensor and LEDs encapsulated with silicone rubber in the top layer and UV responsive hydrogel in the bottom layer form a dense and flexible wound device. The infection induced temperature changes actuate the LEDs and lead to a controlled release of antibiotics from hydrogel[122]. Temperature and pH are the two significant physiological parameters for wound monitoring and Mostafalu et al. worked out an intelligent bandage based on these sensing parameters to realise a stimuli-responsive drug releasing system. Apart from the sensing elements, the bandage contains a microcontroller to control the flexible heater and a hydrogel loaded with thermo-responsive drug carriers in the wound dressing. The temperature and pH sensors integrated onto bandages monitor the wound healing in real-time and a flexible heater actuates the drug delivery from the thermo-responsive hydrogel on demand[421].

The prior mentioned works on smart drug delivery patches that rely on one or mostly two parameters for the on-demand therapy. The heterogenous nature of wound exudate poses substantial challenges for the accurate bioanalysis and timely act of drug delivery. In a more advanced approach, Sani et al. introduced multimodal sensing through a stretchable, flexible bioelectronic platform that can comprehensively monitor the physiological wound condition and can also perform combination therapy (Fig. 8R3). The biomarker analysis is made out of temperature, pH, ammonium, glucose, lactate and uric acid and the combination therapy consists of electro-responsive controlled drug delivery for antimicrobial treatment and exogenous electrical stimulation for tissue regeneration. To mitigate the accuracy issues of the biosensors in wound environments, they make use of a porous PU-based membrane that assists as exudate diffusion limiting layer to protect the sensor, enhance long-term stability, linearity and sensitivity. Preclinical in vivo studies show multiplexed spatial and temporal wound biomarker analysis of metabolic and inflammatory wound parameters in real time with high accuracy and electrochemical stability. The combination therapy enabled the accelerated cutaneous chronic wound healing in diabetic rats[260].

Once the wound forms a scab, it prevents the interface between the sensor and wound creating a barrier to obtain the real status of the wound inside. Surgical sutures and threads are potentially fantastic platform for wound monitoring and drug delivery. Sutures based smart systems can be bioresorbable, non-bulky and a most convenient for the surgical wounds. Most importantly, the sutures are interacting with the wound from inside the skin and not from the surface. In a noteworthy research work, Mostsfalu et al. also worked towards a functionalised multithread-based sensing platform integrated with thread microfluidics to monitor wound repair which was tested on mice wounds. The thread-based sensor system of pH, temperature and strain provides freedom for the integration of the sensing system into current practices of wound dressings[363]. Inspired by ancient animal-derived suture materials, Lee et al. report the development of a biocompatible decellularized gut suture platform capable of monitoring inflammation, delivering a broad array of therapeutics on demand[422]. The authors of this review article themselves recently published a research on the development of functional material deposition patterning on fibres with printing technology enabling the concept of Lab-on-fibre. The patterning capabilities within fibres or threads open an interesting arena for multi-functional sensing, actuation, channel integrated drug delivery, communication and energy storage, within the single fibre[423].

Discussion and outlook

Smart wound dressings exemplify an innovative application of printed electronics, showcasing a synergistic interplay between technology and functionality. The inherent nature of wound dressing platforms necessitates the use of specific types of substrates characterised by absorption, roughness and softness (such as textiles, hydrogels, silicone and sutures), governed by the type of wound. Conventional electronics fabrication methods are unsuitable for such versatile substrates, making printed electronics an obvious choice. The conformability and stretchability attributes of printed electronics provide better wearable quotient and avoid the cumbersome nature of traditional electronics. The demand for smart wound dressings presents a compelling perspective, driven by the increasing number of patients and the growing interest in healthcare digitisation. This demand translates into the need for high-volume production capabilities, such as roll-to-roll manufacturing setups, to ensure economic feasibility. Addressing these substantial yield requirements necessitates the adoption of simple and economic electronics manufacturing processes. Although the application here often does not demand high integration density and long term stability, often lab-scale electronics fabrication facilities use microfabrication tools for developing smart bandages due to the availability and well acceptance of the process. We reviewed such non-printed smart wound dressings due to their novel concepts, and cross-applicability with printed electronics based smart bandages. These microfabricated systems align with printed electronics' scalability, cost-efficiency and compatibility with flexible, sustainable substrates—addressing demands for mass production and eco-friendly solutions in next-generation wound care technologies. The sustainability objective is particularly vital in high-volume, disposable electronics. The disposable nature of wound dressings necessitates the use of nature-derived/environment friendly and biodegradable materials to mitigate the growing threat of e-waste. Moreover, printed electronics technology is the focus of studies and government initiatives with sustainable electronics at their core (for example, 'SUSTRONICS' and[424] Responsible Electronics[425]). Further focus on scaling up these technologies from the proof concept into commercial scale needs much concentrated efforts and high TRL research funding[426].

The world-wide smart wound dressing market generated a revenue of USD 0.81 billion in 2023, and is expected to reach USD 2.90 billion by 2033, with a CAGR of 13.6%[427]. Efforts to commercialise smart wound dressings incorporating flexible sensors are growing. The wound monitoring platform developed at universities—Ve care at National University of Singapore, Caltech's smart bandage are getting in to the - clinical studies[428,429]. Printed functionalities driven innovations, such as the graphene-based smart bandage by CNRS Institute's spin off Grapheal and Absorbest AB's DryMax Sensor are some of them in the line[430]. Despite encouraging in vitro and in vivo outcomes, along with preclinical validation studies, these smart dressings have yet to transition into large-scale clinical trials—a gap attributed to the long regulatory pathways that they need to navigate[431]. The proposed designs, which integrate multifunctional materials, diverse sensors and actuators for combined diagnostics and therapy, face hurdles tied to approvals and validation, given their layered complexity. The intricacy of regulations and validation of such smart wound dressings originates from this diverse material safety and from the system classification where it needs approval as a drug and device causing delays in reaching to market[426]. Beyond the regulations, market entry and success is dependent upon proving cost–effectiveness, seamless integration with health-care protocols

Table 3 | bio-degradable and biocompatible materials for smart wound dressing

Material	Material property	Bioresorbable/ Biodegradable	Synthetic/bio sources	Toxicity	printability	Key applications	notes
PLA	Piezo-electricity	Bioresorbable, biodegradable	Synthetic	Biocompatible	Printable	Actuation/Insulation	451–453
Mg	Conductive	Bioresorbable, biodegradable	Synthetic	Biocompatible	Printable	Sensing elements, bioelectrodes, Antennas, heating elements	389,451
Zinc	Conductive	Bioresorbable, biodegradable	Natural mineral sources	Biocompatible	Printable	Sensing element, Antennas, heating elements	389,454,455
Fe	Conductive	Bioresorbable, biodegradable	Synthetic and bio sources	Biocompatible	Sol. processable	Conductive traces, bioelectrodes	389,392,456
Mo	Conductive	Bioresorbable, biodegradable	Bio sources	Biocompatible	Printable	Sensing elements, bioelectrodes, sensing elements	389,457,458
PEDOT: PSS (p and n type)	Conductive/semi-conduct/ Thermoelectric, redox active	Biodegradable	Synthetic	Biocompatible	Printable	Antennas, Stretchable electronics, Electrochemical, strain and, bio electrodes	257,459,460
CNTs	Conductive/Thermoelectric	Biodegradable	Synthetic	Biocompatible	Printable	Stretchable electronics, bio electrodes, strain, Electrochemical, Temperature sensors	253,461
Prussian blue	Semi-conductor, redox active	Biodegradable	Synthetic	Biocompatible	Sol. processable	Uric acid, Glucose and Gas sensing	462,463
Polyaniline	Semi-conductor, redox active	Biodegradable	Synthetic	Biocompatible	Printable	pH and electrochemical sensing	81
ZnO	Semi-conductor, redox active	Bioresorbable	Synthetic and bio sources	Toxic at high rates	Printable	Transparent conductors, electrochemical sensors, transistors	464–466
MgO	Insulator dielectric	Bioresorbable	Synthetic	Biocompatible	Printable	Dielectric layer, sensors	451,467,468
Si	Semi-conductor	Bioresorbable	Bio sources	Biocompatible	Printable	Transistors, sensors	256,465,469
POMAC	Insulator	Bioresorbable, biodegradable	Synthetic	Biocompatible	Printable	Strain sensor, Elastomeric circuits, substrates	256,451,470,471
PLGA	Insulator	Bioresorbable, biodegradable	Synthetic	Biocompatible	Printable	Strain sensor, Elastomeric circuits, bioelectrodes, Hydrogels, substrates	472
PEG	Organic insulator	Bioresorbable, biodegradable	Synthetic	Biocompatible	printable	Bioelectrodes, temperature and pressure sensors. Hydrogels	473–475
Nano cellulose	Insulator	Bioresorbable, biodegradable	Bio sources	Biocompatible	Printable	Strain, pressure and moisture sensor. Hydrogels and substrates	233,476,477
Chitosan	Insulator	Bioresorbable, biodegradable	Bio source	Biocompatible	Printable	Humidity and gas, sensors. hydrogels	236,478
Pectin	Insulator	Biodegradable	Bio source	Biocompatible	Sol. processable	Temperature sensor	331
Silk	Insulator	Biodegradable, bioresorbable	Bio sources	Biocompatible	Printable	Transistor, Sensor, substrate	390,451,479

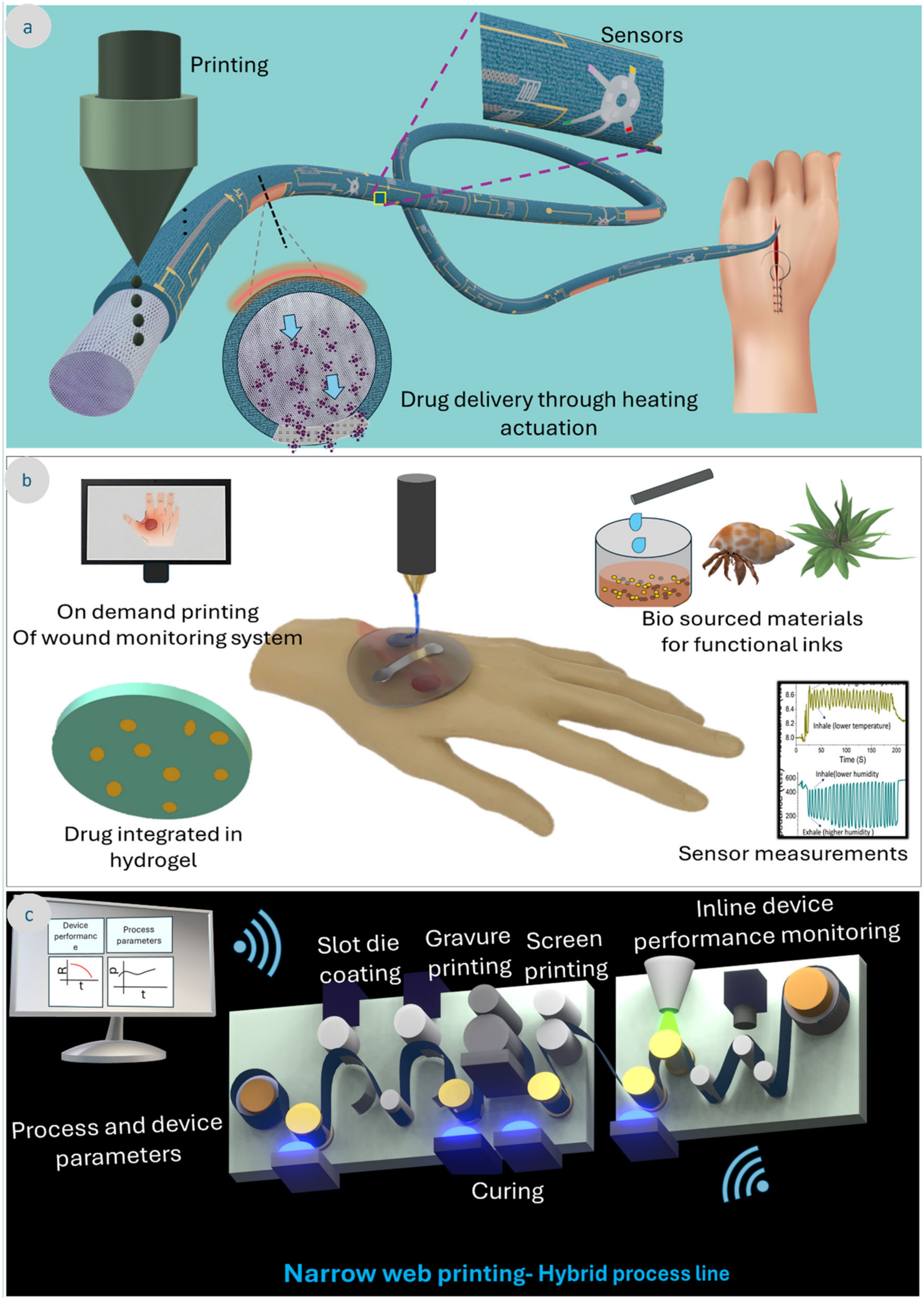


Fig. 9 | Concepts focused on future of printed electronics in wound monitoring applications. Depiction (**a**) shows the sensors and actuators directly printed on sutures. It demonstrates the potential of printing for biodegradable, biocompatible multi sensing and drug delivery platform for surgical wound monitoring. Depiction (**b**) implies the promise of on demand design and fabrication of hydrogel based wound care systems based on the doctor diagnosis. The sensing elements and drugs can be customised depending on the wound condition. Adaptive printing approach with bio sourced material enables skin like features and sensor data driven drug delivery. Depiction (**c**) shows the inline process for the production of wound sensing patches where narrow web line with interchangeable printing process units. The inline characterisation of the sensors together with machine learning assisted optimisation of the production process parameters enable more efficient production line for the smart patches.

and meeting the priorities of users. Over time with advancements in regulations and exploiting the synergy of printed electronics in wound dressings will likely drive the market of sustainable, and economic smart wound care.

The research approaches and product development phases need to explore the opportunities of synergy that exist for printed electronics and smart wound care and there is necessity to address the challenges associated with it. The key learning points from the review are summarised here.

a. The widespread adoption of printed smart wound dressings raises concerns about potential e-waste challenges. One approach to address this issue is through sustainable eco-design, such as adopting a modular platform that incorporates a disposable sensing component and a reusable electronic readout unit. The disposable component, fabricated from printable biodegradable functional materials, requires further research to enhance its attributes and performance.
b. Transient/biodegradable electronic materials are generally categorised as inherently susceptible to instability[432]. As the fabrication of transient electronics for wound monitoring is a critical challenge and very often requires creative and unorthodox approaches to manufacture, where printed electronics can fit well. Table 3 shows the enormous opportunities from emerging ecofriendly materials (mostly we discussed in the previous sections) for printed electronics with the end application towards wound monitoring.
c. A few of the important bio markers in wound monitoring needs biosensing aspects and printed biosensor secured much attention due to its flexibility to be applied in all kinds of substrates, ease of fabrication and disposability. However, the limited time stability of the biosensing elements is a major challenge in the health monitoring sensors. The implementation of bio sensing still needs improved materials and its engineering approaches to bring the biosensors to exploit better for wound care.
d. Establish more specificity regarding wound types (surgical/burn/bedsores) in the selection of biomarkers and type of smart wound dressing. The precise segregation and quantification of the wound fluid composition and to render high-fidelity wearable biomarker data is important[433]. Especially the detection and quantification of biomolecule such as cytokines, MMPs, microbial markers and inflammatory mediators needs further emphasis. Such biomolecules in wound fluid can provide deep insight about healing and the biochemical profile of the wound. The type of smart wound dressings should also need to be specific for wound types/biomarkers (eg. Sutures for surgical wounds).
e. Multi-functionality in compact designs is a key area of interest, focusing on integrating dual or triple sensing capabilities into a single sensor to enhance performance while addressing space constraints. This approach may require innovative solutions, such as novel sensor architectures, strategic material selection, or advanced sensing principles. For instance, a single sensor capable of simultaneously measuring temperature and strain or pressure and moisture could exemplify such advancements.
f. Design principles must guide the functionality development with good wearability form factor: Simple and easy to use, preferably with minimal bulky electronics to maintain inherent dressing properties. For example, single bioresorbable fibres that can perform multi-sensing, communication and drug delivery (Fig. 9a), or a colorimetric sensing platform paired with an AI-enabled smartphone-based evaluation system.
g. It is necessary for sensing elements to keep intact under large deformations during body movement and maintain adhesion in highly hydrated conditions. Therefore, new generation of materials (with tissue-like properties) and biomimetic designs can be adapted to provide opportunities for wound monitoring devices in multiple wearable conditions and to meet biocompatibility.
h. On demand tailored solutions for wound treatment can be the next big step in wound treatments. Tissue like features of hydrogel offers many innovation avenues in wound dressings and also give merits as a biodegradable platform. Adaptive printing system can help to develop diagnosis assisted custom made hydrogel based smart wound dressing (Fig. 9b).
i. The scale-up phase is considered to be the most delicate phase of development[434]. From proof of concept to the upscaleable level, there exists a huge gap. Exploiting the processes available in an industrial scaler to lab scale development of the smart wound dressing (for eg: screen/gravure printing and/slot die coating process) will reduce inertia occurring at the interface of lab to fab transition. Emphasis on pilot scale studies with narrow web process line for easy handling, enabling hybrid interchangeable printing units. Also, make use of machine learning assisted production process would be a great enabler towards more autotuned production (Fig. 9c).

Data availability
No datasets were generated or analysed during the current study.

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Author contributions

M.J.: Investigation, Conceptualisation, Writing—Original Draft, Review & Editing. M.T.V.: Writing, Review & Editing. L.N.: Visualisation, Review & Editing. L.R.: Investigation, Writing. A.C.H.: Investigation, Visualisation, Review & Editing. D.C.: Review & Editing. R.T.: Review & Editing. A.P.: Writing, Review & Editing. J.K.: Supervision, Writing, Review & Editing. W.D.: Conceptualisation, Supervision, Writing, Review & Editing, Funding Acquisition. All authors have reviewed and approved the final version of this paper for publication. Each author agrees to take responsibility for all aspects of the work to ensure that questions about the accuracy or integrity of any part of the work are appropriately investigated and resolved. All authors have read and revised the manuscript.

Competing interests

The authors declare no competing interests.

Additional information

Correspondence and requests for materials should be addressed to Manoj Jose or Wim Deferme.