Volume 2
Number 6
21 March 2021
Pages 1777–2142

Materials Advances

rsc.li/materials-advances

ISSN 2633-5409

ROYAL SOCIETY OF CHEMISTRY

REVIEW ARTICLE

Marco Carlotti, Virgilio Mattoli *et al.*

Conformable on-skin devices for thermo-electro-tactile stimulation: materials, design, and fabrication

Materials Advances

REVIEW

Cite this: *Mater. Adv.*, 2021, 2, 1787

Conformable on-skin devices for thermo-electro-tactile stimulation: materials, design, and fabrication

Arianna Mazzotta, Marco Carlotti * and Virgilio Mattoli *

Conformable electronics is an emerging and innovative research field investigating functional materials and electronic devices capable of adhering and conforming to non-planar surfaces such as human skin. Conformable devices find applications of high economical and scientific interest in healthcare, human-machine interfaces, wearable electronics, robotics, and the internet of things. Compared to the more widely used rigid and bulky electrodes often employed in wearable platforms, the former offers multiple advantages such as imperceptibility, low cost, and the possibility of continuous use, being able to follow the multiple deformations of curvilinear living tissues. In recent years, much attention has been paid to the development of soft and skin-wearable sensors, but an additional area of great interest is certainly that of conformable electrodes capable of providing some kind of stimulation. This review provides an overview of the most attractive and innovative conformable devices developed to stimulate the human body through thermal, electrical, mechanical, and optical stimuli. In particular, it focuses on the functional materials employed, the fabrication techniques involved, and the design solutions proposed to improve the performance of conformable stimulation devices.

Received 19th October 2020,
Accepted 8th February 2021

DOI: 10.1039/d0ma00817f

rsc.li/materials-advances

1. Introduction

The growing demand in recent years for miniaturized and portable electronics has led to the development of numerous conformable and flexible technologies. This was (and still is) driven by the revolutionary opportunities offered by a class of materials and devices able to adhere and conform to nonplanar surfaces.[1] These devices find applications of high economical and scientific interest, such as in healthcare, human-machine interfaces, wearable electronics, robotics, and the internet of things. Technological advancements like new lithographic and microfabrication techniques allow us to reach complex design and unprecedented resolutions,[2–5] and the implementation of innovative functional materials combined the necessary mechanical properties with appealing electronic characteristics.[6]

Compared to the former, most of the electronics employed in the development of wearable platforms are planar and rigid, they are incompatible with the curvilinear and deformable living tissues, they lack conformability, and thus they result in high-impedance, unreliability, and uncomfortable contacts.[1] Instead, conformable electronic skin devices, which have the intrinsic capability to perfectly adhere to surfaces, can follow the multiple deformations of the human skin and have unparalleled advantages like imperceptibility, low cost, and continuous use.[7]

In the medical field, the use of conformable electronics allows the replacement of stiff and bulky devices, which contrast with the soft body tissues and impose severe restrictions to the user comfort and mobility, limiting clinical applications and efficacy. The field of bioelectronics investigates electronic devices that operate as transducers between the signals and functions of biological systems and those of electronic processing systems. Through a myriad of options offered by chemistry, one can design organic electronic materials and devices that show the desired flexibility and conformability, the desired optical properties, and surface characteristics that promote biocompatibility and stability for long periods of time.[8]

So, in recent years, particular attention has been devoted to the development of devices able to monitor human physiological signals, through recording of the electrical activities of the human body (such as electrocardiography, ECG,[9] electromyography, EMG,[1,9] electroencephalography, and EEG,[10]), and measurements of biological parameters (such as pH, pressure, temperature and sweat composition).[11–16]

However, electronic skin devices can accomplish more than just sensing. Another class of conformable devices consists of active platforms, which find use in stimulation, instead of recording. These kinds of devices can apply a specific stimulus

Istituto Italiano di Tecnologia, Center for Materials Interfaces, Viale Rinaldo Piaggio 34, 56025 Pontedera, Pisa, Italy. E-mail: marco.carlotti@iit.it, virgilio.mattoli@iit.it

on the skin – which could be thermal, electrical, luminous, or mechanical – in order to activate the numerous receptors included in the dermis and epidermis. Such platforms can find diverse applications in different therapies, rehabilitation, personal thermal management technologies, bone remodeling, and human-robot interaction.[17–21] Because of their design and characteristics, these devices are particularly promising, in particular considering that one of their paramount characteristics is the limited invasiveness. Their use as smart and self-care treatments is particularly appealing, providing patients that require frequent treatments or suffer chronic disorders with a platform they can use at home or in their daily activities without the need to physically go to the hospital.

As well as this, they also open new channels of communication between machines and the human body. This concept is central in applications such as gaining control of prosthetic limbs: the stimulation of the users is a means to restore tactile feedback and give them the ability to better control the device and to react faster to external stimuli coming from the prosthetic limb, overall resulting in an improvement of the embodiment and acceptability of the prosthesis itself.

Finally, apart from biomedical applications, the capability to provide specific environmental sensation (*e.g.* tactile, kinesthetic, and temperature feedback), while still maintaining a high level of comfort, is of paramount importance in applications, which require virtual reality (VR), like for example remote control, immersive experiences, and gaming.[22]

While applications of recording physiological signals have already been widely reviewed,[23–26] less attention has been paid to devices able to generate a signal and interact directly with the human body. In this review we intend to report the conformable devices, which, according to us, are the most attractive and innovative in stimulating the human body, highlighting the functional materials that are employed, the fabrication techniques involved, and the design solutions that were proposed to improve their performance. In particular, we decided to focus our review on three main groups of stimulation devices, synthesized as follows: (i) devices able to provide thermal stimuli to the human body, (ii) devices providing electrical stimulation, and (iii) haptic displays aiming to restore tactile feedback. For completeness, in the last section, we also briefly examine innovative conformable photostimulation devices, a novel technological platform with a very promising future in healthcare and consumer electronics.

1.1. Requirements for conformable devices

1.1.1. Mechanical properties. The key factor driving research in conformable electronics is the possibility to deliver more comfortable, more efficient, and less invasive devices that can allow continuous, self-applicable, and on-demand monitoring and treatment. In general, human skin consists of a thin layered structure (50–600 μm) with a Young's modulus that can vary in the order of kPa–MPa, and an elastic behavior of up to about 15% overall.[27–29] The actual characteristics of the skin of an individual can, however, vary very much depending on the person (sex, age, and conditions) and on the region of the body we refer to (*e.g.* there can be up to 4 orders of magnitude difference in the Young's moduli of skin on the forearm and the forehead).[27,30]

Conformable electronics need to match these mechanical properties to ensure mechanical compatibility between the skin (or an internal tissue) and the device.[26,30–33] A failure to do so may result in a misfunction of the device, a shortening of its lifetime, or, even worse, tissue damage and infections.[33] This last point is particularly relevant in the case of stimulation. Hence, when designing a new device, one must bear in mind what its purpose is and how it is going to be applied to the user.[25] The limitations posed by the skin in terms of stiffness and stretchability may not apply to other kinds of tissues and organs and implantable devices for the latter may need different materials (*e.g.* because of their limited strain, brain and bone tissues can be addressed with metal electrodes).[34]

In recent years, researchers have come up with many ways to match the mechanical properties of the substrate with those of the electrical components and connectors.[24,31] A straightforward method is to use intrinsically stretchable polymers and composites.[35,36] These offer many combinations of softness and stretchability to meet the device needs and can perform as stretchable substrates for printing and deposition, encapsulation materials for more rigid stuff (*e.g.* regular silicon-based electronic components), or – in the case of opportune functional composites – as themselves.[32,33,37] Among this class of materials, silicone-based polymers (like polydimethylsiloxane (PDMS), Ecoflex, Dragon Skin, and Moldmax; Young's modulus: 1–7 MPa), poly(styrene-butadiene-styrene) (SBS; 2–33 MPa), poly(styrene-ethylene-butylene-styrene) (SEBS; 10–35 MPa), polyacrylates (7–35 MPa), and polyurethanes (PU, 3–51 MPa) are the more frequently used.[32,35] It is worth mentioning that, just like for the current stimulation technology available on the market – which relies on bulky instrumentation and wired, stand-alone electrodes[38] –devices obtained by encapsulation are often thick and require an additional adhesive to conform to the skin.[39]

Careful geometrical engineering can help in obtaining stretchable structures even from tough materials.[40,41] 2D (but also 3D) stretchable structures of metallic wires (as also rigid conductive polymers)[42] are common in epidermal electronics as they allow stretching of the structure as a whole without straining the material much. Serpentines and spring structures, often with fractal elements, are the most common, especially to prepare connectors.[24,37] The downside of this approach is that it is seldom applicable to the active components, limiting its scope. For the latter, a common methodology consists in preparing the desired element on pre-stressed substrates: if the former is thin enough (*i.e.* less than 5 μm), upon releasing it will buckle (without sensing much strain), and it can stretch again without compromising its performance.[41,43,44]

Just the limited thickness of thin structures allows them to bend without suffering the strain originating from the differences in length between the inner and the outer bending arches (which scales with the third power of the thickness for a film). Such principles can also support the fabrication of flexible devices from rigid materials that can work as conformable systems if they are not subjected to strain. This is the reason why it is common to find strong and tough polymers like polyimide (PI) and poly(ethylene naphtalate) (PEN) as substrates for thin skin electronics.[29,45] Moreover, these devices can perfectly conform and adhere to the skin thanks to van der Waals forces, without the need of any adhesive or contact medium. Among these systems, ultrathin-film devices (<100 nm thick) are particularly interesting. Because of the negligible difference in bending radii as we discussed above, and thanks to an incredibly large aspect ratio which favors the adhesion, such systems can perfectly adhere to any substrate, even rough ones.[5,46] This is very appealing for systems like the human skin whose elasticity is predominantly due to its fractal wrinkled structure. In particular, the ultra-thin film devices can elongate with the skin they are adhered to without cracking, no matter the Young's moduli of the materials, thus allowing the use of functional materials that would not be otherwise available because of their mechanical properties.[1,9]

1.1.2. Biocompatibility. Biocompatibility is of primary importance when fabricating conformable devices, since they have to make intimate contact with the skin for an extended amount of time. Adverse responses of the body – including dermatitis, hives, and other skin rashes or allergic reactions, even when mild, may hinder the functioning of the device by altering the chemical and mechanical properties of the skin-device interface. In more severe cases, these may also lead to inflammation, rejection reactions, and infections. Many of the most employed materials, such as PDMS and PI, have been certified biocompatible (although this declination is composition- and application-dependent).[47,48] Similarly, conjugated polymers, like poly(3,4-ethylenedioxythiophene) (PEDOT) and poly(3-hexylthiophene) (P3HT), and natural polymers are generally biocompatible.[42,49] However, in any case the addition of dopants and fillers to improve the properties of a material, such as the use of harmful solvents in the fabrication, may result in non-biocompatible devices from actually compatible starting materials.[36] One may find a clear example of this in PDMS and PI nanocomposites comprising silver nanostructures, which are widely used as conformable electrodes and stretchable connectors (see Section 2.1). While the matrix is biocompatible (as is bulk silver), the silver filler tends to oxidize and release cytotoxic silver(I) ions in the body environment.[50] Similar concerns involve the use of carbon nano-materials:[51] their case exemplarily highlights how, next to the importance of the chemical properties, the biocompatibility is affected by morphology, size, surface interactions, and domain distribution. In this sense, studies that involve novel materials and fabrication methodologies should investigate extensively their suitability – even beyond their proposed application – to better evaluate practical use and encourage technology transfer.

Finally, among the ensemble of biocompatible materials, one may also find some which are biodegradable. This definition brings together all those materials, which our body is capable of decomposing without any harmful consequence or side effect. Many flexible and stretchable polymers are considered biodegradable, like polysaccharides, poly(lactic acid) (PLA), poly(ethylene glycol) (PEG), and poly(vinyl alcohol) (PVA),[52] but also metals including iron, magnesium, zinc, and gold.[53] Devices obtained from these materials find applications as conformable implants or practical disposable electronics, which do not need to be removed.

1.1.3. Safety. Since conformable electronics should form an interface and interact with the human body, safety is a mandatory prerequisite in the development of such devices. In this review we decided to discuss mainly three categories of devices designed for delivering to the user some kind of stimulation: thermal – heaters – electrical, and mechanical stimulation. However, although mainly the electrical and thermal stimulation devices are those that can cause the most harm to the user, one should pay relevant attention to the safety of the user during the development of all the wearable technologies. Of course, the first fundamental issue concerning conformable devices is the biocompatibility of the materials, previously widely discussed. Immunologic and tissue biocompatibility represents a key consideration in particular for implantable devices where long-term stability of the system is crucial for the safety of implantation and the body's immune response.[27] However, the attachment of electronic devices to the skin may cause inflammation even if the materials used are biocompatible, as the long-term use of the system could prevent the skin transpiration. Therefore, epidermal electronics should not only guarantee biocompatibility of the materials but also skin breathability for long-term biocompatibility in order to use the devices with long-term treatment perspectives.[36]

Commercial skin-mountable devices, widely used in the clinical practice, generally consist of a metal electrode that can interface with the skin by different contact methods. In some cases, the electrode can directly adhere to the skin without any interposed layer (dry electrodes), while another possibility is to interpose a gel layer (wet electrode) or a dielectric layer (capacitive contact electrode) between the electrode and the skin.[54] However, the presence of interposed layers can avoid skin irritations and air gap formation, thus making the first configuration not as safe as the others. Air gaps indeed can degrade the electrical signal recorded[54] or, in the case of stimulation devices, they can generate hot spots and burn the user as well as damaging the electrode, especially in electrical stimulations where the current reaches several mA.[55,56] Since a non-uniform and inconsistent adhesion between the device and the skin during stimulation could damage human tissues, another key factor is the perfect contact between the former and the latter and therefore the ability of the device to conform well to non-planar surfaces.[57] Ultra-thin devices, such as dry electrodes transferred from tattoo paper, can perfectly adhere to all the imperfections of human skin, safely recording electrophysiological signals replacing standard electrodes.

Moreover, attention should be paid to the conditioning of electronic components connected to the skin-mountable device. An important issue is to remove, orreduce, the possibility of electrical leakage from the electronics to the body: leakage currents can pose a health risk to the user by causing severe burns, muscle spasm, or may damage the nervous system to the extent that they may provoke cardiac arrests and amnesia.[58]

One should consider that the general standard (IEC 60601-1) for medical electrical equipment imposes a maximum leakage current of 0.5 mA for patient safety.[59] Electrically safe devices could consist of capacitive structures made of an insulation layer that encapsulates the metal electrodes. This design ensures electrical safety in mounting the device directly on the skin but also for implantable electronics where the device has to be mounted on internal organ surfaces. In addition, the device so developed facilitates its own sterilization and cleaning operations, making possible its fast and safe reuse also for different subjects or in different trials.[59] The insulation layer also encapsulates the active parts of the system in order to prevent degradation of the electrodes and, next to this, to prevent possible consequences of malfunctions that could damage the user's tissues.

In recent years, the interest in wearable biochemical sensing has increased, leading to the development of multiple bioelectronic systems that allow epidermal electrochemical monitoring of a variety of metabolites and electrolytes. Enzyme-based devices, with applications like continuous glucose monitoring in diabetic patients, alarm against dangerous chemical threats, or even the detection of alcohol in sweat are just some of the multiple examples with such applications.[60,61] Concerning this latter application, different skin-worn alcohol monitoring platforms are integrated with an iontophoretic-biosensing system so as to locally stimulate sweat secretion by loading the iontophoretic electrodes with a sweat stimulant (for example, pilocarpine and carbachol). The Iontophoresis process consists of applying short-timed and localized current pulses to improve the drug diffusion through the skin. Hence, particular attention should be paid during the design and fabrication of the overall system so as to avoid possible skin burning and/or irritation caused by too high currents.[62]

Finally, another prerequisite for user safety is that wearable devices should work at a low driving voltage.[35] Some applications still need deeper analysis of this factor. For example, mechanical stimulation systems require a high voltage supply to work properly. This aspect makes it difficult to safely integrate them into skin mountable devices.[21] Other examples concern thermal stimulation: the heating module of heaters must not reach excessive temperatures. The designer should pay attention to developing a system capable of continuously remaining within a comfortable heating range for the user, in order to avoid sudden increases in temperature with consequent pain perception.[63] Heaters must also have uniform electrical and thermal properties on their active parts, maintaining electrical stability despite the deformations of the skin once mounted on it: heating inhomogeneity can lead to burns on the user's skin. Similarly, conformable electrodes used in electrical stimulation devices must have a homogeneous current density over the entire electrode; larger electrodes allow more conformable sensations.[64]

The purpose of this review is to provide an overview of conformable stimulation devices (Fig. 1), here classified as (i) devices capable of delivering thermal stimuli to the human body, (ii) devices that provide electrical stimulation,

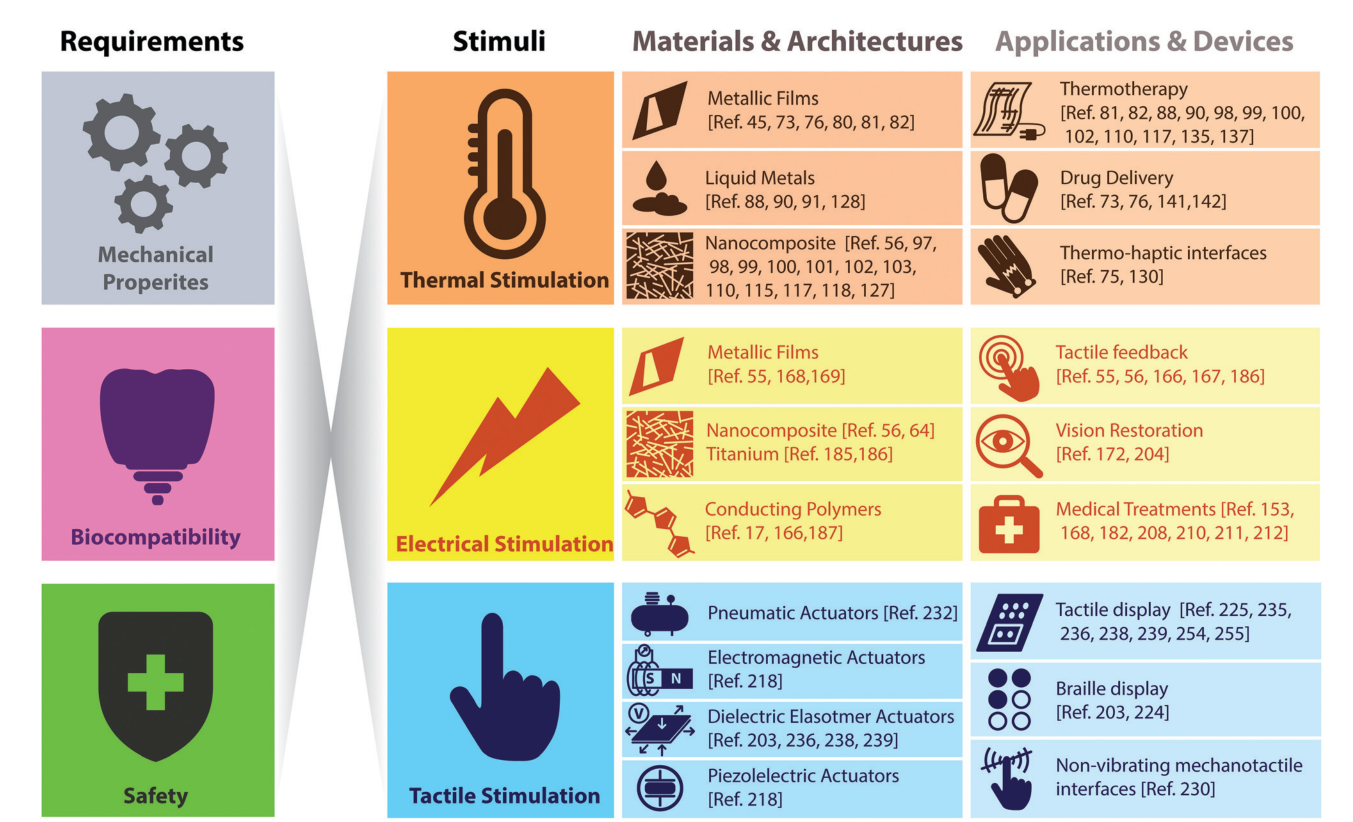

Fig. 1 Summary of the material solutions and device applications for diverse conformable stimulation technologies (thermic, electric, and haptic) covered in this review.

and (iii) tactile displays that aim to restore tactile feedback. In the last section, we also report recent studies about conformable photostimulation devices. In particular, we have only considered "conformable to the body" technologies, emphasizing both the enabling materials exploited for their construction and their working principles.

2. Conformable devices for thermal stimulation

Conformable thermal stimulation devices consist of conductive and often transparent sheets capable of heating or, in some cases, cooling the surface they are in contact with. Their main applications concern mainly transport and building industrial sectors where they find application, for example, as vehicle defrosters and smart heat-retaining windows or solar panels.[65–67] In recent years heaters have attracted increasing attention also in the medical field: the use of new materials and new fabrication techniques allowed the development of numerous conformable and stretchable heaters to directly mount onto human skin. Thermal stimulation is important in personal thermal management (PTM) technologies,[68] where one of the most important applications in this sense is in physical treatments and, in particular, in thermotherapy. The latter can be defined as the therapeutic application of heat to

Table 1 Summary of different properties of conductive materials used in conformable heating devices

System	Resistance/resistivity[a]	Resistance/resistivity at max. strain	Max. strain	Application	Ref.
PEDOT:PSS/rGO in PU	0.05 Ω cm		530%	Heating pad	17
Evaporated Au film (100 nm) on polyethylene naphtalate	188 Ωsq	—		E-skin heater and temperature sensor	45
3D Au@AgNW network in SBS	2.4×10^{-5} Ω cm	$\sim 3 \times 10^{-5}$ Ω cm	265%	E-skin heater, monitoring and stimulating mesh for hearts (implantable)	56
Evaporated Au (70 nm) on PI	550 Ω	630 Ω	Up to 30%	E-skin heater	73
P- and n-type Bi2Te3 thermoelectrics between electrodes	Peltier effect			Haptic device based on heating and cooling	75
Sputtered Zn/Mg/Ag (500 nm) on poly(glycerol sebacate)–poly(caprolactone) nanomesh			20%	E-skin heater for drug delivery	76
Electrodeposited metal films (Au, and Pt) on PI	1.2 Ω	—		E-skin heater and temperature sensor	80
Al(9 μm) on PET	17.2 Ω			E-skin heater and temperature sensor (simultaneous)	81
R2R continuous plasma-sputtered SiNx/Cu(12 nm)/[CNTs/polytetrafluor-oethylene]	5 Ωsq	30 Ωsq	Bending to 2 mm radius	Wearable heater	82
Galinstan/PDMS composite	8.7 Ω	9.1 Ω	>100%	Wearable heater, knee pad	88
EGaIn-Ni composite	2.2×10^{-2} Ω	2.4×10^{-2} Ω	5 mm bend	Tattoo heater	90
Oxidated EGaIn	2.9×10^{-5} Ω cm		180° bending angle	Hypertermia element (wireless inductive heating)	91
2D AgNW network (vacuum filtration) in PDMS	30 Ωsq	180 Ωsq	50%	Heating pad	97
2D AgNW network (vacuum filtration) on PI	<5 Ωsq	<5 Ωsq	450%	Kirigami heater for articulations	98
2D AgNF network (electrospun) in PDMS (or PET)	0.05 Ωsq	0.06 Ωsq	90% (PDMS) 10 mm bend radius (PET)	Wireless heating pad	99
2D CuZr network (evaporated on template) in PDMS	3.8 Ωsq	4.9 Ωsq	70%	Heating pad	100
2D AgNW network in PDMS	5.4 Ωsq	~40 Ωsq	200%	Heating pad	101
3D AgNW network in SBS	8.3×10^{-5} Ω cm	$\sim 10^{-4}$ Ω cm	100%	E-skin for articular thermotherapy	102
2D AgNW network (spun coat) between PDMS/PVA	3.7 Ωsq	3.7 Ωsq	1 mm bend radius	Heating pad	103
2D CuNW (spray coated) between PET/PMMA	17 Ωsq	17 Ωsq	3 mm bending radius	Heating pad	110
SWCNT in PVA	475 Ωsq	475 Ωsq	20 mm bend radius	>10 μm-thin heating pad	115
Sintered AgNP on medical tape	12×10^{-6} Ωsq	13×10^{-6} Ωsq	8 mm bend radius	3D-printed heating pad	117
AgNW on PU sponge	0.05 Ω cm		3D compression-extension	Heating pad	118
Au@SiO_2 in alginate-chitosan matrix	Photothermal		Ultraconformable	Photothermal treatment of tumors	127
Mg in GaIn alloy	Photothermal		Bendable	Photothermal treatment of skin tumors	128
p- And n-type Bi2Te3 thermoelectrics between Cu electrodes	Peltier effect		230%	Heating and cooling pad	130

[a] Resistance is used where the studies did not specify any form factors of the resistive elements.

specific regions of the body resulting in increased temperature in the tissue,[69] often useful for the treatment of musculoskeletal disorders. The heating of body surfaces provides pain relief in wrist, shoulder, or lower back injuries,[69,70] and alleviates disturbances from arthritis, stiff muscles, injuries to deep skin tissue, and joint injuries due to aging or obesity.[17,71] Heat also increases the distensibility and flexibility of soft tissue: physiotherapy can exploit heaters to reduce injuries in sports.[72] Furthermore, heaters find their use also in controlled drug delivery: in this sense, one can control the transdermal diffusion rate of therapeutic drugs – loaded on micro/nano-particles and embedded in conformable devices – by controlling the temperature.[73] Since thermal stimulation allows discrimination between materials with different thermal properties, heaters integrated into haptic interfaces can activate thermoreceptors of the user, transferring cold and hot sensations useful to recognize different virtual objects (*e.g.* touching wood will give a different thermal sensation than touching glass). This impacts the sense of immersion in applications concerning VR.[74] Surely, thermal displays could help blind people to avoid obstacles while walking or interacting with objects in their surroundings, increasing their level of independence.[75]

2.1 Materials

Conformable heaters usually take advantage of Joule heating, thus making the relatively high resistance of many flexible conductors (usually between 10^{-1} and 10^{2} Ω)[11,35] a strength point. One should pay attention to heat dissipation and the working voltage, as uncontrolled overheating might damage both the component and the host matrix.[36,45] As we will discuss in this section, many materials have been proposed as active elements in heating solutions. However, here we decided to focus only on those studies, which made explicit reference to the use of their devices as body-conformable heaters (see Table 1), and therefore also addressed (at least partially) issues like biocompatibility and stability.[36] As the reader will see, the latter only covers a small sub-group of all the flexible heaters, leaving a lot of ideas for improvements in future research.

The simplest, go-to solution for many e-skin systems consists of metallic wires.[29] Usually gold and other noble metals are employed (see Table 1), yet for many applications it would be possible to even use bioabsorbable metals that have higher resistivity and that can be safely degraded by our bodies, like magnesium and zinc.[76–78] Masked evaporation of thin layers of metal on a flexible substrate (*e.g.* polyimide (PI), polyurethane (PU), and polydimethylsiloxane (PDMS) are the most common) yields precise features that can work as connectors, and electrodes, as well as a part of a large selection of components.[45,79] The possibility of obtaining entire circuits with few evaporation steps is very appealing, especially if the aim is to converge many functions in the same device. Next to metal evaporation, electrodeposition, cut-and-paste methodologies, sputtering, and roll-to-roll printing are also widely used techniques, but the level of resolution is much lower.[4,80–82]

Since the resistance increases as the metallic feature shrinks in size, it is no surprise that serpentines realized with thin lines of metal have been widely used from the beginning of epidermal electronics to act as local heaters.[29,80,81] However, such elements are usually small and not able to heat large parts of the human body as often required in thermotherapy. This is because the mechanical properties of the metals – which do not match with those of the human skin or the flexible substrates – allow only limited freedom of movement and thus an increment of the area would result in an increased chance of failure.[20] As the lines in a serpentine need to be dense and tightly packed (in order to provide uniform heat), one can only adopt limited geometric solutions to improve their stress resistance (*e.g.* an array of elements can be a solution to cover larger areas).[73]

Liquid metal (LM) alloys have similar conductive characteristics to metals, but are extremely more flexible thanks to their liquid-like nature.[83] In this regard, the most used materials are Ga-based alloys, like Ga–In at their eutectic composition (EGaIn) or ternary alloys of Ga–In–Sn (Galinstan), as they offer advantages like low toxicity,[84] negligible vapor tension, and can be printed even in 3D architectures, thanks to their intrinsic non-Newtonian behavior.[83,85] Because of these reasons, LM can have a large number of applications in soft electronics.[86,87] Anyhow, one can only find limited use of such materials in fabricating conformable heaters, perhaps as a result of the limited resolutions achievable compared to the previously mentioned lithographic techniques, or because their high conductance makes them more useful to prepare other circuitry elements.[86] We will discuss LM again in this review (Sections 3.1 and 5). Concerning conformable heaters, Wang *et al.* prepared a Galinstan-uncured PDMS composite ink that they could use to write thin lines by direct ink writing (Fig. 2a).[88] By successive encapsulation and polymerization, they printed serpentines that could act as heaters with a minimal change in performance even when stretched to more than 100% of their original length. Compared to pure LM, the composite limited the flow of the LM thus increasing drastically the stability of the device.[89] Guo *et al.* took a different approach for the fabrication of their devices and realized electronic tattoos based on the selective adhesion on polymethylacrylate (PMA) glue of a semi-liquid EGaIn–Ni composite:[90] in their study, they used a skin-friendly PMA glue to draw different designs on skin and then applied the conductive composite on them with a roller (Fig. 2b). Conversely, Wang *et al.* modified the adhesion properties of EGaIn by thorough oxidation.[91] After stirring in air, the adhesion between the LM and the skin increased drastically and the authors were able to write circuits directly on skin using a brush. They also demonstrated that these structures could be used for wireless thermotherapy by inducing a current with an alternating magnetic field.

As well as the opportunities offered by bare metals, (nano-)composite materials also combine an adequate compromise between conductivity and flexibility by relying on percolation networks of conductive fillers in polymer matrices.[33,92] Their resistivity is usually higher than that of the aforementioned systems[93] and thus, as introduced before, it is no surprise that in the literature we can find countless examples of their use as heaters, utilizing both metallic and carbon-based nanomaterials (*e.g.* nanoparticles, nanowires, nanofibers, and flakes).

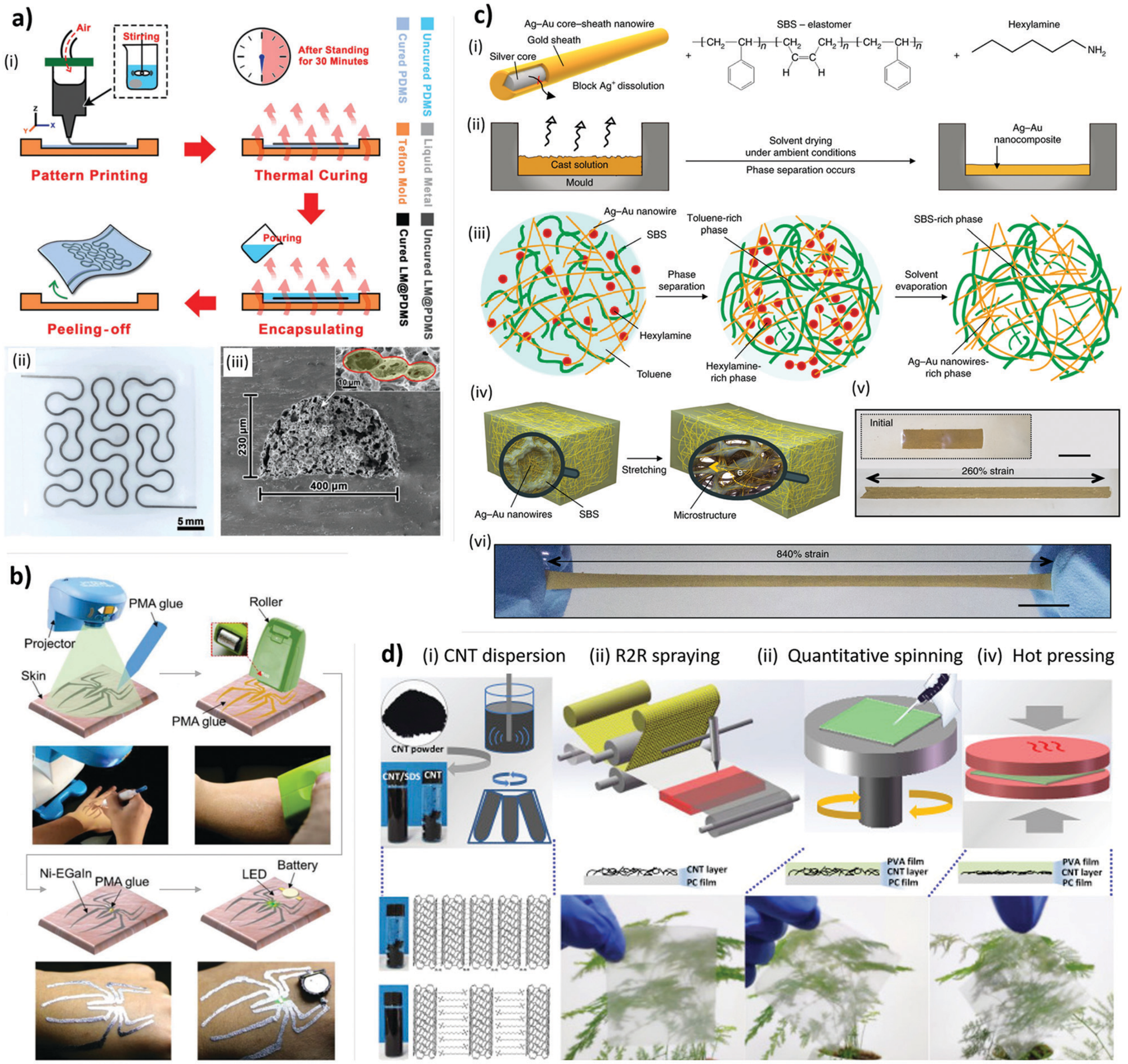

Fig. 2 Materials and fabrication methods employed for the development of conformable heaters. (a) (i) Fabrication procedure for preparing LM@PDMS stretchable, and wearable heaters and (ii) a prepared sample with a complex fractal structure. (iii) Cross-sectional SEM image of LM@PDMS with its internal microstructure. Reproduced from ref. 88 with permission from John Wiley & Sons, DOI: 10.1002/admt.201800435. (b) Fabrication of the Ni-EGaIn electronic tattoo: schematic diagrams and corresponding optical photographs of the Ni-EGaIn electronic tattoo printing procedure. Reproduced from ref. 90 with permission from John Wiley & Sons, DOI: 10.1002/admt.201900183. (c) (i) Ag–Au nanocomposite made by combining a mixture of Ag@Au NWs, SBS, and hexylamine in toluene. (ii) Solvent-drying process under ambient conditions. (iii) Schematic illustration of the initial solution (left) separated into a Ag@Au NW-rich phase and a SBS-rich phase during dry-casting (middle). Subsequent solvent evaporation (right) forms the microstructured Ag–Au nanowire nanocomposite. Red dots, hexylamine; yellow wires, Ag@Au nanowires; green wires, SBS. (iv) Schematic illustration of the microstructured NWs-SBS nanocomposite before and after stretching. (v and vi) Optical camera images of the nanocomposite before (v, inset shows before stretching) and after (vi) the heat rolling-pressed nanocomposite sandwiched between the elastomeric substrates (VHB film). Scale bars, 10 mm. Reproduced from ref. 56 with permission from Springer Nature, DOI: 10.1038/s41565-018-0226-8. (d) Illustration of the fabrication process for PVA/CNTs-HP film: (i) CNT dispersion and corresponding optical image; (ii) R2R spraying method for pre-constructing CNT network on PC surface; (iii) quantitative spinning process to introduce ultrathin PVA layer; (iv) hot pressing treatment to prepare PVA/CNTs-HP film. Reproduced from ref. 115 with permission of Elsevier, DOI: 10.1016/j.compscitech.2019.107796.

For application as conformable electronics, it is worth mentioning that the devices do not necessarily need to have characteristics like ultra-high strain and high transmittance which are appealing for other applications as soft-robotics and smart-architecture.[21,94–96]

One of the most common methodologies to prepare flexible conductive composites is to make use of percolation networks of metallic nanowires (NWs) or nanofibers (NFs) on flexible, and biocompatible substrates. Compared to nanoparticles, the use of these nanostructures results in a lower percolation threshold and thus they influence to a smaller extent the mechanical properties of the matrix.[92] There are several methodologies to achieve such networks: (i) vacuum filtration of dilute NW solutions;[97,98] (ii) electrospinning;[99] (iii) deposition on a sacrificial 2D-network;[100] (iv) drop casting;[101] (v) casting of composite solutions.[56,102] In particular in this last case, the ligands on the NW surface covers an extremely important role to achieve a good dispersion: Choi *et al.*, for example, by carefully tuning the amount of hexylamine to function as the ligand of Au-coated AgNWs in poly(styrene-butadiene-styrene) (SBS), their composite could reach a remarkable 840% stretch strain without sacrificing the conductance of their composite (Fig. 2c).[56] As we show in the entries in Table 2, the most used metal for this kind of device is silver because it combines good electrical properties with simple procedures to prepare long NWs.[56,102] Yet, the resulting networks are prone to the formation of a surface oxide, which hinders the contact between different wires upon oxidation.[97] One may overcome this issue by coating the NWs with Au (which does not oxidize),[56,98] by embedding them in a suitable material,[103] or by adding carbon-based nanomaterials (*e.g.* graphene or carbon nanotubes (CNT), which are reported to increase the stability to mechanical stress and oxidation, in addition to being better heat conductors).[104–106] Coating the Ag NW network with a conductive polymer was reported to increase the thermal properties of a heater (but not the resistivity).[107,108] A valid alternative to silver to prepare similar structures is copper,[109] which has a lower price but is also a worse conductor.[110] Another possibility is to use nanostructures of a different material altogether.[95,100,111–114] For example, recently Zhou *et al.* realized a conformable heating pad by hot pressing a dispersion of CNT in PVA formed with the help of sodium dodecylsulfate as a surfactant (Fig. 2d).[115] Remarkably, the preparation of this composite involved only water-processable materials.

While many groups employ NWs because the resulting composites are highly transparent,[116] Moon *et al.* used a conductive ink containing Ag nanoparticles that they sintered by adding NaCl to prepare a bendable composite that they could print on different substrates;[117] with a different approach, Weng *et al.* realized a conductive 3D sponge by self-assembly of Ag NWs after surface treatment with dopamine: the resulting material could not only stretch and bend but also be compressed and could work as a heater, as well among many other applications.[118]

Other interesting candidates for conformable heaters are intrinsically conductive polymers (CPs).[96,119] Polymers like polypyrrole, polyaniline, and PEDOT:PSS possess good conductivity,[1] are considered biocompatible,[42] and can be patterned easily to make circuits.[5,120,121] The latter polymer in particular attracted much attention because it is water-processable and for its chemical and thermal stability. The main drawback of these materials is their intrinsic rigidity. Pristine PEDOT:PSS, for example, can stretch only about 5% before cracking, limiting its appeal for highly stretchable electronics applications. However, as mentioned before, conformable electronic devices do not need to sustain high strains, and thus one may adopt many solutions to modify both its mechanical and electric properties.[122–124] Conversely, our group showed the use of thin films of PEDOT:PSS with a thickness of a few hundred nanometers as transferable ultra-conformable electrodes for electrophysiology;[1,9] ultra-thin films of rigid materials can show extreme flexibility as bending motions produce only negligible strain (*i.e.* the difference between the inner and outer bending radii). Such devices can adhere perfectly to the skin thanks to van der Waals forces, therefore making them an attractive platform for other skin electronics applications.[125] In any case, despite the known Joule heating properties of PEDOT:PSS,[96,124,126] and its use as a coating for fabrics to be used in heated clothing (we will not address these devices here as Lu *et al.* covered them in detail in a recent review,[68]), one can only find limited examples of CP-based conformable heaters in the literature. Zhou *et al.* developed a conductive composite of PEDOT:PSS and water-borne PU with remarkable stretchability; the addition of *in situ* generated rGO (GO reduction with HI), while improving only slightly the electrical conductivity, increased dramatically the thermal conductivity of the films.[17]

While all the examples introduced so far have to do with Joule heating, there are more mechanisms capable of generating heat. Our group, for example, investigated the photothermal performance of a submicrometric ultraconformal polysaccharide

Table 2 Summary of different properties of conductive materials and electrodes employed in conformable electrostimulation devices and their applications

Electrode materials	Circuit materials	Impedance	Applications	Ref.
Au (evaporated)	Au evaporated on PI	205 kΩ (20 Hz)	Electrotactile stim., electrical muscular stimulation	55
Au@AgNW network in SBS	Au@AgNW-SBS composite in SBS	1.6 kΩ (1 Hz) 0.7 kΩ (10^5 Hz)	ECG, EMG. FES of cardiac muscles	56
Graphene-SEBS composite	Graphene-SEBS composite	12 kΩ (10^3 Hz) 4 kΩ (10^6 Hz)	ECG, EMG. Electrical stimulation	64
Ag/AgCl conductive ink	PEDOT:PSS	—	Electrotactile stimulation	166
Au (evaporated)	AgNW-SBS composite	200 Ω (1 Hz) 20 Ω (10^5 Hz)	FES of cardiac muscles	168
Au (evaporated)	Au evaporated on LCP (Vecstar)	—	Stimulation of bone cell regrowth	169
Ti microneedle	Ti		Eelctrotactile display, data transfer	185 and 186
PEDOT:PSS-*b*-PPEGMEA	PEDOT:PSS-*b*-PPEGMEA	2×10^2 kΩ (0.1 Hz) 0.1 kΩ (10^5 Hz)	Electrotactile element (surface roughness sensations)	187

film containing Au@SiO_2 particles, which produced a dramatic temperature increase when exposed to an NIR laser.[127] Wang *et al.* used a composite of magnesium and Ga–In alloy to produce LM patches with similar properties.[128] Remarkably, Lee *et al.* used p- and n-doped Bi_2Te_3 samples sandwiched between Cu electrodes incorporated in Ecoflex to prepare a series of Peltier elements,[129] which are not only capable of acting as heaters but also as cooling pads just by reversing the applied bias.[130] Kim *et al.* fabricated a device based on the same technology and showed its possible use as a haptic display for guidance of blind users.[75] While these devices are still relatively thick compared to those presented above, the possibility of generating both hot and cold sensation on the skin by the same device is incredibly fascinating. A possible solution to make thinner devices could arrive from organic materials, which, despite having the worst thermoelectric performance achieved so far, possess several advantages concerning fabrication procedures (*e.g.* they can be printed, they are flexible, and they are obtained from abundant resources).[131]

2.2 Devices and applications

Most of the thermal stimulation devices find use in industrial applications, but in recent years they have also become relevant in the medical field. One of the main characteristics required from electrical heaters both for industrial and medical applications is low sheet resistivity in order to obtain low-voltage and power-efficient devices, according to Joule's law. However, medical heaters require some additional attention. For these specific applications (i) biocompatibility of the materials is fundamental so that the device mounted on human skin does not cause damage to the user or skin irritations also in long-term treatment perspectives; (ii) the heating module must not reach too high temperatures: the temperature of the human skin should remain between the safety thresholds of 15 and 45 °C to have comfortable sensations, otherwise nociceptors respond to thermal stimuli with consequent pain for the user;[132] (iii) heaters should be flexible and stretchable in order to follow dynamic strains of the skin, and stable from a mechanical point of view: the inability to conform well leads to a non-uniform and inconsistent adhesion between the heater and the skin. This aspect could cause air voids that generate hot spots[57] and could burn the user if the heater is operating near the 45 °C safety threshold; (iv) heaters must have uniform electrical and thermal properties: heating inhomogeneity can lead to burns on the user's skin;[101] and, finally, to reach the objective of using heaters in healthcare and in thermotherapy two main key issues are (v) their portability and (vi) the possibility to adjust heating temperature in real-time. The latter is important also to avoid the aforementioned overshoots.

2.2.1 Thermotherapy. Different studies, already widely described in the previous section, proved the electrical stability of conformable medical heaters once attached on non-planar surfaces – *i.e.* the human body. These devices should endure deformations of the joints to perform articular thermotherapy: Choi *et al.* and Wang *et al.* demonstrated how resistance and Joule-heating characteristics of the developed heaters remained stable under different strains and motion, respectively, once attached on a volunteer's wrist and knee.[88,102] In the latter case, the authors embedded the heater in a kneepad so to improve its electrical stability, while doing sports.[88] Moreover, Kang *et al.* demonstrated how their A4-size heater mounted on the user's arm can reach electrical stability:[82] larger films are useful in thermotherapy if there is the need to treat a larger area of the body. Heaters can become portable thermotherapy devices because their low working voltage allows using batteries for power supply: the Cu NW-based stretchable heater developed by Zhai *et al.* can elevate its temperature up to 50 °C with 3 V DC bias and the heater fabricated by Choi *et al.* made of Ag-NWs on thermoplastic elastomer reaches 40 °C by applying 1 V.[102,110]

However, most of the studies rely only on the – experimentally obtained – relationship between the applied voltage and the corresponding reached temperature to control the heating level of the device.[82,103–105,110,121,133] However, this method does not take into account the inter-subject variability of heat transfer conditions and provoked sensations, an issue of primary importance for the healthcare market. For example, changes in blood flow can cause changes in skin temperature differently from person to person. Hence, different studies aimed to achieve more precise control over the heat generated and the temperature reached. In some stretchable heaters, which also included temperature sensors, the sensing element was the same as the heating element (usually a metallic serpentine).[73,134] Such a design does not allow one to measure the temperature and to generate heat simultaneously; also, it risks measuring its own temperature instead of the actual skin one. Stier *et al.* proposed instead a stretchable aluminum tattoo-like heater on a PET substrate, which also included a gold resistance temperature sensor, separated from the heater by an insulation layer in order to record skin temperature.[81] The overall device included a proportional-integrative-derivative (PID) control so as to have a real-time temperature feedback. PID control allowed us to maintain a target temperature on human skin over a specific time interval, 38.5 °C for 30 minutes in their experiment, automatically adjusting the DC power applied on the heater (Fig. 3a).

The integration of the heater with electronic circuitries has been another step forward in reaching both the objective of portability and real-time wireless temperature control. An *et al.*, for example, reported an electronic band accompanying the heater, which involved different electronic components such as a Bluetooth module to communicate wirelessly with a smartphone from which the user could both decide the heating mode and control temperature up to 50 °C.[100] The microcontroller unit (MCU) embedded in the band received the signal and adjusted the voltage, which determined the amount of electric current flowing through the heater according to Joule's law. With such a device, however, the user shall rely only on their own thermal cues to regulate the heating mode. Cheng *et al.* proposed to use a digital temperature sensor in addition to the electrical components considered in the previous example.[135] The MCU transmitted wirelessly the value recorded by the sensor to a smartphone so as to show the real value on the display. In this way, the user could decide to adjust the output DC voltage, consequently regulating the temperature as

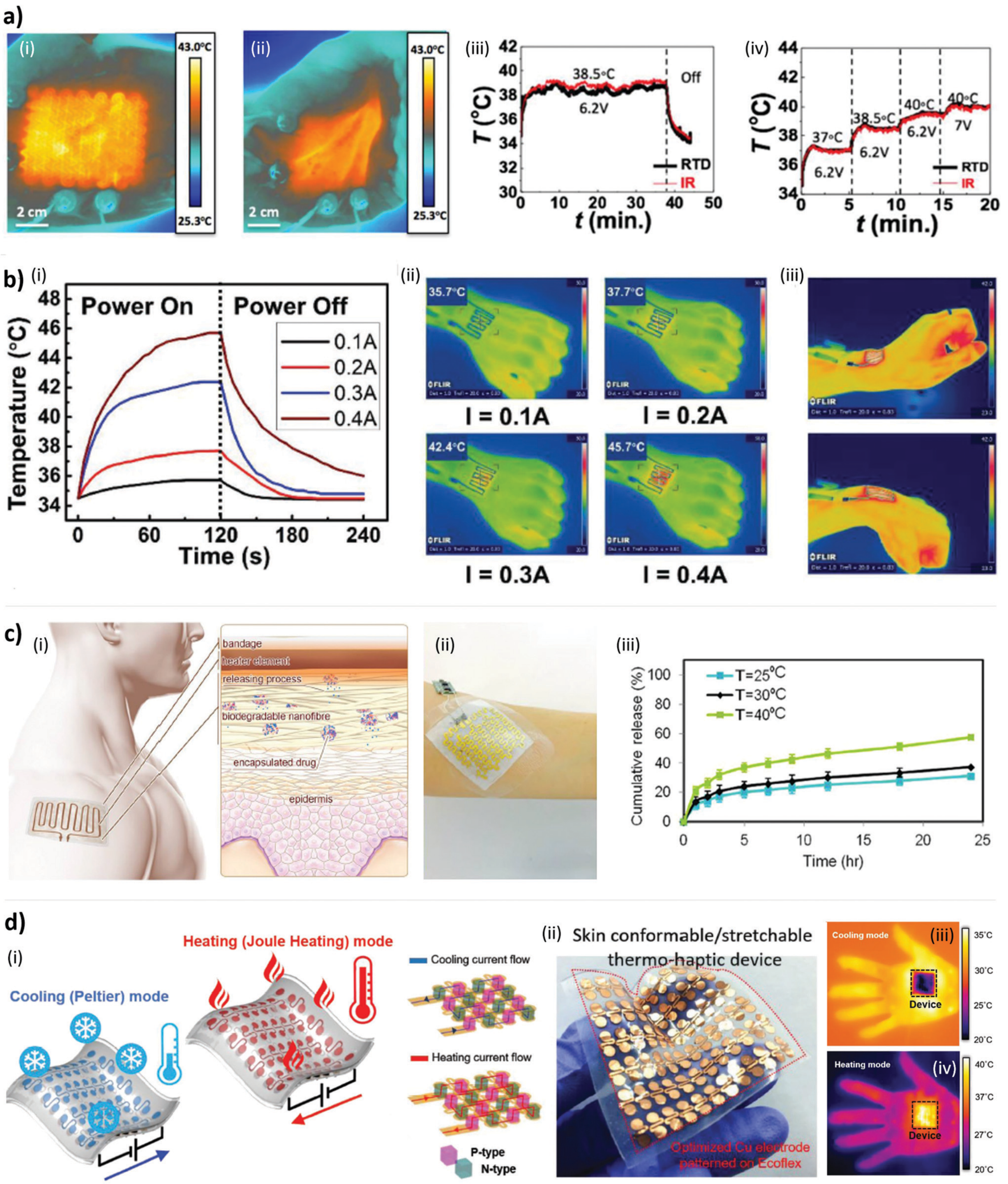

Fig. 3 Functioning principles and performance of conformable heaters. (a) Heater conforms to the hand during (i) opening and (ii) closing without changing its temperature. (iii) Temperature stability with time and (iv) voltage steps. Reproduced from ref. 81 under Open Access CC BY Licence, DOI: 10.3390/mi9040170. (b) Characterization of Ni-EGaIn Joule heating electronic tattoo directly printed on the skin at (i and ii) different voltages and (iii) hand movements. Reproduced from ref. 90 with permission from John Wiley & Sons, DOI: 10.1002/admt.201900183. (c) Thermo-responsive drug nanocarriers released upon temperature increase induced by the integrated flexible heater. (i) Schematic of working principle. (ii) Photograph of the thermal patch. (iii) Temperature response at different voltages. Reproduced from ref. 76 under Open Access CC BY 4.0 Licence, DOI: 10.1038/s41598-017-04749-8. (d) Skin-mountable Peltier-based thermo-haptic device with cooling and heating modes. (i) Working principles. (ii) Photograph of the device. (iii) Thermal characterization. Reproduced from ref. 130 with permission from John Wiley & Sons, DOI: 10.1002/adfm.201909171.

needed. In this study, the heater, MCU, and the temperature sensor were integrated into the heating fabric and clothes so as to be almost not visible but still in contact with the skin. Jang *et al.* demonstrated that the same set-up was used by Cheng *et al.* (same components, with a thermistor as a temperature sensor) can be integrated on a band instead of clothes, depending on the needs.[99]

Another interesting field of research in medical treatments is personalized care, which aims to develop targeted therapies for each individual patient.[136] In order to obtain custom devices, Moon *et al.* proposed to use a 3D scanner to acquire body information, which enabled them to print very personalized heaters.[117] In particular, data acquired from a 3D scan could help in deciding the dimension and shape of the custom heater. The user could select different points on the scan surface and the software calculated the length and width of the single 2D heaters but also how to stitch them together in order to apply the entire film on the interested body surface. In this way, the final heater can have various shapes and sizes depending on the injury or medical condition of the patient. However, low-cost fabrication techniques are necessary to prepare heaters for disposable use. Guo *et al.* exploited a very low-cost fabrication technique in producing their tattoo-like heater: the device could be directly fabricated on the user skin with any shape and size, as already explained in Section 2.1, and as shown in Fig. 3b.[90] Another cost-effective method for heater fabrication is the kirigami technique – the oriental art of cutting and folding paper, used to cut the considered substrates in order to obtain more complex geometries, often with higher stretchability.[98,137]

2.2.2 Drug delivery. In general, in the medical field, treatments for chronic and deep wounds consist in a systemic administration of high doses of antibiotics, resulting however in side effects such as drug resistance or toxicity.[76,138] Transdermal drug delivery offers advantages in this field because of its non-invasiveness in drug administration, also avoiding the first-pass effects of the liver.[139] For these reasons, numerous studies concerning transdermal drug delivery came to the conclusion that the poor permeability of human skin severely limits this application.[140] Park *et al.* demonstrated how the permeability of human skin can improve by locally heating the surface, leading to the possibility to increase transdermal flux of drugs.[139] Hence, in recent years different studies proposed conformable and flexible platforms to mount on the skin including a heating module, aiming to transdermally deliver drugs using thermal stimuli. These devices allowed continuous control of drug delivery rate with specific temporal patterns. In these applications flexible heaters resulted in contact with thermo-responsive drug micro-carriers encapsulated within a substrate. Different groups used PEGylated-chitosan,[76] poly N-isopropylacrylamide-*co*-acrylic acid (PNIPAM),[141] or meso-porous silica (m-silica) particles[73] to encapsulate drugs and, respectively, poly(glycerol sebacate)–poly(caprolactone) (PGS–PCL) nanofibers, hydrogels or hydro-colloids as substrate layers. In particular, the heat generated by the heater degraded the physical bonding between the particles and the drugs and pharmacological agents loaded in the micro-particles diffused transdermally. Tamayol *et al.* demonstrated that drug release rate is proportional to the voltage applied on the heater.[76] Fig. 3c shows the thermo-responsive device developed for drug delivery and the effect of the heat generated, which resulted in an increasing release of the drug, proportional to the reached temperature. The electroresistive heater proposed by Son *et al.* is part of a more complex system, which also includes a strain sensor, a temperature sensor, and resistive random access memory (RRAM) – *i.e.* the resistance of a thin film of electrically switchable material, generally made of oxides, that can assume different physical states depending on the electrical excitation.[73] In this way they developed a multifunctional healthcare system able to monitor the health state of the patient, to store data in the RRAM, and to deliver the corresponding feedback therapy, thanks to the nanoparticle-assisted drug delivery. Lee *et al.* proposed a sweat-based glucose monitoring device for the transdermal thermally-controlled drug delivery in diabetic patients.[142] In particular, the device mounted multiple conformable sensors (humidity, glucose, and pH sensors) and a three-channel stretchable heater with integrated microneedles, which contained the drugs. Thermal activation in response to the glucose level measured from sweat, allowed the dispersion of the drug. Electrically controllable and conformable heaters can pave the way for on-demand and localized drug delivery.

2.2.3 Thermo-haptic interfaces. One of the major applications of heaters is in haptic interfaces, where thermal characteristics of the object can be reproduced to assist users in recognizing a virtual object in virtual environments or a remote object handled by teleoperated robotic systems. This led to development of a large number of thermal displays.[132,143] The number of cold receptors in human skin is higher than the number of heat receptors; hence, a thermo-haptic device shall have both heating and cooling functionalities to deliver more information and to restore true artificial thermal sensations, otherwise the only heating function is limited to a half function of the artificial thermal feeling.[130,143] The most common type of thermoelectric generators are Peltier cells, which generally consist of many thermocouples (or p–n junctions), electrically and thermally connected (in series and in parallel, respectively), inserted between two electrical insulators. The injection of current heats up one plate of the device, while cooling down the opposite one, depending on the relative direction between the current and the p–n junctions.[144] These kinds of devices pave the way for another important role of thermal stimulation: material recognition based on thermal cues. Recently Lee *et al.* proposed a bi-functional thermo-haptic device, already explained in Section 2.1 capable of both heating up (Joule heating) and cooling down (Peltier effect) the body surface, depending on the virtual object the user is in contact with.[130] The device consisted of alternating p- and n-type thermoelectric pellets that induced a rapid thermal response when voltage was applied (Fig. 3d): thanks to the Peltier effect, heat was generated where carriers were highly concentrated, cooling down the other side of the junction. Serpentine-shaped copper electrodes with their highly deformable structure and low electrical resistance connected each pellet. The authors demonstrated that the device could reproduce several thermal properties of objects in virtual reality with a good level of accuracy and wearability, showing that virtual sensations, mainly related now to tactile feedback, could be further augmented with thermal information. More recently, Kim *et al.* developed a two-dimensional thermo-haptic device capable of transferring tactile information to thermally stimulate thermoreceptors placed in the human skin.[75] The 2D array consisted of unit cells each composed of thermoelectric couples of n-type and p-type elements, in order to both warm up and cool down the whole device. An H-bridge inserted in each cell allowed the direction of the current flow to be changed or to bypass the cell, so as to let the thermoelectric unit cell work as a heater, as a cooler, or to turn off both. Moreover, one could access a specific cell so as to control independently its temperature through an active matrix addressing method based on a field-effect transistor switch, activated through the column and row selectors. Also, the temperature of each unit cell could be controlled adjusting the amount of current applied to it. The authors demonstrated how the thermal haptic module so implemented could help blind people if embedded on the handle of their cane. Such a device could indeed give information to the user about the presence of an obstacle as well as its size and shape, using additional electrical components like a depth sensor and a microcomputing unit to control the unit cells of the device, in order to avoid obstacles in a more reliable way. For example, the temperature of the module cooled down proportionally to the distance between the user and the obstacle, informing the user of the actual distance only with thermoception.[75]

3. Conformable devices for electrical stimulation

Nowadays different fields such as rehabilitation or physiotherapy[38,145] but also virtual reality (VR)-based applications[22] often exploit electrical stimulation of the peripheral or central nervous system. However, electrical stimulation and the effects it provides on humans are still a wide field of research. Some examples concern how an electric field can modify cell behavior, including orientation, proliferation, cell migration, and differentiation,[146,147] which are the best stimulation parameters in terms of current amplitude, pulse width, and frequency of the applied electrical signal and the best electrode placement to achieve optimal results in electrical stimulation therapy (ETS);[148] how haptic perception evoked, for example, on the hand by applying transcutaneous electrical nerve stimulation (TENS) can vary in terms of the region perceived and amplitude of the sensation, varying stimulation parameters;[149] the optimal current or voltage waveform to apply. Currently, the most used waveforms for the different applications of electrical stimulation are the sinusoidal, triangular, and mono/biphasic square waveform.[64,150,151] Moreover, in recent years, different studies aimed to restore tactile and pain feedback in amputees by electrically stimulating the subject by applying short current pulses, developing a so-called neuromorphic model, which attempts to mimic the behavior of the nervous system, with the aim of providing more realistic sensations.[152]

Conformability of the device is a central issue in medical applications: different studies propose conformable electrodes that perfectly adhere to soft tissues of the body and are capable of injecting a small amplitude of current flow. In the medical field, such devices find their use in the regeneration of neural networks, in the electrically induced differentiation of osteoblasts and therefore in bone remodeling, or in the electrical stimulation of the heart for smart defibrillation.[20,153] Moreover, as already mentioned, electrical stimulating devices are used in rehabilitation and sports medicine for neuromuscular electrical stimulation (NMES) and functional electrical stimulation (FES). NMES consists of the repeated application of current to produce contraction of innervated muscle by depolarizing local motor nerves producing effects that enhance functions such as muscle strengthening but that do not provide a functional activity. FES, instead, exploits electrical currents to enable a functional activity in order to replace a completely lost movement in patients with neurological impairments, activating the sensory-motor system, allowing the control of walking or grasping in stroke patients, or, also, for tremor suppression.[18,148] Furthermore, experiments have been conducted to demonstrate how targeted spinal cord stimulation could re-establish adaptive control of paralyzed muscles during overground walking in patients with spinal cord injuries.[19] Other applications for stimulating devices regarding human-robot interaction, where artificial components and the user have to form interface with each other.[21] In particular, if a robot is equipped with some kind of sensor, it will "sense" different stimuli from the environment. The sensed signal could then be transferred to the user, stimulated by the wearable conformal device. This concept is central in applications such as prosthetic limbs, where stimulating the user is a means to restore tactile feedback. In this way the users will better control the prosthetic device and they will react faster to external stimuli coming toward the artificial limb, as well as enhancing the embodiment and acceptability of the prosthesis itself. Furthermore, TENS is a treatment generally used against painful conditions, including chronic neuropathic pain,[154] as well as a method that is being explored to restore tactile feedback on amputees.[152] When designing devices and, in particular, electrodes for superficial – *i.e.* non-invasive electrical stimulation of the human body, several factors must be considered, such as skin impedance, perfect contact between the electrode and the skin, and the particular frequencies and current amplitude to apply in order to deliver comfortable stimulation.

3.1 Materials

Electro-dermal stimulation requires the electrodes to be in intimate contact with the skin in order to deliver a reliable and effective signal. At the same time, the circuitry must be isolated from it so that the signals can reach only the desired points. For these reasons, the majority of electrostimulation devices available today consist of a set of wired electrodes that one can apply on the interested areas *via* adhesive pads and gels designed to lower the impedance of the contact.[155–161]

For conformable devices, these limitations translate to a higher complexity in the fabrication processes when compared

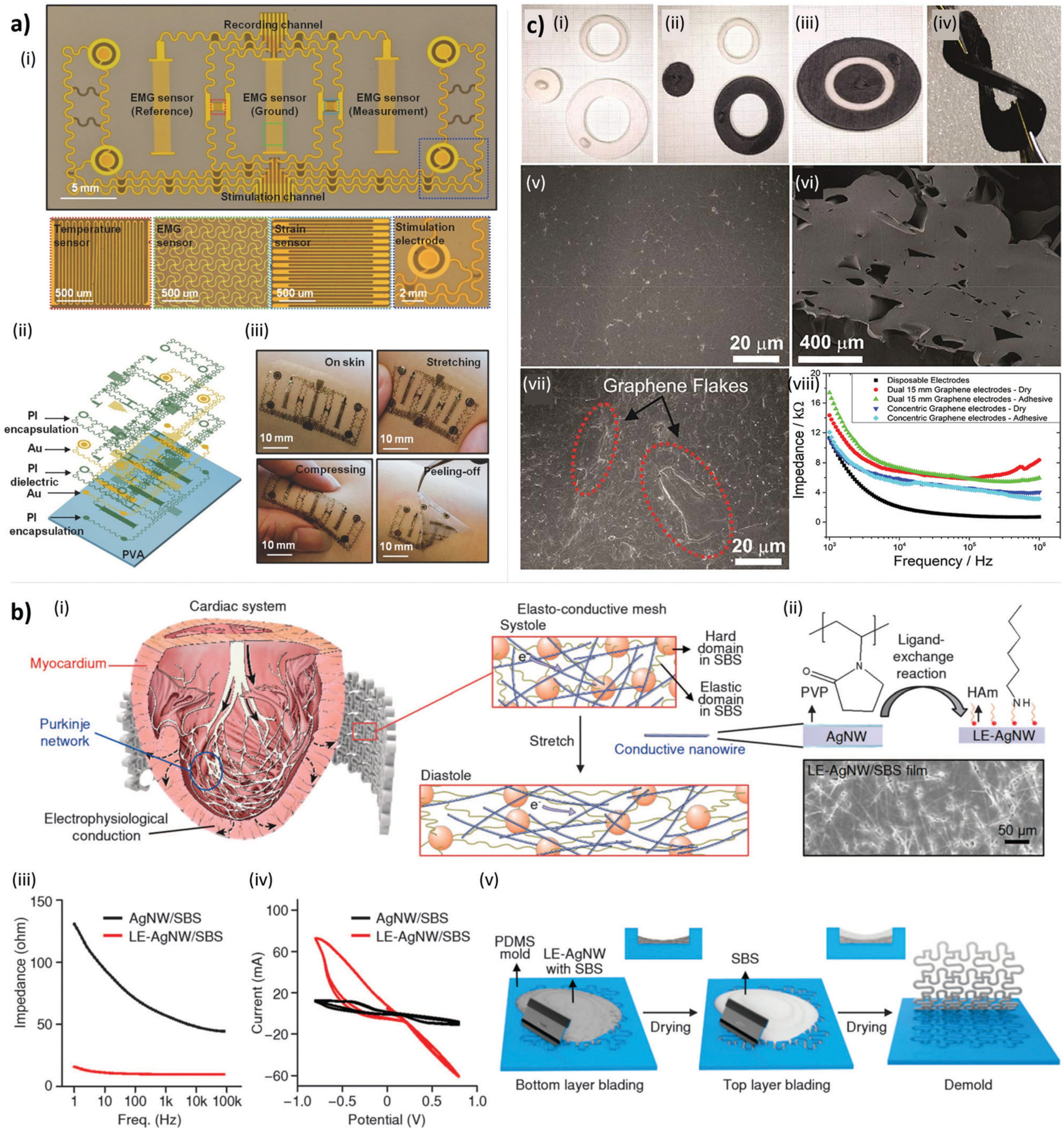

Fig. 4 Materials and design strategies of conformable systems for electrical stimulation. (a) E-skin device capable of both sensing and stimulation. (i) Planar view of the different features of a representative device with insets for the active regions. (ii) Exploded-view of the multilayer construction. (iii) Pictures of the device mounted on the forearm with examples under different deformations. Reproduced from ref. 55 with permission from John Wiley & Sons, DOI: 10.1002/adma.201504155. (b) Elasto-conductive epicardial mesh based on a LE-Ag NWs/SBS composite. (i) Design strategy of the mesh for electromechanical cardioplasty and composite composition. (ii) Scheme of the reaction for LE-AgNW. The ligand of AgNW was exchanged from PVP to Ham; a scanning electron microscopy image of LEAg NW/SBS nanocomposite is shown. (iii) Impedance and (iv) CV (with conductivity measurement) of the initial and LE. (v) Schematic illustration of the epicardial mesh fabrication. Reproduced from ref. 168 with permission from the American Association for the Advancement of Science, DOI: 10.1126/scitranslmed.aad8568. (c) 3D printed, reusable, flexible electrode coated with conductive graphene ink for electrotactile stimulation. Electrode preparation: (i) the electrode components are 3D printed, (ii) coated with graphene ink, and (iii) assembled. (iv) Demonstration of coated electrode's flexibility. (v) Surface and (vi) cross-sectional morphology of the pristine 3D printed component; (vii) surface of the spray coated layer. (viii) Impedeance measurement. Reproduced from ref. 64 under Open Access CC BY Licence, DOI: 10.3389/fbioe.2018.00179.

to the heaters we discussed earlier, requiring distribution of the conductive medium on multiple levels so that only the electrodes are in contact with the skin. On top of this, the proposed devices must also possess a low impedance at the skin electrode interface in order to deliver a reliable signal at low currents and thus limit the discomfort associated with electrostimulation, an issue that often requires an accurate choice of material.[162,163] Finally, all of these characteristics must be suitable to stimulate the target receptors, which respond only to certain electric fields and frequencies.[164] As a result, most of the research and medical applications concern wearable devices,[38,162] which are usually powerful enough to overcome the high impedance of the stratum corneum (*i.e.* the outer layer of the epidermis, with an impedance higher than 10^5 Ω).[165] We will now examine some of the materials and architectures recently proposed in the literature to work as conformable neuromuscular and/or electrotactile stimulators, with their strength points and shortcomings (the most relevant devices are summarized in Table 2).

One fabrication solution to prepare these devices, in which the electrodes require to be in direct contact with the skin, is to deposit a layer of conductive material on a layer of insulator bearing holes in places of the electrodes[55,166] as shown in Fig. 4a (or etch the holes in a sequent step).[167] The difference in height obtained is not an issue for the contact if the size of the electrode is large compared to the thickness of the insulating layer on the skin.[55,167]

An example of this approach is the work of Park *et al.* where the authors obtained a stretchable and conformable Ag NW and SBS composite by mold casting and then covered it with a thin layer of evaporated gold to make it more biocompatible (Fig. 4b).[168] Choi *et al.* used a similar system comprising a network of Au-coated Ag-NWs in SBS encapsulated between SBS sheets.[56] Thanks to a hot pressing step, the thickness of the device was much lower, uniform, and the composite electrodes could already form reliable conformal contact with the skin. Since both the encapsulation and the conductor shared the SBS, the resulting device was intrinsically stretchable and less prone to failure by Young modulus mismatch at the interfaces.[36]

Kim *et al.* realized Au electrodes on a liquid crystal polymer (LCP) substrate (Vecstar) that they could thermo-mold in conformable devices to follow bone structures.[169] LCPs are a class of polymeric material comprising rigid and flexible monomers, which provides them with mechanical toughness and simple processability at high temperatures (even to realize curvilinear manufacts).[170,171] Moreover, they are chemically inert and provide an extremely low gas permeation and humidity uptake (lower than those of polyimmides or parylene), making them an ideal substrate for implantable devices.[171–173] Although our review focuses on external applications, it is worth mentioning that electrical stimulation of cells in the human body with soft and conformable devices can promote regeneration, proliferation, and differentiation, and thus may find many potential applications in therapy.[20,78,87,169,174,175] Notably, the design of some of these devices included the implementation of implantable generators relying on triboelectric effects,[20,87,176] piezoelectricity,[177] or wireless powering[78,178] so as to be energetically self-sufficient, since they would need constant activation. In the future, such solutions could efficiently be applied also to skin-mounted stimulation devices.[179]

Stepping away from gold, Wang *et al.* proposed the use of a liquid Ga–n alloy to prepare an electrode for both muscular stimulation and recording of physiological signals.[153] Ga-Based LMs have the advantage of being non-toxic (even for implantable applications)[180,181] and inherently produce conformable contacts even with the skin;[182] however, their properties have a significant dependence on the thin Ga oxide skin that covers their surface (less than 1 nm thick)[83] and rapidly forms in contact with oxygen.[153]

To reduce the impedance of the contact with the external layer of the skin (the stratum corneum), most of the electrostimulation electrodes require the use of gels or special conductive glues. However, there are solutions to lower the impedance (and thus the working voltage), while avoiding the use of these latter altogether:[183] one may find a lot of research concerning novel dry electrode materials, and also more extensive systems could take advantage of some of these ideas to realize functional conformable devices.

For example, Stephens-Fripp *et al.* realized a 3D printed electrode spray-coated with a conductive graphene-SEBS composite ink (Fig. 4c).[64] The authors claimed it was possible to reuse indefinitely these devices and that they had an impedance similar to that of commercial gel-based electrodes. Wei *et al.* also reported the functional properties of carbon-based materials in lowering the skin impedance.[184]

With a different approach, Tezuka *et al.* avoided the stratum corneum high impedance by implementing microneedles for their tactile displays for electrotactile stimulation.[185,186] The authors used chemical etching of titanium wires with the tip protruding from a soft embedding matrix (PI or Ecoflex) to prepare a matrix of 600 μm tall sharp cones (with a 300 μm diameter at the base). The design of the tips ensured their insertion into the skin without generating pain responses. Moreover, bypassing the stratum corneum lowers drastically the voltage needed to surpass the sensation threshold.

To summarize this section, the main issue conformable electronics must overcome to become a reliable platform for portable, easy-to-use electrostimulation devices is finding solutions to lower the high impedance of the outer skin. The high voltages needed by the wired systems comprising metallic plate electrodes are not compatible with the more delicate flexible electronics. However, materials science still has a lot to offer in this respect: materials like polymer composites and conductive polymers offer a plethora of options – thanks to chemical functionalization and material engineering – to optimize material properties and the way they can interact with the human skin. A recent example of this is the work of Keef *et al.* in which the authors prepared a PEDOT blend with a PSS and poly(polyethylene glycol methyl ether acrylate) block copolymer (PSS-*b*-PPEGMEA) to be used as an electrotactile element in a VR glove.[187] The blend showed high flexibility and low impedance, two ideal qualities for conformable stimulation devices. However, at the moment, only a limited amount of research moves in this direction despite its possible technological and economical values.

3.2 Devices and applications

One of the most innovative objectives in research is to use electrical stimulation to restore lost senses, for example, touch or vision. In recent years numerous studies demonstrated that tactile stimulation increases tactile acuity and manual dexterity in patients suffering from sensory loss, maladaptive plasticity or certain forms of motor impairment, relying on the induction of plasticity in the somatosensory cortex.[188] The user can receive a controlled electrotactile stimulation signal properly modulated in terms of electrical current injected into the skin, which can excite cutaneous mechanoreceptors because of the generation of a potential gradient[21,164] resulting in a tingle, itch, vibration, pinch, or touch sensations.[189] The demand for such systems is high also for human-robot interaction, as required by prosthetic users or remote slave-robot control: adding this feature to a prosthesis can, of course, increase the sense of embodiment of the device[190] as well as allowing the amputee to better manipulate and grasp objects.[191] Moreover, electrotactile stimulation could also substitute the sense of vision in blind people (tactile vision substitution) using tactile displays. Particularly, a two-dimensional matrix of stimulators could give spatial information similarly to the way eyes present spatial information to the retina.[192]

Transcutaneous electrical stimulation (TES) consists of activating excitable tissues in the human body from the skin surface, placing the electrodes over the area of the skin where one intends to activate underlying tissue (see Fig. 5a).[164] Electrodes used in TES shall provide optimal performance without causing permanent skin damage such as burns or irritation and without producing discomfort. To avoid this latter aspect, one should consider designing electrodes that do not allow high current to reach deep into the tissues. An important issue in TES is skin impedance (*i.e.* opposition of the skin to AC flow). It is possible to study physiological operation of the skin thanks to models that show its electrical characteristics, going from a simplified R-C circuit model to more complex models considering the three layers that compose skin – *i.e.* stratum corneum, dermis and hypodermis, as shown in Fig. 5a.[165] Since the skin is an insulator and it stores charges on its outer surface, it acts as a capacitor, whose resistance is inversely proportional to frequency: an increase in the AC frequency results in a decrease of pulse duration, allowing less time to store charges on the skin and, consequently, effectively lowering impedance. Hence, when medium- (1000 Hz) or high-frequency currents are applied, they pass more easily through the skin layer (than low-frequency ones), allowing more current to reach the motor nerves.[148,193] Furthermore, the voltage-current dependence is linear if low currents (in the order of μA) and low voltages (lower than 1 V) are considered, while for higher currents the impedance results in a non-linear function of the current density.[164] The skin presents local inhomogeneities due to the presence of pores or different water content, leading to resistance changes also in the same subject, causing higher or lower current densities.[164] Measurements and models concerning bioimpedance detection could be of help as a feedback reference for adjusting stimulation parameters such as current amplitude, pulse width, and frequency, promoting methods to reduce impedance differences in the human body or between subjects.[165] Moreover, the electrodes should have good contact with the skin allowing the largest surface at the interface. The current density should be homogeneous over the entire electrode. The size of the electrodes depends on the size of the region to stimulate, for example the size of the muscle.[164] Smaller electrodes allow higher selectivity, but the case of using electrical stimulation for restoring sensory feedback requires larger electrodes to produce a comfortable sensation.[64]

Regarding muscular activation, especially in FES, multi-electrode arrays have some advantages with respect to a single pair of electrodes. Arrays allow the user to try many different electrode sizes and positions so as to find the best ones for stimulation, without removing the electrode from the skin.[194]

For all the applications mentioned above, conformable devices could be particularly practical as they inherently provide good contact with the substrate, they are less bulky and more comfortable than applied electrodes, their application is less invasive, and the user could wear them when they prefer, even when not in use, thus making self-use at home possible.[11,23,35,36]

Implantable electrodes for electrical stimulation often rely on conformable technology so as to adapt to the human body tissues. Such devices can be implanted (i) in the brain: deep brain stimulation (DBS) helps patients with epilepsy or Parkinson's disease,[173] and neuroprosthesis implants (such as auditory brainstem implants) can provide sound sensations stimulating neurons;[195] (ii) in muscles: functional electrical stimulation (FES) can restore sensory or motor functionality if applied on a targeted muscle, against atrophy or for generating limb movements like walking or grasping;[164,196] (iii) in the spinal cord: epidural electrical stimulation of the spinal cord enables the use of functionally silent descending neural pathways in order to produce movements of paralyzed limbs and also improves the ability of the spinal cord to translate sensory information into muscle activities that control standing and walking.[19,197,198]

In recent years, the interest in non-invasive techniques has grown, with the objective of avoiding surgical implantation of electrodes, thus renouncing the high selectivity provided by invasive methods. However, studies on the use of conformable electrodes to mount on the skin are still very few. This could be due to the difficulty in reaching low impedances in conformable devices compared to the disposable electrodes generally used in TES – which usually make use of a suitable gel,[64] thus resulting in a higher voltage to maintain the desired current and higher power consumption. Usually, the conductive gel enhances contact between the electrode and the skin; however, conformable electrodes cannot use it.[166] Although we want to remind the reader that the stimulation parameters used in tactile feedback are very subjective and can widely vary inter-subject, rarely the sensation threshold in terms of current amplitude is less than 1 mA, reaching tens of mAs in FES applications.[199,200] Because the current injected could have significant values to electrically stimulate the skin, having a consistent and firm electrical contact with the skin is a central issue in order to avoid sparks that can damage the electrodes and can burn the skin. Another factor is

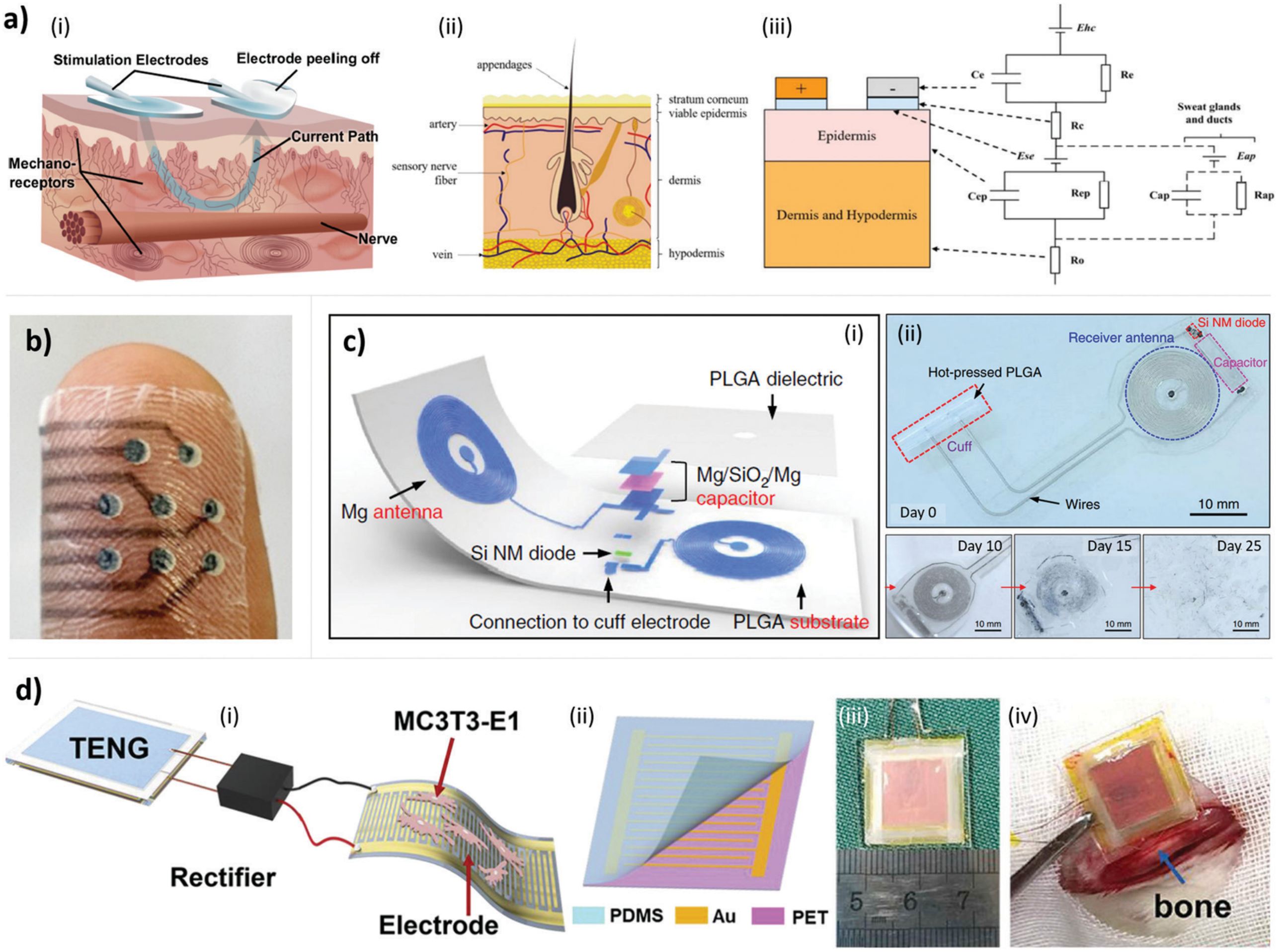

Fig. 5 Functioning principles, schematics, and performance of conformable devices for electrical stimulation. (a) Schematic illustrations of (i) transcutaneous electrical stimulation (TES) (reproduced from ref. 200 with permission from the American Association for the Advancement of Science) DOI: 10.1126/SCIROBOTICS.AAP9770 and (ii) the skin structure with (iii) an equivalent circuit of human skin impedance measurement. Reproduced from ref. 165 under Open Access CC BY Licence, DOI: 10.3390/bios8020031. (b) Picture of a temporary tattoo electrode interface, named Tacttoo, providing a feel-through interface for electro-tactile output on the user's skin. The Tacttoo can be applied on complex body geometries, including the fingertip. Overall thickness is less than 35 μm. Reproduced from ref. 166 with permission from the authors, DOI: 10.1145/3242587.3242645. (c) Example of a conformable wireless and bioresorbable implantable device for electrical-stimulation-assisted tissue regeneration: (i) schematic illustration of the device and (ii) images of the fabricated bioresorbable stimulator, with its dissolution over several days once immersed in PBS, and its implantation. Reproduced from ref. 78 with permission of Springer Nature, DOI: 10.1038/s41591-018-0196-2. (d) Example of conformable self-powering implantable device for enhancing osteoblasts proliferation and differentiation. (i) Schematic illustration of the device and (ii) of the interdigitated electrode. (iii) The size of the implantable TENG and (iv) the surface of the mouse femur region where the flexible TENG was implanted. Reproduced from ref. 20 with permission from Elsevier, DOI: 10.1016/j.nanoen.2019.02.073.

the increased complexity characterizing the manufacturing of these systems. Some examples of conformable electrical stimulating devices will follow.

3.2.1. Tactile feedback. Xu *et al.* designed a multifunctional and fully conformable platform capable both of recording (electromyography, temperature, and strain) and stimulating.[55] The authors demonstrated how a user controlling the grasping force of a robot through EMG signal had better performance in grasping a bottle if provided with tactile feedback. In particular, the current used for electrical stimulation was directly proportional to the force measured by force sensors on the griper, thus resulting in a better modulation of the applied force. Without feedback, the subject could not consistently close the gripper without causing the bottle to collapse. Tezuka *et al.* developed a thin and flexible electrotactile display on a PDMS substrate provided with titanium microneedle electrodes that one could attach on the forearm of the user so as to reproduce tactile sensation.[186] The device consisted in an array of microneedle electrodes. The activation in sequence of each column and the variation of the stimulation flow (*i.e.* the speed in switching columns) reproduced 2D touch sensation. Moreover, microneedles with a length of 600 μm allowed penetrating the stratum corneum resulting in a much lower voltage supply than superficial electrodes for tactile stimulation. Something less invasive was proposed by Ying *et al.* The device consisted of electrode arrays capable of providing electrotactile stimulation using silicon nanomembrane diodes directly, wearing the platform on the finger. The very thin profile and the serpentine-shaped interconnections allowed a conformal contact of the device with human skin. Modulated currents sent through each pair of electrodes can create a localized tactile sensation (*i.e.* tingle or vibration feeling).[167] More recently, Choi *et. al* also presented an ultra-conformable tactile stimulation device, coupled with a heater for thermal stimulation. Their bipolar stimulation Ag/Au electrodes had a low current threshold due to a low impedance resulting in a high-power efficient device.[56]

Withana *et al.* developed a very thin tactile interface (35 μm thick) consisting of an electronic temporary tattoo able to provide safe electrical operation of skin-worn electrotactile devices through a three-layer architecture.[166] To overcome the challenge of achieving perfect contact with the skin, the authors selected a combination of elastic and rigid materials with different thicknesses in order to create a micro-spring mechanism that pushes electrodes toward the skin. Moreover, the design of the three layers allowed control of the current flowing through the skin so as to avoid and prevent over-currents and leakage currents. Screen printing of PEDOT:PSS and Ag/AgCl on a substrate of commercially available temporary tattoo paper allowed fabrication of the tattoos. Thus designed, the tattoos could be easily used by the subjects themselves without the intervention of specialized-personal to attach such devices on the skin (see Fig. 5b).

3.2.2. Vision restoration. Some other applications of electrical stimulation concern visual substitution methods that aim to replace vision with touch senses, placing touch stimulation electrodes on the skin – fingers – or on the tongue.[201–203] For example, one may consider the latter three studies mentioned in the previous subsection as sensory substitution platforms where tactile sensations generated from the electrically activated tactile displays give information to blind people, substituting the sense of vision with touch.[56,166,167] However, such set-ups could be very uncomfortable for everyday activities. Hence, Zhang *et al.* proposed an electrical stimulation device to be directly applied on the cornea of the blind for visual reconstruction. In particular, the idea is to use a camera able to acquire image information sent to a controller that encoded it with the consequent generation of electrical pulses, and then transmitted to the conformable device so as to stimulate the cornea. The device, fully conformable and biocompatible, made of parylene and platinum electrodes, can work with a low voltage (around 1.7 V) due to the high sensitivity of the cornea.[204]

Another innovative field of research is retinal prosthetic devices that aim to partially restore vision in patients suffering from retinal degeneration pathologies. Flexible and conformable materials are perfectly suitable with such application, since the available space between the eye and orbital rim is limited. Furthermore, the continuous movement of the eyeball may impose mechanical stress also on the device, leading to a higher risk of device failure and discomfort to patients. Jeong *et al.* proposed an eye conformable structure that can fit the curvature of the eyeball made of a liquid crystal polymer (LCP). LCPs are compatible with microfabrication techniques and different LCP films can be thermally bonded together because of their thermoplasticity. The authors used a monolithic fabrication process consisting of a multilayered integration of electrical components, thermal deformation of the layers, LCP-power packaging, and laser-machining. The implanted device so constructed could generate biphasic current pulses delivering electrical stimulation.[172]

Both the two studies involved *in vivo* experiments only on animals: the possibility to restore vision in humans using a conformable stimulation device, and restoring tactile feedback in patients suffering from the loss of tactile sensations are still a broad and innovative field of research.

3.2.3. Medical treatments. Electrically stimulating devices also find their use in multiple medical or medical-related fields. In particular such devices can help in solving or ameliorating some of the major global health problems: conformable and implantable electronics can help in various therapeutic applications such as cellular proliferation and differentiation for tissue regeneration, sweat analysis, drug delivery, tumor treatments, and smart defibrillation.[205]

All the internal mechanisms of the human body base their working principle on bioelectricity, which is present also in almost all cell events. Hence, numerous cell types respond to electrical stimulation: this aspect can help in tissue repairment (with cellular proliferation) and in cell differentiation.[146] Some examples are electrical-stimulation-assisted tissue regeneration (see Fig. 5c),[78] cell cultivation on scaffolds made of conductive and high biocompatible materials creating substrates for the attachment, proliferation and neural differentiation of stem cells,[174,206] and enhancing osteoblast proliferation and differentiation against osteoporosis and its related fractures (Fig. 5d).[20]

Iontophoresis is a process that makes use of short-timed and localized current pulses (0.1–1.0 mA cm^{-2}) to improve the drug diffusion through the skin. One of its major applications consists of inducing sweat by stimulating sweat glands infusing pilocarpine. In particular, the measurement of the levels of sodium and chloride in sweat is useful in the diagnosis of cystic fibrosis.[207,208] For example, Heikenfeld *et al.* fabricated a sweat-sensing patch that took advantage of iontophoresis to stimulate sweat production (by releasing pilocarpine) and analyze its composition, also transmitting wirelessly this information to a smartphone.[208] In particular, the measurement of the levels of sodium and chloride in sweat is useful in the diagnosis of cystic fibrosis.[207,208] Since sweat can give additional information related to glucose monitoring, iontophoresis could also help in diabetes management: reverse iontophoresis is the passage of a current over the skin to drive ions from the interstitial fluid and onto the surface of the skin, where they can be analyzed.[209] Bandodkar *et al.* proposed an iontophoretic-biosensing tattoo-based platform targeting non-invasive glucose extraction and monitoring:[210] thanks to the application of a mild current to the epidermis, ions move across the skin and toward the electrodes. The device consisted of a pair of reverse iontophoresis electrodes (Ag/AgCl ink), a pseudo reference/counter (Ag/AgCl ink), and working electrodes (Prussian blue ink). In particular, each tattoo consisted of anodic and cathodic contingents and an amperometric biosensing module using a glucose oxidase

(GOx)-modified Prussian blue transducer able to work at a low voltage. Once the tattoos were mounted on subjects, a 0.2 mA cm^{-2} current applied between the two iontophoretic electrodes for 10 minutes allowed extraction of the interstitial fluid. Then a voltage of −0.1 V applied for 5 minutes at the GOx level allowed the amperometric glucose response to be recorded. Kim *et al.* reported a conformable device able both to sample and analyze in a non-invasive way epidermal biofluids such as glucose and sweat, thanks to the parallel operation of reverse iontophoresis for fluid extraction and iontophoretic delivery of the sweat-inducing drug pilocarpine into the skin.[211] In particular, they developed an epidermal temporary tattoo integrated with a flexible circuit to drive the iontophoretic electrodes, control the biosensors, and wirelessly transmit the sensed information to a smartphone. Thanks to electrorepulsion, the tattoo delivers the positive charged policarpine from the anode so as to generate sweat, while glucose moves toward the cathode because of electro-osmosis and the consequent conductive flow.

In addition, the application of an electric field finds use in tumor treatments. One of the techniques used in electro-chemotherapy is the electroporation, *i.e.* the application of short, millisecond and high-voltage (5–500 V) electrical pulses on the skin through closely spaced electrodes, so as to help diffuse the drug into cancer cells by enhancing permeability as a result of the formation of small pores.[207] To date, most of the electrodes used in this application are rigid and do not adapt to soft body tissues. Hence, because of the possibility to completely wrap the tumoral tissue with the electrode, conformable electrodes could be of central importance to increase the effectiveness of electrochemotherapy in tumor treatments. Such electrodes should deliver an electric field in the range of 20 and 30 kV m^{-1} so as to obtain effective electroporation. Nenzi *et al.* proposed a polymer-metal-hybrid conformable flexible electrode realized thanks to the conversion of porous silicon into nano-porous metals filled by a biocompatible thermoplastic polymer, able to treat large areas of the human body (16 cm^2).[212] They demonstrated that the most effective configuration for this application is the one where an electrode at a specific potential is encircled by electrodes with opposite polarity: they developed an electrode made of 40 concentric square loops 250 μm wide, with the outermost 14 loops set at 0 V, and the other 25 inner loops at 1000 V. The authors showed in simulation that this configuration could generate an effective electric field of 27 kV m^{-1} up to 1 cm depth in biological tissues.

Another non-invasive regional treatment against tumors consists of low-intensity, intermediate-frequency, and alternating electric fields delivered through transducer arrays on the skin around the region of the body containing the tumor in order to kill cancer cells. This particular therapy relies on tuning the frequency of the electric field according to specific types of cancer.[182] Indeed, the applied pulsed electric fields induce the formation of nanopores in cell membranes that play a significant role in destroying cancer cells: cell death occurs by immediate membrane rupture or by programmed cell death mechanisms like apoptosis or autophagy.[213] In this application, rigid circuits present disadvantages such as poor coupling performance, discomfort or hurt on the human body. Conformal electrodes could instead increase the contact area with the skin and enhance therapeutic effects. To this aim, LM electrodes could be optimal candidates for flexible electronics in tumor treatments, directly printing them on the skin in correspondence to the tumor or injecting them in the body.[91,214] Li *et al.* conducted in-vivo experiments on mice and demonstrated how this method slowed down malignant melanoma growth after 6 days of electrical stimulation treatment.[182] A combination of such treatment with chemotherapy could increase killed cell numbers.

Next to tumor treatment, also other medical applications use LM because of their capability to perfectly conform to soft tissues. The study conducted by Wang *et al.* demonstrated, through finite element method simulations, the effectiveness of using conformable LM electrodes for smart defibrillation.[153] They modelled a simplified human trunk applying the electrode on the chest. The results showed that conformal LM electrodes reached more uniform electric potential and electric field than rigid electrodes, probably because of their direct and immediate skin contact, high conformability, and good conductivity. The growing interest in conformable and flexible devices led to the development of heart-conformable electrodes, aiming at an integration of the devices with the epicardial surface. Heart activity relies on pumping blood thanks its muscle contraction (myocardium) induced by electrical impulses. A failure in its conductive system leads to heart impairments. Park *et al.* introduced a soft, conductive and elastic epicardial mesh designed to mechanically adapt to the heart thanks to its elasticity, almost identical to the epicardium one, enabling global pacing to stimulate the whole ventricle.[168] The mesh could detect abnormal activities such as spontaneously occurring ventricular tachycardia and ventricular fibrillation and could deliver an electrical shock to terminate ventricular fibrillation, demonstrating that the device could serve both for pacing and defibrillation in the clinical setting.

4. Conformable devices for mechanical stimulation

Haptic interaction refers to the process by which humans touch, explore, and manipulate objects, but it is also an important means for humans to perceive the external world, receiving stimulation from the environment.[215] The sense of touch is one the most important senses of the human body, allowing humans to interact with the world. In particular, two sub-senses build the sense of touch: kinesthetic and cutaneous feedback. The first refers to receptors located in muscles, tendons, and joints, which provide proprioception (*i.e.* limb position) and muscle tension. Cutaneous sensing consists of mechanoreceptors embedded in the skin and is responsible for pressure and touch sensations: the latter have become the primary means for haptic interfaces.[216]

Despite most of the modern technologies relying on visual and audio feedback,[217] the significance and potential of the tactile sense led to the development of numerous applications concerning virtual or augmented reality that requires tactile feedback to recreate the mechanical properties of objects – *i.e.* mass, stiffness, and temperature – for virtual interaction. The development of such technology will have a strong impact on many different aspects of our life, from social media and communications, to gaming and entertainment, to clinical medicine, rehabilitation and recovery (Fig. 7c).[218] Haptic feedback is becoming progressively important to increase immersion in such applications. Different researchers focus on non-rigid interfaces between the user and the device and, also, in using soft materials for interactive applications.[219]

One of the most investigated methods to provide sensory feedback to users are vibrations,[220] although in recent years the interest related to the possibility to generate displacements and morphological changes of the haptic device has also grown.[221] In particular, in the following sections, the term 'actuator' refers to specific devices piloted with different driving forces developed just to exert some pressure on the human skin. Given that the different types of mechanoreceptors distributed in the glabrous skin respond to mechanical pressure, vibration, or deformation,[216] tactile sensations are provided by mechanically stimulating the skin through pressures/forces applied by actuators, thus reproducing the true nature of the sense of touch.[186] The request for soft materials has led to focus on polymer-based dielectric elastomers because of their cost-effective fabrication techniques and their flexibility and elastic properties being similar to those of human skin, allowing adhesion on body surfaces.[222] Also shape memory alloy (SMA) platforms could be of interest for the development of haptic devices because of their capability to stretch skin once attached, creating shear forces similar to light pinches.[223] Numerous tactile displays have been produced, consisting of tactile stimulators that stimulate human skin.[203] Tactile displays are especially a means for communication for the visually impaired: new fabrication techniques allowed very thin, flexible, and lightweight sheet-type Braille displays to be created.[203,224] In rehabilitation and, in particular, neurorehabilitation, studies demonstrated how robotic therapy provided with haptic interaction with the user improved performance of subjects following stroke or paralysis.[225,226] In the medical field, haptic feedback has relevant importance in minimally invasive surgery, in particular in laparoscopic surgery, where the surgeon loses his own tactile feedback due to the interposition of long instruments between his hand and the interested tissue. Haptic feedback devices could help the surgeon in identifying tissue properties and in handling tissues in a safe manner.[215,227]

4.1 Materials and architectures

Devices capable of providing a tactile sensation to a human being has to actively perform a displacement in the range between microns and millimeters. On top of this, by matching the frequencies of the different mechanoreceptors in our skin, they can more efficiently provoke a sensation, while limiting the required power.[30,228] To achieve this, many diverse solutions have been brought up relying on different actuation technologies, some of which have been successfully employed in conformable devices. However, we must make a distinction between wearable and actually conformable devices. The former – as in many examples of gloves for VR applications – can just serve as a substrate on which one can mount pneumatic and mechanical actuators. They are very efficient in providing a sensation, but are also generally bulky, they require external moduli, and limit the freedom of movement of the user.[217,223,229,230] On the other hand, the design of conformable tactile stimulators allow one to maximize the comfort of the user and simplify the utilization procedures.[231] However, it is worth mentioning, that there are many technologies from these formers to be applied to conformable devices with few modifications (Table 3). Song *et al.*, for instance, realized a soft-pneumatic actuator, which did not require the use of an external compressor.[232] To do so, they encapsulated an air pocket between two silicone sheets, which bared two concentric electrodes (Fig. 6a): the electrostatic attraction between the latter would squeeze the side of the pocket thus pushing up the central part of the element not covered by the electrodes to keep the volume constant. In another recent example, Yu *et al.* prepared an array of Lorentz-type actuators embedded in a silicone rubber thus resulting in a flexible tactile display (Fig. 6b).[218] Such actuators work based on the electromagnetic force generated from a current flowing in a magnetic field. Remarkably, their device could power wirelessly thanks to a flat antenna embedded in the same matrix. These two studies point out that, compared to the thermal and electro stimulation, in this case the skin contact does not need to be as intimate, and therefore the devices can be less compliant. However, the latter are still quite thick. Thinner systems would be preferable to facilitate their use.

A particularly studied solution involves the use of dielectric elastomer actuators (DEAs).[30,233] They consist of a layer of insulating elastomer sandwiched between two electrodes: once charged, a Coulomb force pushes them together, deforming the elastomer (which will try to retain its volume).[222] Since the deformation is rapid and reversible, they are capable of producing a large force and reaching high frequencies. They, however, require high working voltages (in the order of kV) to obtain a significant displacement. To improve the performance of these devices, the elastomers employed should have low Young and shear moduli (allowing a large strain with a high frequency), a large dielectric constant, and a high breakdown threshold (larger than 10^{-6} V m^{-1}).[234] The most frequently employed polymers are polyacrylate elastomers, polyurethanes, and silicones. By rational design of the device architecture and choice of the materials, these systems can be very versatile with respect to the type of motion they can provide and their dimensions, allowing the preparation of wearable and conformable devices.[203,235] Mun *et al.* fabricated a compliant, 500 μm-thick DEA device comprising a flat multilayered PDMS/AgNW-electrode structure (Fig. 6c).[236] Thanks to a slight curvature in the active element in the OFF state that induced asymmetry, the authors could obtain a vertical displacement from an otherwise flat geometry.

Table 3 Examples of materials and architectures employed in conformable haptic devices based on different working principles and their properties

Working principle	Material/design	Force or displacement	Freq.	Application	Ref.
DEA	8 Elastomer layers between carbon electrodes of alternating polarity (200 μm in total)	Up to 0.9 mm	0–150 Hz	Tactile display	203
Lorentz force	Moving magnet over metal coil	Up to 0.3 mm	0–300 Hz	Wireless tactile display	218
Soft pneumatic act	Air pocket squeeze by electrostatic attraction between electrodes	0.10–0.13 mm	0.2–1 Hz	Elements for VR glove	232
DEA	5 PDMS layers between AgNW electrodes of alternating polarities (500 μm in total)	Up to 0.33 mm	1–330 Hz	Tactile display	236
DEA	PDMS between AgNW electrode perpendicular to the device (total thickness 50 μm)	Up to 12 μm	20–240 Hz	Transparent tactile display and sensor	238
DEA	25 PVDF layers between Ag electrodes of alternating polarity (700 μm in total)	Up to 4 μm	250 Hz	Haptic stimulation and sensing	239
Piezo	500 nm lead zirconate titanate nanoribbon between Au and Pt electrodes		1 kHz	Measurement of mechanical properties of biological tissues	240

Note: in this table we reported only conformable devices as discussed in the text, leaving out the wearable options.

Not only the fabrication of these devices can result in thin, conformable systems, but it can also incorporate other functionalities.[237] Yun *et al.* realized a flexible and transparent tactile display capable of both delivering haptic stimuli and sense touch.[238] Using an ingenious fabrication involving masked evaporation on vertical PDMS patterns, they obtained the Ag/PDMS/Ag actuators perpendicular to the device surface and thus capable of exerting their force directly on the skin (Fig. 6d). Yoon *et al.* used a multistack of PVDF layers between Ag electrodes to produce tactile sensations, while at the same time working as a pressure sensor taking advantage of the piezoelectric properties of PVDF.[239] Piezoelectric materials also find applications as stimulators; however, despite them reaching very high frequencies (kHz), the displacement they can achieve is very limited.[240,241]

Electro-active systems, such as those introduced so far, have the advantage to reach high frequencies that can improve the interaction with the skin without increasing the power. Other electro-responding actuators based on functional materials – such as ionic conductive polymer metal composites or electro-adhesive displays – have been employed for tactile stimulation but never implemented in conformable devices.[224] A recent review specifically about haptics by Biswas *et al.* introduces them nicely.[30] However, there are many more materials that one could use instead, whose actuating principles do not rely on electromagnetic forces.

SMAs are deformable alloys which are capable of reforming their original shape upon heating.[242] A current that flows in a SMA wire can generate enough Joule heating to trigger the response in many alloys of specific compositions, making them suitable for skin-contact applications. SMA-based actuators make use of the force that accompanies this transformation, which can be of several N for a mm of wire, and enhanced even further with shape engineering.[243] In the literature, one can find recent examples of devices relying on this technology for tactile stimulation.[220,223,230] Compared to previous examples, such systems have longer actuation time (1–2 s) but can provide a much more realistic sensation when compared to vibro- and elctro-tactile stimulation by actually stretching and pinching the skin.

A different type of functional material often used as actuators in soft robotics are liquid crystal elastomers (LCEs). These are polymeric materials obtained *via* the polymerization of molecules characterized by liquid crystalline behavior (*i.e.* comprising a flat, rigid, and long region and flexible side chains).[244,245] While at low temperature most of the molecules are aligned and stacked together in a well ordered nematic phase, upon application of suitable stimuli, the order is lost and the material changes shape, shortening along what was the alignment direction and broadening in the other directions. Such movement is reversible and can be triggered by different stimuli such as heat, light, and electric fields, thus providing actuation in different systems, often with tensile strengths of several kPa.[246–249] While LCEs have found application in soft electronics, deeper studies should better explore their potential in conformable devices. Similar remarks can be made on conductive polymers (CPs), one of the go-to choices for actuators in soft robotics, as they require low voltages to work (but relatively high currents).[221,224] However, both CPs and LCEs, usually require longer actuation time than DEAs.[49,146]

4.2. Devices and applications

Mechanical stimulation is the most commonly used method to restore tactile sensations of the user: mechanoreceptors are very sensitive to vibrations, forces, deformations, and skin stretching. Numerous mechanoreceptors are distributed across the skin, in particular in the dermis. In general, the speed of their adaptation to changes in the pressure applied to some points on the skin classify the type of tactile receptors.[192] Hence, mechanoreceptors respond to stimuli coming from the world generating an electrical signal, transmitted to the brain. This incoming signal defines the mechanical sense of our surrounding.[218] In particular, the changes in the firing rate of action potentials on the corresponding afferent nerve could be a way for measuring mechanoreceptor response. Four types of mechanoreceptors exist in the glabrous skin (Merkel cells, Ruffini endings, Meissner corpuscles, and Pacinian corpuscles), which respond to mechanical pressure, vibration or deformation.[250] Since mechanoreceptors are sensitive to

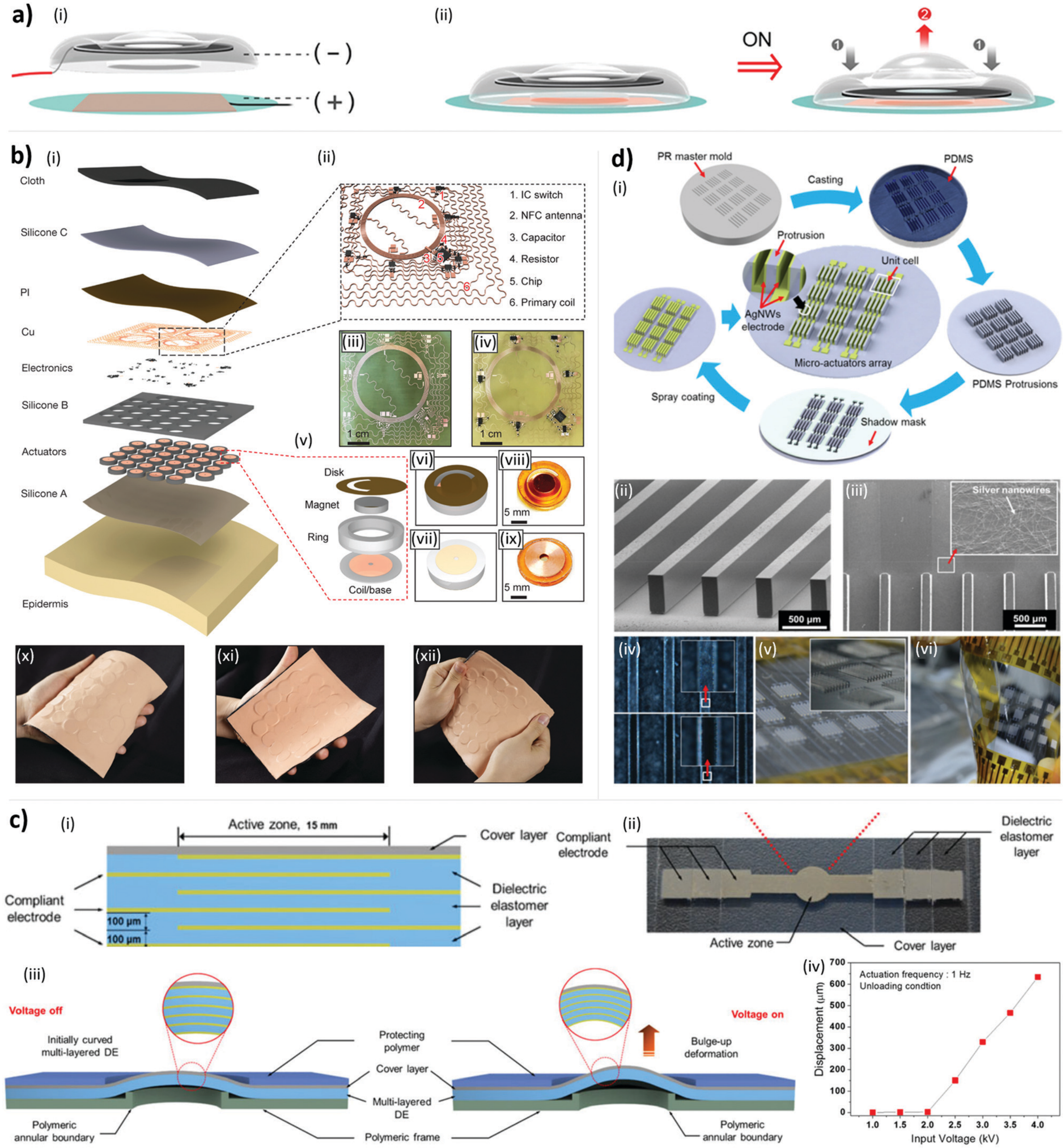

Fig. 6 Fabrication strategies and materials employed for the development of conformable haptic interfaces. (a) Soft pneumatic actuator with internal air pressure generated by an electrostatic force for providing tactile feedback. (i) Structure schematic and (ii) operation mechanism. Adapted from ref. 232 under Open Access Attribution License CC BY, DOI: 10.1038/s41598-019-45422-6. (b) Wireless and battery-free platform composed of electronic systems and haptic interfaces capable of communicating information *via* programmable patterns using mechanical vibrations. (i) Exploded-view of the device and (ii) schematic illustration of the electronics and circuit. (iii and iv) Optical images of an NFC coil; (iii) before and (iv) after integrating the electronic components. (v) Exploded-view schematic diagram of a haptic actuator with the schematic diagram of an actuator viewed from (vi) above and (vii) below, and (viii–ix) their respective optical images. (x–xii) Qualitative mechanical tests of the device under (x) bending, (xi) folding, and (xii) twisting. Reproduced from ref. 218 with permission from Springer Nature, DOI: 10.1038/s41586-019-1687-0. (c) Soft actuator for tactile stimulation interfaces prepared by multilayered accumulation of thin electro-active polymer (EAP) films. (i) Cross-sectional view of the multilayered DE and (ii) a picture of the fabricated device. (iii) Schematic illustration of the operating principle of the DE actuator. (iv) Induced displacement profile as a function of the input voltage at 1 Hz. Reproduced from ref. 236 with permission from IEEE, DOI: 10.1109/TOH.2018.2805901. (d) Preparation of a visuo-haptic interface composed of a touch-sensitive visual display based on polymer waveguides and a dielectric elastomer microactuators (DEMA) array. (i) Schematic illustration of the fabrication process. (ii) SEM images of PDMS protrusions peeled off from a photoresist mold on a Si wafer and (iii) AgNWs patterns on the 3-D structure, respectively. (iv) Optical microscope images of the AgNW-coated protrusions (top) and its change after eliminating the AgNW coating eliminated from the top surface of the protrusions (bottom). (v) Picture of tactile-actuator array with the AgNW electrode patterns and their 3-D geometry (inset). (vi) Transparency and flexibility of the tactile-actuator array. Reproduced from ref. 238 with permission form IEEE, DOI: 10.1109/TIE.2019.2898620.

mechanical stimuli, the tactile sensation can arise stimulating the skin in a mechanical way, reproducing the real nature of the sense of touch.[192] However, each of the four aforementioned mechanoreceptors respond to different stimuli and can therefore be classified on the basis of the different frequencies and displacements of the stimuli to which they respond: (i) Merkel Cells are stimulated at 0.3–3 Hz and need 8 μm displacements; (ii) Meissner Corpuscles respond to 3–40 Hz frequency and 3 μm displacements; (iii) Pacinian corpuscles 40–500 Hz with lower displacements, 0.01 μm; (iv) Ruffini Corpuscles (located deeply) 40–500 Hz and 40 μm.[237] Hence, a mechanotactile device should have high bandwidth and generate relatively high displacements, as we will discuss later.

4.2.1 Tactile displays. The most common means to provide mechanical stimuli to the user's skin is vibration. Vibrotactile displays are considered safer in comparison to electrotactile systems, which can more easily cause pain and fatigue to the skin. Vibrotactile feedback systems are currently used in sports, rehabilitation, and information tactile display, as well as for providing tactile feedback to upper limb amputees through vibrotactile sensory substitution systems.[216,251–253] The increasing demand for wearable electronics led to the development of conformable devices to directly attach to the skin. Conformable and thin devices allow a more efficient contact with the skin and therefore a more effective mechanical coupling. Many recent examples of vibrotactile displays aimed to provide tactile feedback basing their operation principle on soft actuators. One notable example is the use of soft pneumatic actuators made of flexible and compliant polymers to develop devices capable of generating internal air pressure by electrostatic forces.[232] Sonar *et al.* proposed a fluidic haptic display that consisted of silicone layers, aiming to artificially reproduce skin. The artificial skin consisted of piezoceramic sensors for detecting normal forces in order to create a close-loop controlled device, capable of responding to external forces by adjusting vibration amplitude (Fig. 7a). In particular, the device embedded soft pneumatic components activated thanks to air passing through one of the layers, resulting in inflation for reaching the desired shape.[225] The work from Yoon *et al.* proposed a dual functionalities device, both for force sensing and vibrotactile actuation for feedback.[239] The authors utilized PVDF piezoelectric polymer films that deformed with an applied electrical potential. The driving voltage, in this case, was less than 300 V, however reaching output forces lower than 7 mN.

Multiple studies used polymer-based systems that respond to electrical stimulation with shape or size change, called electroactive polymers, which are flexible, lightweight, fracture-tolerant, and easy to mould.[30] Their mechanical properties are comparable to those of the human skin, making these actuators attractive for haptic devices.[238]

Conformable and wearable platforms for haptics can efficiently employ DEAs. Thanks to their lightweight, flexibility, cost-effectiveness, and fast response, DEAs have opened many opportunities in applications such as tactile or Braille displays (Section 4.2.2).[222] The aim is to provide tactile sensations using a localized tuneable mechanical pressure applied to the skin through a soft interface. For example, Carpi *et al.* demonstrated the possibility to employ an acrylic elastomeric film to create a VHB-based DEA wearable tactile interface for use in portable and more cost-effective virtual reality (VR) or augmented reality (AR) systems.[254,255] Some other recent studies demonstrated the possibility of using DEAs for developing conformable devices. Mun *et al.* designed a tactile display to embed into clothes, integrating flexible soft actuators onto the inside fingertip surface of a rubber glove or a patched display that could give vibrotactile feedback to the forearm (Fig. 7b).[236] The soft actuator consisted of a multilayer dielectric elastomer with compliant electrodes made of AgNWs, stacked to obtain six layers. They demonstrated that the device could exert up to 255 mN at the maximum driving voltage of 4 kV when working at the resonance frequency of the membrane structure, considering, however, that the force perception threshold at the human fingertip is much lower ($\sim$2.2 mN at 240 Hz frequency).[228] Such a high range of forces could be useful to provide different sensations, for example distinguishing "hard" touch from "soft" touch. The device presented a response delay of actuation below 1 ms and the actuators produced outputs that covered a wide frequency range.

Fingertips have a very high sensitivity; hence, most of the studies concerning haptic feedback aim at designing devices that fit this region of the body. However, this configuration leads to difficulties in using the hands for everyday activities, such as grasping or manipulating objects.[256] For this reason, Zhao *et al.* proposed a soft device to wear on the arm instead of fingertips, with the drawback that the arm is less sensitive than fingers, requiring higher forces and displacements for stimulation.[235] The devices consisted of a 2-by-2 array of actuators: the authors demonstrated how an array of such components could create an illusion of motion, allowing them to reproduce a moving touch sensation through sequential activation of two actuators. This latter aspect could be of importance in applications concerning VR or AR for more realistic feedback of tactile sensations, as well as for prosthetic users who could detect movements of the grasped object avoiding its slippage. Yun *et al.* designed a perfectly skin-mountable interface based on DEAs arranged in arrays with AgNW-compliant electrodes. The device was able to reach an output force of 130 mN with, however, a very high driving electric field (20 MV m^{-1}).[238]

The great issue related to the devices that base their operating principle on DEAs is the charge and discharge cycle time. In particular, slow charging time led to the slow response of the device because of the limited current flowing into the actuators. This aspect affected frequency response, resulting in high driving voltage requirement and low bandwidth. The devices needed driving voltages of several kVs (electric fields in the order of tens of MV m^{-1})[238] resulting in severe limitations for wearable purposes. A step forward in this sense could be the development of flexible, stretchable, lightweight, and wearable devices consisting in wirelessly powered actuators so as to use haptic display in numerous applications. Moreover, the device should

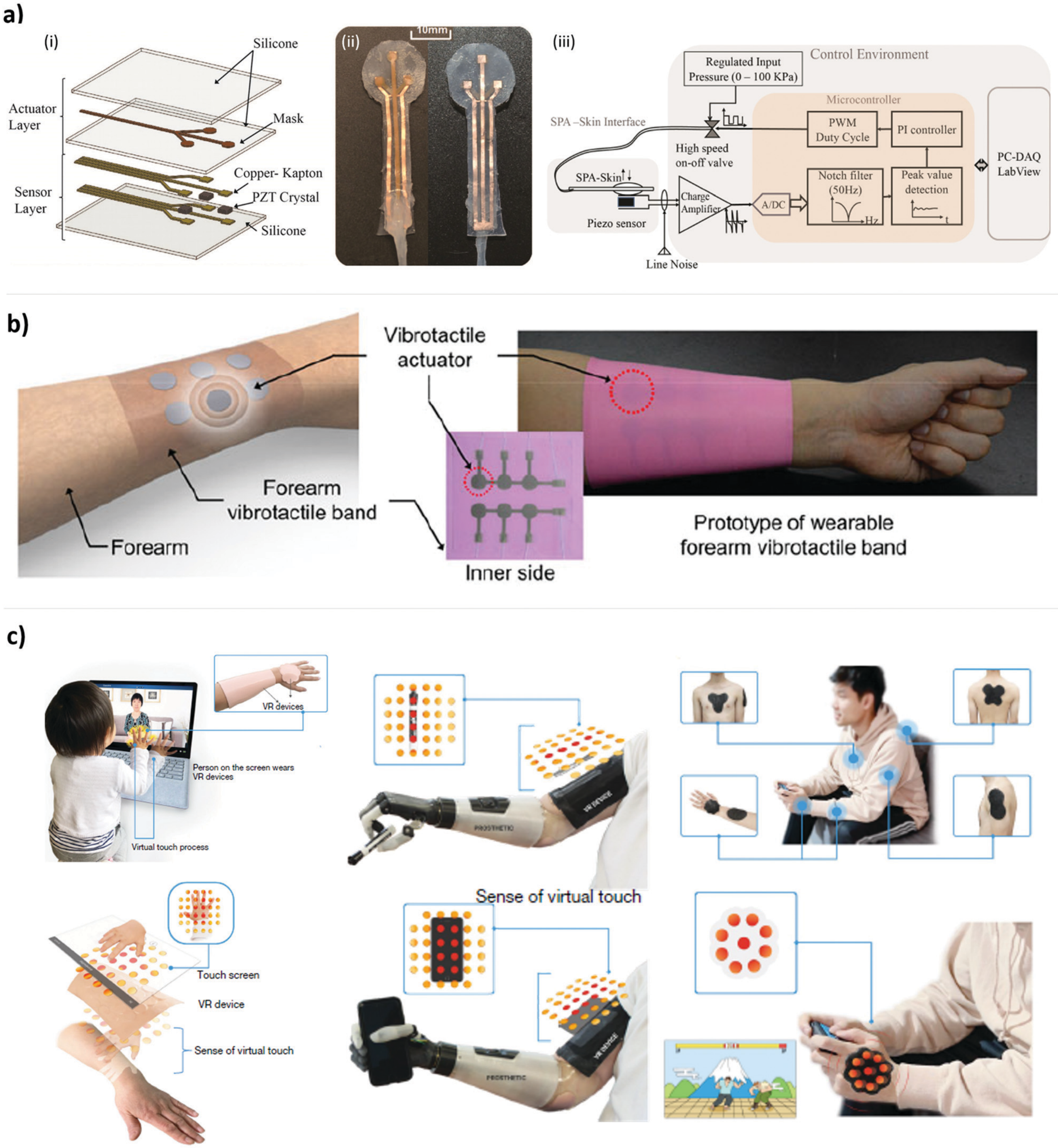

Fig. 7 Examples of conformable devices for haptic interfaces and their applications. (a) Piezoelectric haptic conformable devices. (i) Exploded view of the device. (ii) Optical image of the fabricated device (front and back). (iii) Schematic of the closed-loop control of the actuation mechanism for adjusting vibration amplitude. Adapted from ref. 225 under Open Access Attribution License CC BY, DOI: 10.3389/frobt.2015.00038. (b) Haptic devices based on DEA embedded into a patched display. Reproduced from ref. 236 with permission from IEEE, DOI: 10.1109/TOH.2018.2805901. (c) Examples of VR applications of a battery-free epidermal system used as a haptic interface capable of communicating information *via* mechanical vibrations. Reproduced from ref. 218 with permission from Springer Nature, DOI: 10.1038/s41586-019-1687-0.

include a proper insulation layer because of high-operating voltage, thus leading to higher fabrication costs and output force degradation.[222] Another technical challenge of DEA is the relatively low vertical force they are able to provide and small displacements, despite high driving voltages:[203,236] an ideal tactile display should provide a 4 mm stroke.[257] By causing the device to work at its resonant frequency, it is possible to reach higher forces and displacements.

4.2.2 Braille displays. The possibility to develop conformable devices paves the way for new types of Braille displays, enabling people with visual impairments to carry their own display wherever they go. Kato *et al.* presented one of the first examples of Braille display based on soft actuators demonstrating that recent developments in materials and fabrication techniques allowed the creation of cost-effective devices using, for example printing machines to fabricate electronics – such as organic field transistors – directly on plastic films.[224] Koo *et al.* proposed an example of a conformable Braille display based on DEA:[203] the application of the voltage to the DEA resulted in a curvature of the elastomer. This mechanism generated displacement and an effective mechanical pressure along the thickness direction proportional to the square of the applied voltage: keeping material properties constant, a higher electric field led to a larger interaction force, which was a driving force to generate deformations. The whole device consisted of multiple stimulation cells (a 4×5 array) and fit on the fingertip of the user because of its flexibility due to a very small thickness ($\sim$210 μm), despite it consisting of eight layers of stacked dielectric elastomer. The device exerted a force output of 14 mN at the maximum driving voltage (3.5 kV).

4.2.3 Non-vibrating mechanotactile interfaces. Some interesting devices exploit elasticity of the skin to provide haptic feedback, designing super-cutaneous wearable tactile devices that gently stretch and squeeze the surface of the skin.[220] In fact, vibrations cannot convey a large range of tactile gestures that mimic real-world experiences.[230] An example is proposed by Hamdan *et al.* who developed a tactile interface based on a shape memory alloy (SMA) spring, exploiting it as a mechanotactile actuator: when current flowed into the spring, it heated up, and generated silent-movements, contracting to shorten. SMAs have a very high force-to-weight ratio, thus allowing their use as smart actuators.[242] In this way, when the SMA spring is attached to the skin, the mechanotactile actuator stretched the skin itself. The contraction stopped when the spring reached its shortest length, or when a counteracting force – *i.e.* the skin resistance – became higher than the spring's contraction force. The latter is also provided to restore the initial shape of the spring, once the power is removed. However, the device could provide multiple touch sensations, depending on how the spring is mounted on the skin, for example, press, pull, drag, and expand on the skin surface. The control over the amount and the duration of the current flow allowed to adjust actuation displacement, actuation force, and actuation speed. Nevertheless, SMAs show some drawbacks related to their low controllability due to hysteresis behaviour, low bandwidth and low energy-efficiency (slow heating and cooling rates).[230]

In general, all the interfaces considered above still present some challenges related to adaptation (loss of sensitiveness) and habituation (getting used to certain stimuli) of the user to the provided sensations due to the continuous sensory stimuli. These aspects could result in not completely effective transmission of sensory information in the long-term use of haptic interfaces.[258]

5. Photostimulation devices

In modern medicine, light and optical techniques are profoundly impacting clinical practice in the health assessment and disease treatments.[259] When photons interact with biological tissues, they can either scatter the photons (used to monitor diverse body parameters)[259,260] or absorb them, activating photochemical and biological reactions of the tissues themselves, as well as generating heat.[261–263] New therapeutic strategies exploit these phenomena, leading to an increasing interest in light-based therapies, such as (i) photodynamic therapy, in which a light source with a specific wavelength activates a photosensitizing agent located in a tumor, which causes irreversible photodamage to its tissues;[261] (ii) photothermal therapy, that is a minimally-invasive therapy in which photon energy produces heat (also used in cancer treatment);[262] (iii) optogenetic therapies, which provide initial genetic modifications of cells in order to express photosensitive traits, subsequently utilized for directly irradiating cells with visible light enabling users to photocontrol molecules, proteins, and cells *in vitro* and *in vivo*.[264,265] Most of these kinds of therapies base their working principle on synthetic chromophores that undergo light-dependent structural or electronic changes and can target unique cellular traits, notably proteins. They have been proven useful in the treatment of diabetes, peripheral nerves, pain, and bladder diseases, as well as in regulating cardiac pacing when applied to cardiomyocytes;[259,260,266] (iv) photobiomodulation exploits visible and near-infrared portions of the light spectrum in order to induce chemical changes within the cells, provoking biological reactions that benefit the body in triggering neuroprotective responses, improvements in metabolism, blood flow, and neurogenesis. The relatively low power density ensures no burn damage to the tissues;[263] (v) photochemotherapy depends on the connection of drug and light: the drug exerts locally an antiproliferative effect only after interaction with suitable light, being otherwise ineffective and unable to target more cells than desired;[267] (vi) in photopharmacology light influences the biological activity of synthetic molecules by changing their pharmacokinetic or pharmacodynamic properties: light precisely controls the bioactivity of smart drugs with incorporated molecular photoswitches, which change the molecular structure.[268]

All these therapies can benefit from the use of dermal and transdermal opto-devices as the skin can let through several wavelengths. For example, light with a wavelength of 400–500 nm will stop at the epidermal layer, while light with a wavelength longer than 700 nm can reach deeper tissues beyond the dermis.[260] Hence, one must select the light with a proper wavelength depending on the specific objective of the application (Fig. 8a).

Recent research led to the development of new materials and new techniques to manufacture conformable and stretchable devices. In particular, the possibility of fabricating LEDs with micrometric dimensions (μLEDs) and preparing inherently flexible OLEDs based on polymeric emitters propelled this field forward.[44,269–271] Bioelectronic medicine aims to develop engineered systems that can relieve clinical conditions by photostimulating the body of the patient.[272,273] Hence, while in past years pain relief relied largely on electrical stimulation, now bio-optoelectronic devices can overcome some limits of the

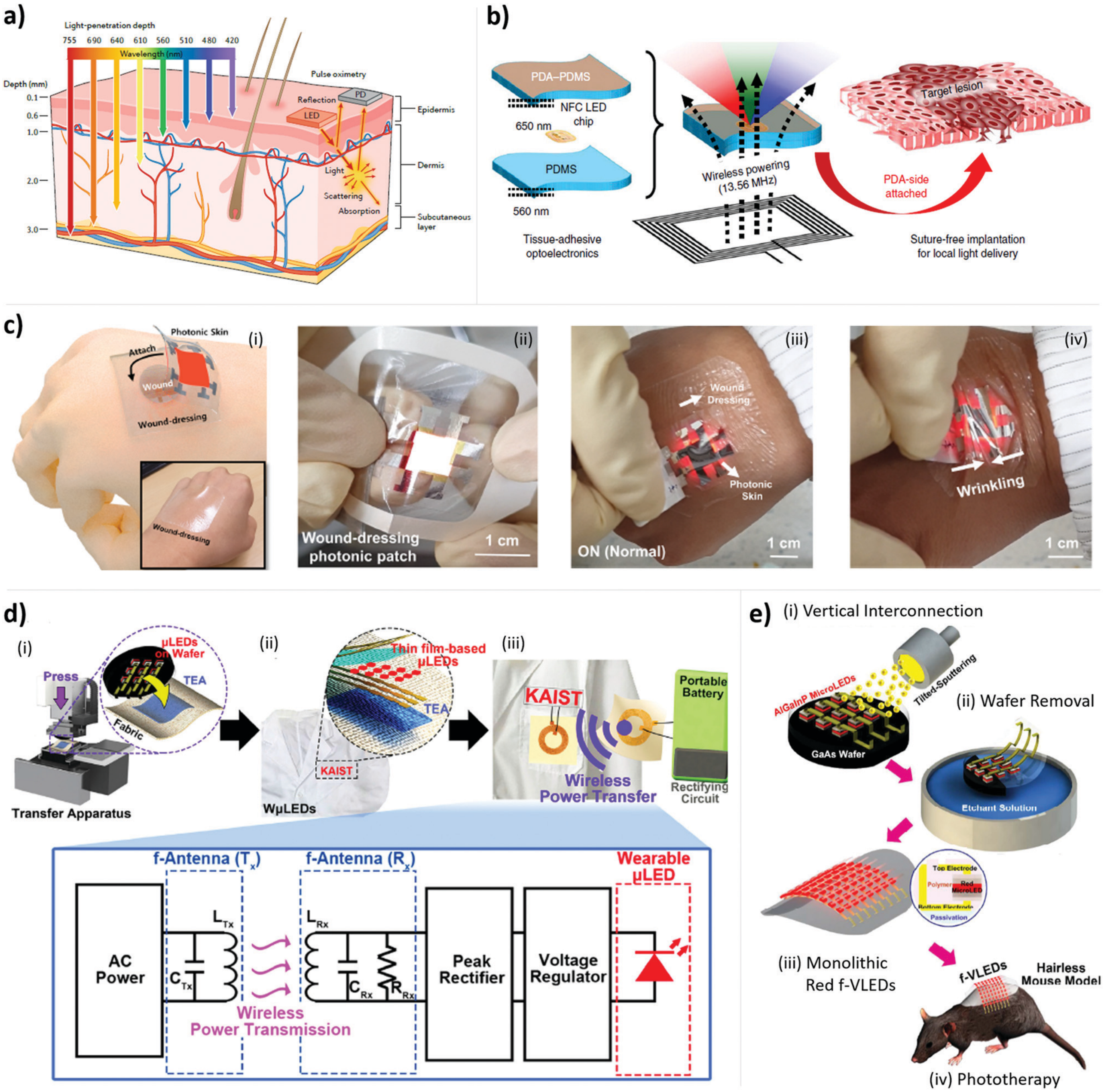

Fig. 8 Examples of photostimulation devices. (a) Schematic depicting how light can interact with human skin. Reproduced from ref. 260 with permission from Springer Nature, DOI: 10.1038/s41578-019-0167-3. (b) Implantable μLEDs embedded in PDMS provided with wireless communication for cancer treatments. Reproduced from ref. 275 with permission from Springer Nature, DOI: 10.1038/s41551-018-0261-7. (c) Flexible OLED for wound treatments. (i) Schematic of device. Optical images of the (ii) device as prepared, (iii) mounted on skin, and (iv) working upon wrinkling. Reproduced from ref. 277 with permission from John Wiley & Sons, DOI: 10.1002/jsid.882. (d) Schematic illustration of a wirelessly powered wearable μLED. (i) Schematic of preparation and (ii) exploded device. (iii) Working principle and circuit schematics. Reproduced from ref. 270 with permission from Elsevier, DOI: 10.1016/j.nanoen.2018.11.017. (e) Schematic illustration of monolithic fabrication process of red μLEDs on a polymer substrate and application of the wearable device on a mouse to test treatments against alopecia. Adapted with permission from ref. 279. Copyright 2018 American Chemical Society, DOI: 10.1021/acsnano.8b05568.

former. Some of these limits are possible discomfort and pain caused by a continuous stimulation protocol, direct physical contact between the electrodes and the nerve, which can lead to injuries and inflammations, and lack of organ specificity.[273]

Most of the conformable photonic healthcare devices found in the literature consist of implantable devices, developed in particular for optogenetic and photodynamic therapies.[260] The most attractive systems consist of μLEDs transferred on a flexible substrate so as to be directly implanted in the body, without the necessity of using an external light source for the specifically considered therapy. Moreover, studies developed devices with LEDs mounted on microneedles, with the ability to operate at wavelengths ranging from UV to blue, green-yellow, and red.[260,274] Their use is particularly growing in biomedical applications that include biosensors, healthcare devices, and optogenetic stimulators, because of their high biocompatibility, low heating property, and excellent stability.[27,271] Indeed, implantable photonic devices must be highly biocompatible and fixed inside the body for continuous local light delivery without causing any damage to other tissues or organs.[260] For this reason, photostimulation implantable devices widely use PDMS as substrates or encapsulation material. Moreover, many are also provided with a wireless power supply system. For example, Yamagishi *et al.* used μLEDs embedded in PDMS for developing an implantable photodynamic device against cancer (Fig. 8b).[275] Similarly, Mickle *et al.* used μLEDs on a PDMS substrate to fabricate a closed-loop device that exploited biophysical sensors for continuously monitoring and controlling bladder function through a smart device in order to eliminate pathological behaviours as they occur in real-time.[273] Some other applications in which implantable photonic devices can offer benefits concern the stimulation of the spinal cord and peripheral nerves, against chronic pain, itch, and other neurological disorders.[276]

Wearable photonic healthcare devices are currently part of several applications including skin rejuvenation, wound treatment, mental healthcare and brain-disease therapy.[260] Current rigid and bulky conventional light sources are, however, difficult to transport and use for regular irradiation treatments.[277] Skin-mountable conformable devices – comprising μLEDs, flexible OLEDs, quantum dot displays and other thin technologies – can overcome these limitations, paving the way for developing lightweight, portable, and comfortable devices capable of providing on-demand, and continuous treatment.[43,260,277,278]

However, to the best of our knowledge, only a few examples of skin-mountable photonic devices exist. One application concerns the management of surgical wounds in order to ensure quick skin regeneration, pain relief, and scar prevention. Surgical wounds typically require periodic light treatment: Jeon *et al.* recently proposed a very thin flexible photonic skin, comprising a iridium(III) OLED, having a total thickness of 6 μm, that they developed for skin attachable phototherapy (Fig. 8c).[277] The authors studied the regeneration effects on an artificial skin model composed of fibroblasts and keratinocytes: OLED light could have beneficial effects in cell metabolism, hence regeneration improved control of the peak wavelengths (600–700 nm) and irradiation interval of the flexible OLED. The authors designed the OLED with a bottom-emitting structure so that light could safely spread out through the substrate without exposing organic materials to the skin, so avoiding possible harm to the user. Moreover, they demonstrated that it is possible to mount the photonic skin on commercial dressing film and successfully applied it to patients with surgical wounds. Stretchability is surely important in wearable devices, so that the platforms can follow all the skin movements and deformations, without degrading. Yin *et al.* proposed another example of a wearable and flexible device composed of highly stretchable iridium(III) OLEDs that could exhibit 100% strain without a decrease in performance.[44] In particular, the authors proposed a laser-programmable buckling process in order to create ordered buckles on the OLEDs, which permitted controllable stretch-release cycles and enhanced stretchability and mechanical robustness of the devices. In general, such platforms work through wireless power-transfer systems.[272] Lee *et al.* proposed a method to supply electrical power to μLEDs using a wearable wireless power transmission/reception system consisting of flexible antennas, a peak rectifier, and a voltage regulator (Fig. 8d).[270] An AC power supply source supplied radio frequency power to the wearable transmitter antenna wirelessly, which was collected by the wearable receiver antenna and then transmitted to the peak rectifier for converting AC to DC power. The converted DC power hence powered μLEDs through the voltage regulator. Another interesting application of the skin-mountable photostimulation devices concerns the acceleration of hair growth against alopecia. Lee *et al.* developed a methodology to prepare flexible microscale red LEDs fabricated through a simple monolithic fabrication process, and successfully used these elements to stimulate hair follicles in a rat, which are located 2 mm under the epidermis (Fig. 8e).[279] The 650 nm radiation could penetrate deep into the skin compared to green and blue ones. This method aims to replace the use of laser stimulation, generally used in the clinical practice against alopecia, which has numerous drawbacks (*e.g.* high-power consumption, large size, and restrictive use in daily life due to its low controllability).

In conclusion, photostimulation conformable devices can surely bring benefits in numerous health areas. However, this field has been developed only in recent years, when technological breakthrough allowed the preparation of microscale and flexible light sources, and the research on wearable opto-stimulating devices is still in its infancy.

6. Conclusions and future outlook

The skin has a fundamental role in the human body: it is not only the largest organ of the human body, but it also allows humans to interact with the external world, thanks to the continuous exchange of information with the environment. In recent years, skin has been the base to develop new platforms and devices, creating new generations of technologies now recognized as conformable devices. Such devices perfectly adhere to non-planar surfaces such as the human body, paving the way for substituting the rigid and bulk devices now used especially in clinical practice with flexible and stretchable platforms. Such devices have advantages concerning: the possibility to follow strains and movements of the human body because of their similar mechanical properties to those of the skin; their imperceptibility; low cost due to innovative fabrication technologies; the possibility of a continuous use of such devices. Hence, many of them found their use in healthcare or as human-machine interfaces. In particular, we extensively reviewed conformable devices able to provide some kind of stimulation to the human body, their working principles, the materials they comprise, and their fabrication. We decided to focus on three categories based on the method used to generate feedback – thermal, electrical, or mechanical stimulation – since these three categories require different design approaches due to their different purposes. In the last section, we also reviewed the most innovative conformable photostimulation devices and discussed their promising application in healthcare. We highlighted the recent steps forward research has made in new material explorations, fabrication techniques for the production of increasingly cost-effective devices, and innovative powering systems: all these factors contribute to creating more and more wearable, comfortable, conformable, and disposable devices. We also outlined the most important parameters (*Tables*) to consider and to assess during the investigations concerning new materials for developing more and more effective devices.

What has emerged from the reviewed devices is that for the three categories the number of conformable devices already developed is high and, in particular, we have found that the demand for such devices is particularly growing in the medical field. Thermal stimulation provided by conformable heaters find use in personal treatment management and, so, in thermotherapy applications against muscular disorders or in controlled therapeutic drug delivery. At the same time, electrical stimulation can help patients with neurological impairments or for tremor suppression. Also, haptic feedback provided through mechanical stimulation could help in developing tactile displays for blind people. In addition, in this review, we discussed many other applications of conformable stimulating devices. Conformable devices capable of providing transcutaneous electrical stimulation through electrodes applied on the surface of the skin, which are the ones that present more difficulties in their development. So far, such conformable devices are mainly implantable. We think this aspect correlates with the difficulties concerning the development of conformable devices that have low impedance at the interface but also concerning the generation of a uniform electric field. Conformable devices used in electrical stimulation must have only the electrodes in contact with the skin in order to stimulate just the interested point: this translates in a complex fabrication process to manufacture multilevel interconnected layers.

In particular for the medical field, we have noticed the importance of developing portable, disposable, but also customizable devices. One of the main problems is that most of the devices considered required high driving voltage and, so, high power-consumption. This leads to the addition of multiple electronic components on the platform, necessary for the operating principle of such devices, but also brings severe limitations for wearability. Hence, further investigations are necessary aiming to obtain less bulky platforms: basing device power supply on wireless technologies could be a disruptive strategy, as already proposed by some of the studies here reviewed. Also, one can investigate and exploit technologies and materials concerning energy storage and energy harvesting: triboelectric generators added to the platform can constitute self-powered devices, transducing mechanical signals such as movements of the human body into electrical energy. However, conformable stimulation devices could benefit from wireless communication, in addition to miniaturization of electronic components, also because it could guarantee a continuous exchange of data between the device mounted on the skin, and the user. The latter could indeed directly control the temperature reached on the skin using the heating module or the stimulation parameters for transcutaneous electrical stimulation, simply from his smartphone. A more complete vision of such platforms could be to develop a closed-loop control based on feedback signals provided directly from the mounted device, in order to have a system capable of auto-regulating thermal, electrical, or mechanical stimulation parameters on the basis of the data transmitted.

The long-term perspective concerns the use of these devices in clinical practice, in order to replace rigid and bulk electrodes and to reach a level of complexity in controlling these devices, which also suits the capabilities of non-specialized users. More in-depth studies concerning both the materials and fabrication techniques, which allow simplifying the application of the device on the skin, are necessary. Materials investigation and materials engineering are of paramount importance to achieve thinner stimulating systems (in the order of microns), and thus develop bendable and flexible devices that perfectly adhere to all the pores and imperfections of the human skin. Moreover, the miniaturization of different components such as sensors or communication modules for wireless connections and power supply is of crucial importance. This would allow all the electronic components in the stimulating device to be directly embeded so as to transfer all the electronics to the device. This aspect is important for such platforms to be used by individuals themselves at home, making their use in a non-clinical environment also possible.

In conclusion, we presented numerous systems capable of delivering thermal, electrical, mechanical, or optical stimulation, which could be valid platforms to exploit in practical usage. However, all the considered studies were conducted on bench-scale experiments: their use by individuals at home can only be achieved after careful clinical tests, in order to assess the efficacy and effectiveness of the stimulating systems. In doing so, collaborations between specialists like engineers, medical specialists, and chemists will be fundamental to improve the field of soft and conformable electronics.

Conflicts of interest

The authors declare no conflict of interests.

References

1 A. Zucca, C. Cipriani, Sudha, S. Tarantino, D. Ricci, V. Mattoli and F. Greco, *Adv. Healthcare Mater.*, 2015, **4**, 983–990.
2 S. C. Chang, J. Bharathan, Y. Yang, R. Helgeson, F. Wudl, M. B. Ramey and J. R. Reynolds, *Appl. Phys. Lett.*, 1998, **73**, 2561–2563.

3 Z. Bao, Y. Feng, A. Dodabalapur, V. R. Raju and A. J. Lovinger, *Chem. Mater.*, 1997, **9**, 1299–1301.
4 S. Chun, D. Grudinin, D. Lee, S. H. Kim, G. R. Yi and I. Hwang, *Chem. Mater.*, 2009, **21**, 343–350.
5 A. Zucca, K. Yamagishi, T. Fujie, S. Takeoka, V. Mattoli and F. Greco, *J. Mater. Chem. C*, 2015, **3**, 6539–6548.
6 Z. Fan, J. C. Ho, T. Takahashi, R. Yerushalmi, K. Takei, A. C. Ford, Y. L. Chueh and A. Javey, *Adv. Mater.*, 2009, **21**, 3730–3743.
7 J. A. Rogers, T. Someya and Y. Huang, *Science*, 2010, **327**, 1603–1607.
8 D. T. Simon, E. O. Gabrielsson, K. Tybrandt and M. Berggren, *Chem. Rev.*, 2016, **116**, 13009–13041.
9 L. M. Ferrari, S. Sudha, S. Tarantino, R. Esposti, F. Bolzoni, P. Cavallari, C. Cipriani, V. Mattoli and F. Greco, *Adv. Sci.*, 2018, **5**, 1–11.
10 L. M. Ferrari, U. Ismailov, J. M. Badier, F. Greco and E. Ismailova, *npj Flex. Electron.*, 2020, **4**, 1–9.
11 M. Wang, P. Baek, A. Akbarinejad, D. Barker and J. Travas-Sejdic, *J. Mater. Chem. C*, 2019, **7**, 5534–5552.
12 D. A. May-Arrioja, V. I. Ruiz-Perez, Y. Bustos-Terrones and M. A. Basurto-Pensado, *IEEE Sens. J.*, 2016, **16**, 1956–1961.
13 J. Kim, M. Lee, H. J. Shim, R. Ghaffari, H. R. Cho, D. Son, Y. H. Jung, M. Soh, C. Choi, S. Jung, K. Chu, D. Jeon, S. T. Lee, J. H. Kim, S. H. Choi, T. Hyeon and D. H. Kim, *Nat. Commun.*, 2014, **5**, 1–11.
14 A. J. Bandodkar, V. W. S. Hung, W. Jia, G. Valdés-Ramírez, J. R. Windmiller, A. G. Martinez, J. Ramírez, G. Chan, K. Kerman and J. Wang, *Analyst*, 2013, **138**, 123–128.
15 Y. Y. Hsu, H. James, R. Ghaffari, B. Ives, P. Wei, L. Klinker, B. Morey, B. Elolampi, D. Davis, C. Rafferty and K. Dowling, Proc. Tech. Pap. - Int. Microsystems, Packag. Assem. Circuits Technol. Conf. IMPACT, 2012, 228–231.
16 S. Taccola, F. Greco, A. Zucca, C. Innocenti, C. De Julián Fernández, G. Campo, C. Sangregorio, B. Mazzolai and V. Mattoli, *ACS Appl. Mater. Interfaces*, 2013, **5**, 6324–6332.
17 R. Zhou, P. Li, Z. Fan, D. Du and J. Ouyang, *J. Mater. Chem. C*, 2017, **5**, 1544–1551.
18 A. D. Koutsou, J. C. Moreno, A. J. Del Ama, E. Rocon and J. L. Pons, *J. Neuroeng. Rehabil.*, 2016, **13**, 56.
19 F. B. Wagner, J. B. Mignardot, C. G. Le Goff-Mignardot, R. Demesmaeker, S. Komi, M. Capogrosso, A. Rowald, I. Seáñez, M. Caban, E. Pirondini, M. Vat, L. A. McCracken, R. Heimgartner, I. Fodor, A. Watrin, P. Seguin, E. Paoles, K. Van Den Keybus, G. Eberle, B. Schurch, E. Pralong, F. Becce, J. Prior, N. Buse, R. Buschman, E. Neufeld, N. Kuster, S. Carda, J. von Zitzewitz, V. Delattre, T. Denison, H. Lambert, K. Minassian, J. Bloch and G. Courtine, *Nature*, 2018, **563**, 65–93.
20 J. Tian, R. Shi, Z. Liu, H. Ouyang, M. Yu, C. Zhao, Y. Zou, D. Jiang, J. Zhang and Z. Li, *Nano Energy*, 2019, **59**, 705–714.
21 N. Lu and D. H. Kim, *Soft Robot.*, 2014, **1**, 53–62.
22 P. Lopes, S. You, L. P. Cheng, S. Marwecki and P. Baudisch, Conf. Hum. Factors Comput. Syst. - Proc., 2017, 2017-May, 1471–1482.
23 D. Gao, K. Parida and P. S. Lee, *Adv. Funct. Mater.*, 2020, **30**, 1–30.
24 W. Dong, Y. Wang, Y. Zhou, Y. Bai, Z. Ju, J. Guo, G. Gu, K. Bai, G. Ouyang, S. Chen, Q. Zhang and Y. A. Huang, *Int. J. Intell. Robot. Appl.*, 2018, **2**, 313–338.
25 A. Nag, S. C. Mukhopadhyay and J. Kosel, *IEEE Sens. J.*, 2017, **17**, 3949–3960.
26 M. Amjadi, K. U. Kyung, I. Park and M. Sitti, *Adv. Funct. Mater.*, 2016, **26**, 1678–1698.
27 Z. W. K. Low, Z. Li, C. Owh, P. L. Chee, E. Ye, D. Kai, D. P. Yang and X. J. Loh, *Small*, 2019, **15**, 1–23.
28 M. L. Cohen, *J. Invest. Dermatol.*, 1977, **69**, 333–338.
29 D. H. Kim, N. Lu, R. Ma, Y. S. Kim, R. H. Kim, S. Wang, J. Wu, S. M. Won, H. Tao, A. Islam, K. J. Yu, T. Il Kim, R. Chowdhury, M. Ying, L. Xu, M. Li, H. J. Chung, H. Keum, M. McCormick, P. Liu, Y. W. Zhang, F. G. Omenetto, Y. Huang, T. Coleman and J. A. Rogers, *Science*, 2011, **333**, 838–843.
30 S. Biswas and Y. Visell, *Adv. Mater. Technol.*, 2019, **4**, 1–30.
31 K. D. Harris, A. L. Elias and H. J. Chung, *J. Mater. Sci.*, 2016, **51**, 2771–2805.
32 Y. Liu, M. Pharr and G. A. Salvatore, *ACS Nano*, 2017, **11**, 9614–9635.
33 X. Chen, *Small Methods*, 2017, **1**, 1600029.
34 Y. H. Jung, J. U. Kim, J. S. Lee, J. H. Shin, W. Jung, J. Ok and T. il Kim, *Adv. Mater.*, 2020, **32**, 1–25.
35 B. Wang and A. Facchetti, *Adv. Mater.*, 2019, **31**, 1–53.
36 J. C. Yang, J. Mun, S. Y. Kwon, S. Park, Z. Bao and S. Park, *Adv. Mater.*, 2019, **31**, 1–50.
37 M. L. Hammock, A. Chortos, B. C. K. Tee, J. B. H. Tok and Z. Bao, *Adv. Mater.*, 2013, **25**, 5997–6038.
38 J. S. Knutson, N. S. Makowski, K. L. Kilgore and J. Chae, *Neuromuscular Electrical Stimulation Applications*, Elsevier Inc., Fifth Edn, 2019.
39 J. Park, Y. Lee, H. Lee and H. Ko, *ACS Nano*, 2020, **14**, 12–20.
40 D. Bin Moon, J. Lee, E. Roh and N. E. Lee, *Thin Solid Films*, 2019, **688**, 137435.
41 D. F. Fernandes, C. Majidi and M. Tavakoli, *J. Mater. Chem. C*, 2019, **7**, 14035–14068.
42 S. H. Park, Y. J. Kang and S. Majd, *Adv. Mater.*, 2015, **27**, 7583–7619.
43 T. Yokota, P. Zalar, M. Kaltenbrunner, H. Jinno, N. Matsuhisa, H. Kitanosako, Y. Tachibana, W. Yukita, M. Koizumi and T. Someya, *Sci. Adv.*, 2016, **2**, 1–9.
44 D. Yin, J. Feng, R. Ma, Y. F. Liu, Y. L. Zhang, X. L. Zhang, Y. G. Bi, Q. D. Chen and H. B. Sun, *Nat. Commun.*, 2016, **7**, 1–7.
45 M. Kaltenbrunner, T. Sekitani, J. Reeder, T. Yokota, K. Kuribara, T. Tokuhara, M. Drack, R. Schwödiauer, I. Graz, S. Bauer-Gogonea, S. Bauer and T. Someya, *Nature*, 2013, **499**, 458–463.
46 J. Barsotti, I. Hirata, F. Pignatelli, M. Caironi, F. Greco and V. Mattoli, *Adv. Electron. Mater.*, 2018, **4**, 1–9.
47 J. C. McDonald and G. M. Whitesides, *Acc. Chem. Res.*, 2002, **35**, 491–499.

48 C. P. Constantin, M. Aflori, R. F. Damian and R. D. Rusu, *Materials*, 2019, **12**, 3166.
49 R. Balint, N. J. Cassidy and S. H. Cartmell, *Acta Biomater.*, 2014, **10**, 2341–2353.
50 N. Hadrup, A. K. Sharma and K. Loeschner, *Regul. Toxicol. Pharmacol.*, 2018, **98**, 257–267.
51 M. Fedel, *C—J. Carbon Res.*, 2020, **6**, 12.
52 T. P. Haider, C. Völker, J. Kramm, K. Landfester and F. R. Wurm, *Angew. Chem., Int. Ed.*, 2019, **58**, 50–62.
53 E. Aghion, *Metals*, 2018, **8**, 10–13.
54 E. K. Lee, M. K. Kim and C. H. Lee, *Annu. Rev. Biomed. Eng.*, 2019, **21**, 299–323.
55 B. Xu, A. Akhtar, Y. Liu, H. Chen, W. H. Yeo, S. Park, B. Boyce, H. Kim, J. Yu, H. Y. Lai, S. Jung, Y. Zhou, J. Kim, S. Cho, Y. Huang, T. Bretl and J. A. Rogers, *Adv. Mater.*, 2016, **28**, 4462–4471.
56 S. Choi, S. I. Han, D. Jung, H. J. Hwang, C. Lim, S. Bae, O. K. Park, C. M. Tschabrunn, M. Lee, S. Y. Bae, J. W. Yu, J. H. Ryu, S. W. Lee, K. Park, P. M. Kang, W. B. Lee, R. Nezafat, T. Hyeon and D. H. Kim, *Nat. Nanotechnol.*, 2018, **13**, 1048–1056.
57 M. L. Cohen, *J. Invest. Dermatol.*, 1977, **69**, 333–338.
58 J. Kasper, D. L. Fauci, A. S. Hauser, S. L. Longo, D. L. 1. Jameson and J. L. Loscalzo, *Harrison's principles of internal medicine*, 19th edn, New York, McGraw Hill Education, 2015.
59 J. W. Jeong, M. K. Kim, H. Cheng, W. H. Yeo, X. Huang, Y. Liu, Y. Zhang, Y. Huang and J. A. Rogers, *Adv. Healthcare Mater.*, 2014, **3**, 642–648.
60 J. Kim, I. Jeerapan, J. R. Sempionatto, A. Barfidokht, R. K. Mishra, A. S. Campbell, L. J. Hubble and J. Wang, *Acc. Chem. Res.*, 2018, **51**, 2820–2828.
61 A. S. Campbell, J. Kim and J. Wang, *Curr. Opin. Electrochem.*, 2018, **10**, 126–135.
62 J. Kim, I. Jeerapan, S. Imani, T. N. Cho, A. Bandodkar, S. Cinti, P. P. Mercier and J. Wang, *ACS Sens.*, 2016, **1**, 1011–1019.
63 E. R. Lee, T. R. Wilsey, P. Tarczy-Homoch, D. S. Kapp, P. Fessenden, A. Lohrbach and S. D. Prionas, *IEEE Trans. Biomed. Eng.*, 1992, **39**, 470–483.
64 B. Stephens-Fripp, V. Sencadas, R. Mutlu and G. Alici, *Front. Bioeng. Biotechnol.*, 2018, **6**, 1–9.
65 R. G. Gordon, *MRS Bull.*, 2000, **25**, 52–57.
66 H. S. Jo, S. An, J. G. Lee, H. G. Park, S. S. Al-Deyab, A. L. Yarin and S. S. Yoon, *NPG Asia Mater.*, 2017, **9**, e347.
67 S. Selkowitz, *Transparent heat mirrors for passive solar heating applications*, California Univ., Lawrence Berkeley Lab., Berkeley, USA, 1978.
68 R. Hu, Y. Liu, S. Shin, S. Huang, X. Ren, W. Shu, J. Cheng, G. Tao, W. Xu, R. Chen and X. Luo, *Adv. Energy Mater.*, 2020, **10**, 1–23.
69 S. F. Nadler, K. Weingand and R. J. Kruse, *Pain Physician*, 2004, **7**, 395–399.
70 S. Michlovitz, L. Hun, G. N. Erasala, D. A. Hengehold and K. W. Weingand, *Arch. Phys. Med. Rehabil.*, 2004, **85**, 1409–1416.
71 R. B. Roemer, *Annu. Rev. Biomed. Eng.*, 1999, 347–376.
72 J. S. Petrofsky, M. Laymon and H. Lee, *Med. Sci. Monit.*, 2013, **19**, 661–667.
73 D. Son, J. Lee, S. Qiao, R. Ghaffari, J. Kim, J. E. Lee, C. Song, S. J. Kim, D. J. Lee, S. W. Jun, S. Yang, M. Park, J. Shin, K. Do, M. Lee, K. Kang, C. S. Hwang, N. Lu, T. Hyeon and D. H. Kim, *Nat. Nanotechnol.*, 2014, **9**, 397–404.
74 G. Chernyshov, K. Ragozin, C. Caremel and K. Kunze, *Proc. ACM Symp. Virtual Real. Softw. Technol. VRST*, 2018, 52.
75 S. Kim, T. Kim, C. S. Kim, H. Choi, Y. J. Kim, G. S. Lee, O. Oh and B. J. Cho, *Soft Robot.*, 2020, **00**, 1–7.
76 A. Tamayol, A. Hassani Najafabadi, P. Mostafalu, A. K. Yetisen, M. Commotto, M. Aldhahri, M. S. Abdel-Wahab, Z. I. Najafabadi, S. Latifi, M. Akbari, N. Annabi, S. H. Yun, A. Memic, M. R. Dokmeci and A. Khademhosseini, *Sci. Rep.*, 2017, **7**, 1–10.
77 S. W. Hwang, H. Tao, D. H. Kim, H. Cheng, J. K. Song, E. Rill, M. A. Brenckle, B. Panilaitis, S. M. Won, Y. S. Kim, Y. M. Song, K. J. Yu, A. A. Ameen, R. Li, Y. Su, M. Yang, D. L. Kaplan, M. R. Zakin, M. J. Slepian, Y. Huang, F. G. Omenetto and J. A. Rogers, *Science*, 2012, **337**, 1640–1644.
78 J. Koo, M. R. MacEwan, S. K. Kang, S. M. Won, M. Stephen, P. Gamble, Z. Xie, Y. Yan, Y. Y. Chen, J. Shin, N. Birenbaum, S. Chung, S. B. Kim, J. Khalifeh, D. V. Harburg, K. Bean, M. Paskett, J. Kim, Z. S. Zohny, S. M. Lee, R. Zhang, K. Luo, B. Ji, A. Banks, H. M. Lee, Y. Huang, W. Z. Ray and J. A. Rogers, *Nat. Med.*, 2018, **24**, 1830–1836.
79 J. M. Kim, D. R. Oh, J. Sanchez, S. H. Kim and J. M. Seo, *Proc. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. EMBS*, 2013, 1716–1719.
80 P. Kassanos, F. Seichepine, D. Wales and G. Z. Yang, *BioCAS 2019 - Biomed. Circuits Syst. Conf. Proc.*, 2019, 1–4.
81 A. Stier, E. Halekote, A. Mark, S. Qiao, S. Yang, K. Diller and N. Lu, *Micromachines*, 2018, **9**, 1–14.
82 T. W. Kang, S. H. Kim, M. Kim, E. Cho and S. J. Lee, *Plasma Process. Polym.*, 2020, **17**, e1900188.
83 M. D. Dickey, R. C. Chiechi, R. J. Larsen, E. A. Weiss, D. A. Weitz and G. M. Whitesides, *Adv. Funct. Mater.*, 2008, **18**, 1097–1104.
84 L. Jing and Y. Liting, *Liquid Metal Biomaterials*, Springer, Singapore, 2018, pp. 149–186.
85 Y. G. Park, H. S. An, J. Y. Kim and J. U. Park, *Sci. Adv.*, 2019, **5**, 1–10.
86 J. Yang, W. Cheng and K. Kalantar-Zadeh, *Proc. IEEE*, 2019, **107**, 2168–2184.
87 J. Wang, H. Wang, N. V. Thakor and C. Lee, *ACS Nano*, 2019, **13**, 3589–3599.
88 Y. Wang, Z. Yu, G. Mao, Y. Liu, G. Liu, J. Shang, S. Qu, Q. Chen and R. W. Li, *Adv. Mater. Technol.*, 2019, **4**, 1–9.
89 M. G. Kim, H. Alrowais, S. Pavlidis and O. Brand, *Adv. Funct. Mater.*, 2017, **27**, 1604466.
90 R. Guo, X. Sun, S. Yao, M. Duan, H. Wang, J. Liu and Z. Deng, *Adv. Mater. Technol.*, 2019, **4**, 1–11.
91 X. Wang, L. Fan, J. Zhang, X. Sun, H. Chang, B. Yuan, R. Guo, M. Duan and J. Liu, *Adv. Funct. Mater.*, 2019, **29**, 1–13.

92 D. X. Song, Y. Z. Du, W. G. Ma and X. Zhang, *Int. J. Heat Mass Transf.*, 2020, **147**, 119014.
93 S. Sorel, D. Bellet and J. N. Coleman, *ACS Nano*, 2014, **8**, 4805–4814.
94 L. Veeramuthu, B. Y. Chen, C. Y. Tsai, F. C. Liang, M. Venkatesan, D. H. Jiang, C. W. Chen, X. Cai and C. C. Kuo, *RSC Adv.*, 2019, **9**, 35786–35796.
95 J. Kang, H. Kim, K. S. Kim, S. K. Lee, S. Bae, J. H. Ahn, Y. J. Kim, J. B. Choi and B. H. Hong, *Nano Lett.*, 2011, **11**, 5154–5158.
96 D. T. Papanastasiou, A. Schultheiss, D. Muñoz-Rojas, C. Celle, A. Carella, J. P. Simonato and D. Bellet, *Adv. Funct. Mater.*, 2020, **30**, 1–33.
97 S. Hong, H. Lee, J. Lee, J. Kwon, S. Han, Y. D. Suh, H. Cho, J. Shin, J. Yeo and S. H. Ko, *Adv. Mater.*, 2015, **27**, 4744–4751.
98 P. Won, J. J. Park, T. Lee, I. Ha, S. Han, M. Choi, J. Lee, S. Hong, K. J. Cho and S. H. Ko, *Nano Lett.*, 2019, **19**, 6087–6096.
99 J. Jang, B. G. Hyun, S. Ji, E. Cho, B. W. An, W. H. Cheong and J.-U. Park, *NPG Asia Mater.*, 2017, **9**, e432–e432.
100 B. W. An, E. J. Gwak, K. Kim, Y. C. Kim, J. Jang, J. Y. Kim and J. U. Park, *Nano Lett.*, 2016, **16**, 471–478.
101 D. Han, Y. Li, X. Jiang, W. Zhao, F. Wang, W. Lan, E. Xie and W. Han, *Compos. Sci. Technol.*, 2018, **168**, 460–466.
102 S. Choi, J. Park, W. Hyun, J. Kim, J. Kim, Y. B. Lee, C. Song, H. J. Hwang, J. H. Kim, T. Hyeon and D. H. Kim, *ACS Nano*, 2015, **9**, 6626–6633.
103 W. Lan, Y. Chen, Z. Yang, W. Han, J. Zhou, Y. Zhang, J. Wang, G. Tang, Y. Wei, W. Dou, Q. Su and E. Xie, *ACS Appl. Mater. Interfaces*, 2017, **9**, 6644–6651.
104 M. Cao, M. Wang, L. Li, H. Qiu and Z. Yang, *ACS Appl. Mater. Interfaces*, 2018, **10**, 1077–1083.
105 J. C. Goak, T. Y. Kim, D. U. Kim, K. S. Chang, C. S. Lee and N. Lee, *Appl. Surf. Sci.*, 2020, **510**, 145445.
106 J. Ahn, J. Gu, Y. Jeong, K. Kim, J. Jeong and I. Park, *IEEE 32nd Int. Conf. Micro Electro Mech. Syst.*, 2019, 303–306.
107 K. Hee Pyo and J. W. Kim, *Compos. Sci. Technol.*, 2016, **133**, 7–14.
108 J. Park, D. Han, S. Choi, Y. Kim and J. Kwak, *RSC Adv.*, 2019, **9**, 5731–5737.
109 X. Chen, S. Nie, W. Guo, F. Fei, W. Su, W. Gu and Z. Cui, *Adv. Electron. Mater.*, 2019, **5**, 1–8.
110 H. Zhai, R. Wang, X. Wang, Y. Cheng, L. Shi and J. Sun, *Nano Res.*, 2016, **9**, 3924–3936.
111 J. Koo, C. Lee, C. R. Chu, S. K. Kang and H. M. Lee, *Adv. Mater. Technol.*, 2020, **5**, 1–7.
112 L. Yang, W. Weng, J. Yang, Y. Zhang, Y. Liang, X. Luo and M. Zhu, *ACS Appl. Nano Mater.*, 2018, **1**, 4781–4787.
113 R. Kolisnyk, M. Korab, M. Iurzhenko, O. Masiuchok, A. Shadrin, Y. Mamunya, S. Pruvost and V. Demchenko, *Advances in Thin Films, Nanostructured Materials, and Coatings*, Springer, Singapore, 2019, pp. 215–224.
114 S. Kim, W. Oh, J. S. Bae, S. B. Yang, J. H. Yeum, J. Park, C. Sung, J. Kim and J. C. Shin, *Int. J. Polym. Sci.*, 2019, **2019**, 3478325.
115 B. Zhou, X. Han, L. Li, Y. Feng, T. Fang, G. Zheng, B. Wang, K. Dai, C. Liu and C. Shen, *Compos. Sci. Technol.*, 2019, **183**, 107796.
116 N. M. Abbasi, H. Yu, L. Wang, Z. Ul-Abdin, W. A. Amer, M. Akram, H. Khalid, Y. Chen, M. Saleem, R. Sun and J. Shan, *Mater. Chem. Phys.*, 2015, **166**, 1–15.
117 D. Il Moon, G. Plečkaitytė, T. Choi, M. L. Seol, B. Kim, D. Lee, J. W. Han and M. Meyyappan, *Adv. Healthcare Mater.*, 2020, **9**, 1–7.
118 C. Weng, Z. Dai, G. Wang, L. Liu and Z. Zhang, *ACS Appl. Mater. Interfaces*, 2019, **11**, 6541–6549.
119 L. J. Romasanta, P. Schäfer and J. Leng, *Sci. Rep.*, 2018, **8**, 1–7.
120 F. Greco, A. Zucca, S. Taccola, B. Mazzolai and V. Mattoli, *ACS Appl. Mater. Interfaces*, 2013, **5**, 9461–9469.
121 J. Chang, J. He, Q. Lei and D. Li, *ACS Appl. Mater. Interfaces*, 2018, **10**, 19116–19122.
122 C. Wang, K. Sun, J. Fu, R. Chen, M. Li, Z. Zang, X. Liu, B. Li, H. Gong and J. Ouyang, *Adv. Sustainable Syst.*, 2018, **2**, 1800085.
123 H. He, L. Zhang, X. Guan, H. Cheng, X. Liu, S. Yu, J. Wei and J. Ouyang, *ACS Appl. Mater. Interfaces*, 2019, **11**, 26185–26193.
124 M. N. Gueye, A. Carella, R. Demadrille and J. P. Simonato, *ACS Appl. Mater. Interfaces*, 2017, **9**, 27250–27256.
125 F. Greco, A. Zucca, S. Taccola, A. Menciassi, T. Fujie, H. Haniuda, S. Takeoka, P. Dario and V. Mattoli, *Soft Matter*, 2011, **7**, 10642–10650.
126 S. Taccola, F. Greco, E. Sinibaldi, A. Mondini, B. Mazzolai and V. Mattoli, *Adv. Mater.*, 2015, **27**, 1668–1675.
127 E. Ridolfi Riva, A. Desii, E. Sinibaldi, G. Ciofani, V. Piazza, B. Mazzolai and V. Mattoli, *ACS Nano*, 2014, **8**, 5552–5563.
128 X. Wang, W. Yao, R. Guo, X. Yang, J. Tang, J. Zhang, W. Gao, V. Timchenko and J. Liu, *Adv. Healthcare Mater.*, 2018, **7**, 1–10.
129 J. H. Bahk, H. Fang, K. Yazawa and A. Shakouri, *J. Mater. Chem. C*, 2015, **3**, 10362–10374.
130 J. Lee, H. Sul, W. Lee, K. R. Pyun, I. Ha, D. Kim, H. Park, H. Eom, Y. Yoon, J. Jung, D. Lee and S. H. Ko, *Adv. Funct. Mater.*, 2020, **30**, 1–11.
131 D. Beretta, A. J. Barker, I. Maqueira-Albo, A. Calloni, G. Bussetti, G. Dell'Erba, A. Luzio, L. Duò, A. Petrozza, G. Lanzani and M. Caironi, *ACS Appl. Mater. Interfaces*, 2017, **9**, 18151–18160.
132 H. N. Ho, *Temperature*, 2018, **5**, 36–55.
133 Z. Ma, S. Kang, J. Ma, L. Shao, A. Wei, C. Liang, J. Gu, B. Yang, D. Dong, L. Wei and Z. Ji, *ACS Nano*, 2019, **13**, 7578–7590.
134 R. C. Webb, A. P. Bonifas, A. Behnaz, Y. Zhang, K. J. Yu, H. Cheng, M. Shi, Z. Bian, Z. Liu, Y. S. Kim, W. H. Yeo, J. S. Park, J. Song, Y. Li, Y. Huang, A. M. Gorbach and J. A. Rogers, *Nat. Mater.*, 2013, **12**, 938–944.
135 Y. Cheng, H. Zhang, R. Wang, X. Wang, H. Zhai, T. Wang, Q. Jin and J. Sun, *ACS Appl. Mater. Interfaces*, 2016, **8**, 32925–32933.
136 A. Margaret, M. D. Hamburg and F. S. Collins, *N. Engl. J. Med.*, 2010, **363**, 301–304.

137 N. S. Jang, K. H. Kim, S. H. Ha, S. H. Jung, H. M. Lee and J. M. Kim, *ACS Appl. Mater. Interfaces*, 2017, **9**, 19612–19621.
138 S. S. Gharaie, S. M. H. Dabiri and M. Akbari, *Polymers*, 2018, **10**, 1317.
139 J. H. Park, J. W. Lee, Y. C. Kim and M. R. Prausnitz, *Int. J. Pharm.*, 2008, **359**, 94–103.
140 M. R. Prausnitz, S. Mitragotri and R. Langer, *Nat. Rev. Drug Discovery*, 2004, **3**, 115–124.
141 S. Bagherifard, A. Tamayol, P. Mostafalu, M. Akbari, M. Comotto, N. Annabi, M. Ghaderi, S. Sonkusale, M. R. Dokmeci and A. Khademhosseini, *Adv. Healthcare Mater.*, 2016, **5**, 175–184.
142 H. Lee, C. Song, Y. S. Hong, M. S. Kim, H. R. Cho, T. Kang, K. Shin, S. H. Choi, T. Hyeon and D. H. Kim, *Sci. Adv.*, 2017, **3**, 1–9.
143 L. A. Jones and H. N. Ho, *IEEE Trans. Haptics*, 2008, **1**, 53–70.
144 D. Beretta, A. Perego, G. Lanzani and M. Caironi, *Sustainable Energy Fuels*, 2017, **1**, 174–190.
145 A. V. Roy, J. Camchong and K. O. Lim, *Principles and applications of transcranial electrical stimulation*, Elsevier Inc., 2018.
146 H. Palza, P. A. Zapata and C. Angulo-Pineda, *Materials*, 2019, **12**, 277.
147 C. Galli, G. Pedrazzi and S. Guizzardi, *Bioelectromagnetics*, 2019, **40**, 211–233.
148 E. L. Nussbaum, P. Houghton, J. Anthony, S. Rennie, B. L. Shay and A. M. Hoens, *Physiother. Canada*, 2017, **69**, 1–76.
149 E. D'Anna, F. M. Petrini, F. Artoni, I. Popovic, I. Simanić, S. Raspopovic and S. Micera, *Sci. Rep.*, 2017, **7**, 1–15.
150 P. E. Houghton, *Chronic Wound Care Manag. Res.*, 2017, **4**, 25–44.
151 S. D. Bennie, J. S. Petrofsky, J. Nisperos, M. Tsurudome and M. Laymon, *Eur. J. Appl. Physiol.*, 2002, **88**, 13–19.
152 L. E. Osborn, A. Dragomir, J. L. Betthauser, C. L. Hunt, H. H. Nguyen, R. R. Kaliki and N. V. Thakor, *Sci. Robot.*, 2018, **3**, eaat3818.
153 X. Wang, Y. Zhang, R. Guo, H. Wang, B. Yuan and J. Liu, *J. Micromechanics Microengineering*, 2018, **28**, aaa80f.
154 W. Navaraj, C. Smith and R. Dahiya, *E-skin and wearable systems for health care*, Elsevier Ltd., 2020.
155 J. C. Loitz, A. Reinert, A. K. Neumann, F. Quandt, D. Schroeder and W. H. Krautschneider, *Curr. Dir. Biomed. Eng*, 2016, **2**, 391–394.
156 H. P. Wang, A. W. Guo, Z. Y. Bi, F. Li, X. Y. Lu and Z. G. Wang, *Proc. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. EMBS*, 2017, 714–717.
157 L. Z. Popović and N. M. Malešević, *Proc. 31st Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. Eng. Futur. Biomed. EMBC 2009*, 2009, 6785–6788.
158 C. De Marchis, T. S. Monteiro, C. Simon-Martinez, S. Conforto and A. Gharabaghi, *J. Neuroeng. Rehabil.*, 2016, **13**, 1–9.
159 M. Perovic, M. Stevanovic, T. Jevtic, M. Strbac, G. Bijelic, C. Vucetic, L. Popovic-Maneski and D. Popovic, *J. Autom. Control*, 2013, **21**, 13–18.
160 S. Cheng and D. Zhang, *Proc. 10th Int. Conf. on Human System Interactions (HSI)*, 2017, 120–124.
161 T. Štrbac, M. Belić, M. Isaković, M. Kojić, V. Bijelić, G. Popović and I. Keller, *J. Neural Eng.*, 2016, **13**, 046014.
162 H. Zhou, Y. Lu, W. Chen, Z. Wu, H. Zou, L. Krundel and G. Li, *Sensors*, 2015, **15**, 17241–17257.
163 M. Rahimi, F. Jiang and Y. Shen, *IEEE Access*, 2019, **7**, 169844–169852.
164 T. Keller and A. Kuhn, *J. Autom. Control*, 2008, **18**, 35–45.
165 F. Lu, C. Wang, R. Zhao, L. Du, Z. Fang, X. Guo and Z. Zhao, *Biosensors*, 2018, **8**, 31.
166 A. Withana, D. Groeger and J. Steimle, *UIST 2018 - Proc. 31st Annu. ACM Symp. User Interface Softw. Technol.*, 2018, 365–378.
167 M. Ying, A. P. Bonifas, N. Lu, Y. Su, R. Li, H. Cheng, A. Ameen, Y. Huang and J. A. Rogers, *Nanotechnology*, 2012, **23**, 344004.
168 J. Park, S. Choi, A. H. Janardhan, S. Y. Lee, S. Raut, J. Soares, K. Shin, S. Yang, C. Lee, K. W. Kang, H. R. Cho, S. J. Kim, P. Seo, W. Hyun, S. Jung, H. J. Lee, N. Lee, S. H. Choi, M. Sacks, N. Lu, M. E. Josephson, T. Hyeon, D. H. Kim and H. J. Hwang, *Sci. Transl. Med.*, 2016, **8**, 344ra86.
169 C. Kim, H. J. Yang, T. H. Cho, B. S. Lee, T. M. Gwon, S. Shin, I. S. Kim, S. J. Kim and S. J. Hwang, *Med. Biol. Eng. Comput.*, 2020, **58**, 383–399.
170 J. K. Fink, *High Performance Polymers*, Second Edition, Elsevier Inc., 2014, pp. 381–400.
171 J. Jeong, K. S. Min and S. J. Kim, *Microelectron. Eng.*, 2019, **216**, 111096.
172 J. Jeong, S. H. Bae, K. S. Min, J. M. Seo, H. Chung and S. J. Kim, *IEEE Trans. Biomed. Eng.*, 2015, **62**, 982–989.
173 N. Obidin, F. Tasnim and C. Dagdeviren, *Adv. Mater.*, 2020, **32**, 1–26.
174 O. Akhavan, E. Ghaderi, S. A. Shirazian and R. Rahighi, *Carbon N. Y.*, 2016, **97**, 71–77.
175 J. Pfau, D. Ganatra, A. Weltin, G. Urban, J. Kieninger and T. Stieglitz, *Proc. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. EMBS*, 2019, 3762–3765.
176 H. Ouyang, Z. Liu, N. Li, B. Shi, Y. Zou, F. Xie, Y. Ma, Z. Li, H. Li, Q. Zheng, X. Qu, Y. Fan, Z. L. Wang, H. Zhang and Z. Li, *Nat. Commun.*, 2019, **10**, 1–10.
177 D. Y. Park, D. J. Joe, D. H. Kim, H. Park, J. H. Han, C. K. Jeong, H. Park, J. G. Park, B. Joung and K. J. Lee, *Adv. Mater.*, 2017, **29**, 1–9.
178 R. Guo, X. Wang, H. Chang, W. Yu, S. Liang, W. Rao and J. Liu, *Adv. Eng. Mater.*, 2018, **20**, 1–9.
179 S. Park, S. W. Heo, W. Lee, D. Inoue, Z. Jiang, K. Yu, H. Jinno, D. Hashizume, M. Sekino, T. Yokota, K. Fukuda, K. Tajima and T. Someya, *Nature*, 2018, **561**, 516–521.
180 R. David and N. Miki, *Proc. IEEE Int. Conf. Micro Electro Mech. Syst.*, 2018, **2018-Janua**, 373–375.
181 J. Liu and L. Yi, *Liquid Metal Enabled Skin Electronics*, 2018, 255–323.
182 J. Li, C. Guo, Z. Wang, K. Gao, X. Shi and J. Liu, *Clin. Transl. Med.*, 2016, **5**, 21.

183 M. Leandri, L. Marinelli, A. Siri and L. Pellegrino, *J. Neurosci. Methods*, 2018, **293**, 17–26.
184 Y. Wei, K. Yang, M. Browne, L. Bostan and P. Worsley, *IEEE Sensors Lett.*, 2019, **3**, 2017–2020.
185 M. Tezuka, N. Kitamura and N. Miki, *Jpn. J. Appl. Phys.*, 2016, **55**, 6S1.
186 M. Tezuka, K. Ishimaru and N. Miki, *Sens. Actuators, A*, 2017, **258**, 32–38.
187 C. V. Keef, L. V. Kayser, S. Tronboll, C. W. Carpenter, N. B. Root, M. Finn, T. F. O'Connor, S. N. Abuhamdieh, D. M. Davies, R. Runser, Y. S. Meng, V. S. Ramachandran and D. J. Lipomi, *Adv. Intell. Syst.*, 2020, **2**, 2000018.
188 F. H. Parianen Lesemann, E. M. Reuter and B. Godde, *Neurosci. Biobehav. Rev.*, 2015, **51**, 126–137.
189 C. Antfolk, M. D'alonzo, B. Rosén, G. Lundborg, F. Sebelius and C. Cipriani, *Expert Rev. Med. Devices*, 2013, **10**, 45–54.
190 M. R. Mulvey, H. J. Fawkner, H. E. Radford and M. I. Johnson, *Neuromodulation*, 2012, **15**, 42–47.
191 E. D'Anna, G. Valle, A. Mazzoni, I. Strauss, F. Iberite, J. Patton, F. M. Petrini, S. Raspopovic, G. Granata, R. Di Iorio, M. Controzzi, C. Cipriani, T. Stieglitz, P. M. Rossini and S. Micera, *Sci. Robot.*, 2019, **4**, eaau8892.
192 K. A. Kaczmarek, J. G. Webster, P. Bach-y-Rita and W. J. Tompkins, *IEEE Trans. Biomed. Eng.*, 1991, **38**, 1–16.
193 D. A. Lake, *Sport. Med. An Int. J. Appl. Med. Sci. Sport Exerc.*, 1992, **13**, 320–336.
194 A. Popović-Bijelić, G. Bijelić, N. Jorgovanović, D. Bojanić, M. B. Popović and D. B. Popović, *Artif. Organs*, 2005, **29**, 448–452.
195 A. A. Guex, A. E. Hight, S. Narasimhan, N. Vachicouras, D. J. Lee, S. P. Lacour and M. C. Brown, *Hear. Res.*, 2019, **377**, 339–352.
196 L. Guo and S. P. Deweerth, *Proc. 31st Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. Eng. Futur. Biomed. EMBC 2009*, 2009, 1623–1626.
197 K. W. Meacham, L. Guo, S. P. DeWeerth and S. Hochman, *Front. Neuroeng.*, 2011, **4**, 1–12.
198 I. R. Minev, P. Musienko, A. Hirsch, Q. Barraud, N. Wenger, E. M. Moraud, J. Gandar, M. Capogrosso, T. Milekovic, L. Asboth, R. F. Torres, N. Vachicouras, Q. Liu, N. Pavlova, S. Duis, A. Larmagnac, J. Vörös, S. Micera, Z. Suo, G. Courtine and S. P. Lacour, *Science*, 2015, **347**, 159–163.
199 G. M. Graham, T. A. Thrasher and M. R. Popovic, *IEEE Trans. Neural Syst. Rehabil. Eng*, 2006, **14**, 38–45.
200 A. Akhtar, J. Sombeck, B. Boyce and T. Bretl, *Sci. Robot.*, 2018, **3**, eaap9770.
201 P. Bach-y-Rita, M. E. Tyler and K. A. Kaczmarek, *Int. J. Hum. Comput. Interact.*, 2003, **15**, 285–295.
202 H. Tang and D. J. Beebe, *J. Microelectromechanical Syst.*, 2003, **12**, 29–36.
203 I. M. Koo, K. Jung, J. C. Koo, J. Do Nam, Y. K. Lee and H. R. Choi, *IEEE Trans. Robot.*, 2008, **24**, 549–558.
204 Y. Zhang, Y. Chen, B. Zhao, Y. Hu and Z. Li, *Proc. IEEE Int. Conf. Micro Electro Mech. Syst.*, 2018, 415–418.
205 Y. Song, J. Min and W. Gao, *ACS Nano*, 2019, **13**, 12280–12286.
206 J. Li, X. Liu, E. Tomaskovic-Crook, J. M. Crook and G. G. Wallace, *Colloids Surf., B*, 2019, **176**, 87–95.
207 N. Akhtar, V. Singh, M. Yusuf, R. A. Khan and R. A. Khan, *Biomed. Tech.*, 2020, **65**, 243–272.
208 J. Heikenfeld, *IEEE Spectr*, 2014, **51**, 46–63.
209 D. Bruen, C. Delaney, L. Florea and D. Diamond, *Sensors*, 2017, **17**, 1–21.
210 A. J. Bandodkar, W. Jia, C. Yard, X. Wang, J. Ramirez and J. Wang, *Anal. Chem.*, 2015, **87**, 394–398.
211 J. Kim, J. R. Sempionatto, S. Imani, M. C. Hartel, A. Barfidokht, G. Tang, A. S. Campbell, P. P. Mercier and J. Wang, *Adv. Sci.*, 2018, **5**, 1800880.
212 P. Nenzi, A. Denzi, K. Kholostov, R. Crescenzi, F. Apollonio, M. Liberti, P. Marracino, A. Ongaro, R. Cadossi and M. Balucani, *Proc. - Electron. Components Technol. Conf.*, 2013, 486–493.
213 W. E. Ford, W. Ren, P. F. Blackmore, K. H. Schoenbach and S. J. Beebe, *Arch. Biochem. Biophys.*, 2010, **497**, 82–89.
214 X. Sun, B. Yuan, W. Rao and J. Liu, *Biomaterials*, 2017, **146**, 156–167.
215 K. Salisbury and C. Tarr, *Am. Soc. Mech. Eng. Dyn. Syst. Control Div. DSC*, 1997, **61**, 61–67.
216 A. U. Alahakone and S. M. N. A. Senanayake, *IEEE/ASME Int. Conf. Adv. Intell. Mechatronics, AIM*, 2009, 1148–1153.
217 S. Jadhav, V. Kannanda, B. Kang, M. T. Tolley and J. P. Schulze, *IST Int. Symp. Electron. Imaging Sci. Technol.*, 2017, 19–24.
218 X. Yu, Z. Xie, Y. Yu, J. Lee, A. Vazquez-Guardado, H. Luan, J. Ruban, X. Ning, A. Akhtar, D. Li, B. Ji, Y. Liu, R. Sun, J. Cao, Q. Huo, Y. Zhong, C. M. Lee, S. Y. Kim, P. Gutruf, C. Zhang, Y. Xue, Q. Guo, A. Chempakasseril, P. Tian, W. Lu, J. Y. Jeong, Y. J. Yu, J. Cornman, C. S. Tan, B. H. Kim, K. H. Lee, X. Feng, Y. Huang and J. A. Rogers, *Nature*, 2019, **575**, 473–479.
219 A. Boem and G. M. Troiano, *DIS 2019 - Proc. 2019 ACM Des. Interact. Syst. Conf.*, 2019, 885–906.
220 A. Haynes, M. F. Simons, T. Helps, Y. Nakamura and J. Rossiter, *IEEE Robot. Autom. Lett.*, 2019, **4**, 1641–1646.
221 M. Watanabe, H. Shirai and T. Hirai, *J. Appl. Phys.*, 2002, **92**, 4631–4637.
222 J. H. Youn, S. M. Jeong, G. Hwang, H. Kim, K. Hyeon, J. Park and K. U. Kyung, *Appl. Sci.*, 2020, **10**, 640.
223 S. Muthukumarana, D. S. Elvitigala, J. P. F. Cortes, D. J. C. Matthies and S. Nanayakkara, *SIGGRAPH Asia 2019 XR, SA*, 2019, **2019**, 29–30.
224 Y. Kato, T. Sekitani, M. Takamiya, M. Doi, K. Asaka, T. Sakurai and T. Someya, *IEEE Trans. Electron Devices*, 2007, **54**, 202–209.
225 H. A. Sonar and J. Paik, *Front. Robot. AI*, 2016, **2**, 1–11.
226 P. Morasso, M. Casadio, V. Sanguineti, V. Squeri, S. Member and E. Vergaro, *2007 Virtual Rehabilitation*, 2007, 70–77.
227 S. Kim, P. Kim, C. Y. Park and S. B. Choi, *Sens. Actuators, A*, 2016, **239**, 61–69.
228 C. Hatzfeld and R. Werthschützky, *Lect. Notes Comput. Sci. (including Subser. Lect. Notes Artif. Intell. Lect. Notes Bioinformatics)*, 2010, **6191 LNCS**, 99–104.

229 M. Raitor, J. M. Walker, A. M. Okamura and H. Culbertson, *Proc. - IEEE Int. Conf. Robot. Autom.*, 2017, 427–432.
230 N. Al-huda Hamdan, A. Wagner, S. Voelker, J. Steimle and J. Borchers, *Conf. Hum. Factors Comput. Syst. - Proc.*, 2019, 488.
231 C. Larson, B. Peele, S. Li, S. Robinson, M. Totaro, L. Beccai, B. Mazzolai and R. Shepherd, *Science*, 2016, **351**, 1071–1074.
232 K. Song, S. H. Kim, S. Jin, S. Kim, S. Lee, J. S. Kim, J. M. Park and Y. Cha, *Sci. Rep.*, 2019, **9**, 1–8.
233 D. J. Lipomi, C. Dhong, C. W. Carpenter, N. B. Root and V. S. Ramachandran, *Adv. Funct. Mater.*, 2020, **30**, 1–15.
234 P. Le Floch, N. Molinari, K. Nan, S. Zhang, B. Kozinsky, Z. Suo and J. Liu, *Nano Lett.*, 2020, **20**, 224–233.
235 H. Zhao, A. M. Hussain, A. Israr, D. M. Vogt, M. Duduta, D. R. Clarke and R. J. Wood, *Soft Robot.*, 2020, **7**, 451–461.
236 S. Mun, S. Yun, S. Nam, S. K. Park, S. Park, B. J. Park, J. M. Lim and K. U. Kyung, *IEEE Trans. Haptics*, 2018, **11**, 15–21.
237 H. Phung, P. T. Hoang, C. T. Nguyen, T. Dat Nguyen, H. Jung, U. Kim and H. R. Choi, *IEEE Int. Conf. Intell. Robot. Syst.*, 2017, **2017**, 886–891.
238 S. Yun, S. Park, B. Park, S. Ryu, S. M. Jeong and K. U. Kyung, *IEEE Trans. Ind. Electron.*, 2020, **67**, 717–724.
239 S. H. Yoon, J. D. Ma, L. Siyuan, T. Woo Suk, S. Shantanu, R. Di and F. P. Holbery, *Proceedings of the 32nd Annual ACM Symposium on User Interface Software and Technology*, 2019, pp. 949–961.
240 C. Dagdeviren, Y. Shi, P. Joe, R. Ghaffari, G. Balooch, K. Usgaonkar, O. Gur, P. L. Tran, J. R. Crosby, M. Meyer, Y. Su, R. C. Webb, A. S. Tedesco, M. J. Slepian, Y. Huang and J. A. Rogers, *Nat. Mater.*, 2015, **14**, 728–736.
241 D. G. Seo and Y. H. Cho, *J. Mech. Sci. Technol.*, 2018, **32**, 631–636.
242 J. Mohd Jani, M. Leary, A. Subic and M. A. Gibson, *Mater. Des.*, 2014, **56**, 1078–1113.
243 J. Mohd Jani, M. Leary and A. Subic, *J. Intell. Mater. Syst. Struct*, 2017, **28**, 1699–1718.
244 R. S. Kularatne, H. Kim, J. M. Boothby and T. H. Ware, *J. Polym. Sci., Part B: Polym. Phys.*, 2017, **55**, 395–411.
245 M. E. Prévôt, S. Ustunel and E. Hegmann, *Materials*, 2018, **11**, 377.
246 N. Torras, K. E. Zinoviev, C. J. Camargo, E. M. Campo, H. Campanella, J. Esteve, J. E. Marshall, E. M. Terentjev, M. Omastová, I. Krupa, P. Teplický, B. Mamojka, P. Bruns, B. Roeder, M. Vallribera, R. Malet, S. Zuffanelli, V. Soler, J. Roig, N. Walker, D. Wenn, F. Vossen and F. M. H. Crompvoets, *Sens. Actuators, A*, 2014, **208**, 104–112.
247 C. Wang, K. Sim, J. Chen, H. Kim, Z. Rao, Y. Li, W. Chen, J. Song, R. Verduzco and C. Yu, *Adv. Mater.*, 2018, **30**, 1–9.
248 N. Torras, K. E. Zinoviev, J. Esteve and A. Sánchez-Ferrer, *J. Mater. Chem. C*, 2013, **1**, 5183–5190.
249 V. S. R. Jampani, R. H. Volpe, K. R. De Sousa, J. F. Machado, C. M. Yakacki and J. P. F. Lagerwall, *Sci. Adv.*, 2019, **5**, eaaw2476.
250 K. O. Johnson, *Curr. Opin. Neurobiol.*, 2001, **11**, 455–461.
251 E. Van Breda, S. Verwulgen, W. Saeys, K. Wuyts, T. Peeters and S. Truijen, *BMJ Open Sport Exerc. Med*, 2017, **3**, 1–12.
252 A. Manuscript, *J. Neurol.*, 2011, **34**, 98–104.
253 C. Cipriani, M. Dalonzo and M. C. Carrozza, *IEEE Trans. Biomed. Eng.*, 2012, **59**, 400–408.
254 G. Frediani, D. Mazzei, D. E. De Rossi and F. Carpi, *Front. Bioeng. Biotechnol.*, 2014, **2**, 1–7.
255 H. Boys, G. Frediani, M. Ghilardi, S. Poslad, J. C. Busfield and F. Carpi, 2018 *IEEE Int. Conf. Soft Robot. RoboSoft 2018*, 2018, 270–275.
256 M. Sarac, A. M. Okamura and M. Di Luca, 2019, arXiv preprint arXiv:1911.08528.
257 G. Moy, U. Singh, E. Tan and R. S. Fearing, *Haptics-e*, 2000, **1**, 1–20.
258 M. Pielot and R. De Oliveira, *Proceed. 15th Int. Conf. on Human-computer interaction with mobile devices and services*, 2013, pp. 1–10.
259 S. H. Yun and S. J. J. Kwok, *Nat. Biomed. Eng.*, 2017, **1**, 1–16.
260 G. H. Lee, H. Moon, H. Kim, G. H. Lee, W. Kwon, S. Yoo, D. Myung, S. H. Yun, Z. Bao and S. K. Hahn, *Nat. Rev. Mater*, 2020, **5**, 149–165.
261 M. Yasumoto, T. Shimoda, R. Tsubouchi and S. Kawana, *Nishinihon J. Dermatology*, 2009, **71**, 201–205.
262 X. Huang and M. A. El-Sayed, *Alexandria J. Med.*, 2011, **47**, 1–9.
263 M. Hennessy and M. R. Hamblin, *J. Opt.*, 2017, **19**, 013003.
264 S. Kellner and S. Berlin, *Appl. Sci.*, 2020, **10**, 805.
265 J. Hopkins, L. Travaglini, A. Lauto, T. Cramer, B. Fraboni, J. Seidel and D. Mawad, *Adv. Mater. Technol.*, 2019, **4**, 1–10.
266 V. K. Samineni, A. D. Mickle, J. Yoon, J. G. Grajales-Reyes, M. Y. Pullen, K. E. Crawford, K. N. Noh, G. B. Gereau, S. K. Vogt, H. H. Lai, J. A. Rogers and R. W. Gereau, *Sci. Rep.*, 2017, **7**, 1–14.
267 L. Via and S. Magno, *Curr. Med. Chem.*, 2001, **8**, 1405–1418.
268 M. Wegener, M. J. Hansen, A. J. M. Driessen, W. Szymanski and B. L. Feringa, *J. Am. Chem. Soc.*, 2017, **139**, 17979–17986.
269 S. G. R. Bade, X. Shan, P. T. Hoang, J. Li, T. Geske, L. Cai, Q. Pei, C. Wang and Z. Yu, *Adv. Mater.*, 2017, **29**, 1–7.
270 H. E. Lee, D. Lee, T. I. Lee, J. H. Shin, G. M. Choi, C. Kim, S. H. Lee, J. H. Lee, Y. H. Kim, S. M. Kang, S. H. Park, I. S. Kang, T. S. Kim, B. S. Bae and K. J. Lee, *Nano Energy*, 2019, **55**, 454–462.
271 H. E. Lee, J. H. Shin, J. H. Park, S. H. Hong, S. K. Park, S. H. Lee, J. H. Lee, I.-S. Kang and K. J. Lee, *Adv. Funct. Mater.*, 2019, **29**, 1808075.
272 J. Kim, G. A. Salvatore, H. Araki, A. M. Chiarelli, Z. Xie, A. Banks, X. Sheng, Y. Liu, J. W. Lee, K. I. Jang, S. Y. Heo, K. Cho, H. Luo, B. Zimmerman, J. Kim, L. Yan, X. Feng, S. Xu, M. Fabiani, G. Gratton, Y. Huang, U. Paik and J. A. Rogers, *Sci. Adv.*, 2016, **2**, e1600418.
273 A. D. Mickle, S. M. Won, K. N. Noh, J. Yoon, K. W. Meacham, Y. Xue, L. A. McIlvried, B. A. Copits, V. K. Samineni, K. E. Crawford, D. H. Kim, P. Srivastava, B. H. Kim, S. Min, Y. Shiuan, Y. Yun, M. A. Payne, J. Zhang,

H. Jang, Y. Li, H. H. Lai, Y. Huang, S. Il Park, R. W. Gereau and J. A. Rogers, *Nature*, 2019, **565**, 361–365.
274 G. Shin, A. M. Gomez, R. Al-Hasani, Y. R. Jeong, J. Kim, Z. Xie, A. Banks, S. M. Lee, S. Y. Han, C. J. Yoo, J. L. Lee, S. H. Lee, J. Kurniawan, J. Tureb, Z. Guo, J. Yoon, S. Il Park, S. Y. Bang, Y. Nam, M. C. Walicki, V. K. Samineni, A. D. Mickle, K. Lee, S. Y. Heo, J. G. McCall, T. Pan, L. Wang, X. Feng, T. il Kim, J. K. Kim, Y. Li, Y. Huang, R. W. Gereau, J. S. Ha, M. R. Bruchas and J. A. Rogers, *Neuron*, 2017, **93**, 509–521.e3.
275 K. Yamagishi, I. Kirino, I. Takahashi, H. Amano, S. Takeoka, Y. Morimoto and T. Fujie, *Nat. Biomed. Eng.*, 2019, **3**, 27–36.
276 S. Il Park, D. S. Brenner, G. Shin, C. D. Morgan, B. A. Copits, H. U. Chung, M. Y. Pullen, K. N. Noh, S. Davidson, S. J. Oh, J. Yoon, K. I. Jang, V. K. Samineni, M. Norman, J. G. Grajales-Reyes, S. K. Vogt, S. S. Sundaram, K. M. Wilson, J. S. Ha, R. Xu, T. Pan, T. Il Kim, Y. Huang, M. C. Montana, J. P. Golden, M. R. Bruchas, R. W. Gereau and J. A. Rogers, *Nat. Biotechnol.*, 2015, **33**, 1280–1286.
277 Y. Jeon, H. R. Choi, K. C. Park and K. C. Choi, *J. Soc. Inf. Disp.*, 2020, **28**, 324–332.
278 D. Han, Y. Khan, J. Ting, S. M. King, N. Yaacobi-Gross, M. J. Humphries, C. J. Newsome and A. C. Arias, *Adv. Mater.*, 2017, **29**, 1606206.
279 H. E. Lee, S. H. Lee, M. Jeong, J. H. Shin, Y. Ahn, D. Kim, S. H. Oh, S. H. Yun and K. J. Lee, *ACS Nano*, 2018, **12**, 9587–9595.

