Flexible and Stretchable Devices for Human-Machine Interfaces

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18.1 Introduction

The discovery of the transistor in the mid-twentieth century has revolutionized modern electronics and transformed the way we live, experience, and interact with the world today. Vast developments in semiconductor materials and fabrication technologies enabled exponential growth of transistor counts and miniaturization of electronic systems, down from logic gates to integrated circuits (IC), as predicted by Moore’s law. With further advances in large-scale integration, compact packaging, high durability, and low-power design of microelectronics, electronic devices with many functionalities have become ubiquitous and embedded in all aspects of our life. They have now been applied to a broad range of applications, from entertainment, health-care, business development, to information technology, all converging to augment the way we interact with digital information and support our daily needs and goals.

The miniaturization, reduced cost, and improved performance of these electronic devices and systems have also sparked various technological breakthroughs in computing interfaces and pushed forward their
Human-computer or human-machine interface (HMI) requires an input mechanism that sends information or requests from a human to the machine. The machine then processes this command and relays feedback in the form of output. This information can be ergonomically communicated through various means, either through visual, auditory, or other cognitive or physical interaction channels (Jaimes and Sebe 2007). As illustrated in Figure 18.1, the first computing devices required simple mechanical switches for user controls. After the appearance of the first series of personal computers in the 1980s, there has been a rapid development in new interface technologies that enable users to interact with the digital world intuitively. Several user interface technologies have become matured and widespread, such as the mouse and keypad set to control a graphical user interface (GUI) in a personal computer, touch-sensitive devices that have become default controls of today’s tablets and smartphones, as well as wearables that actively respond to the activity or physiological state of the users.

Continuous innovation in sensing, display, hardware processing, and software developments around the early 21st century have also triggered advances in technologies for immersive sensory experiences, such as virtual and augmented reality. With haptic feedback, gesture-sensing, voice-recognition devices, and head-mounted displays, these technologies attempt to radically change the way users perceive and interact with digital information. It is done by bringing them into a computer-simulated reality or by overlaying digital information to the physical world respectively. They also aim to reduce users’ memory load and provide intuitive, seamless interaction between human and computers. However, they are still under early developmental stage before becoming universal. Their relatively large size, limited mobility, and in some cases, additional setup for the controllers limit the applications and restrain them from becoming truly wearable and being used continuously. Extensive research in brain-computer interfaces and neuro-prosthetics has also been recently conducted to ultimately accomplish a direct bi-directional control between the nervous system and external devices. Current research trends show various invasive technologies for a high-quality sensing, actuation, and direct access to the neural tissues. Nonetheless, invasive neural interfacing approaches provide several technological challenges. Standard electronic devices are not compatible with the human body, as they are commonly planar, rigid, and several orders of magnitude stiffer than the body tissue.
In this light, recent advances in new materials, device designs, and fabrication strategies have established a new form of soft electronics that are biocompatible and can be flexed and stretched to bridge the biological, geometrical, and mechanical mismatch between electronics and the human body (Rogers 2015). They enable a myriad of novel wearable and implantable applications, from physiological and activity monitoring, physical interaction media, prosthetics, to robotics. As discussed in the previous chapters of this book and illustrated in Figure 18.2, the performance and applications of these flexible and stretchable devices are defined by their material properties, choice of substrates, and fabrication techniques. Most of the state-of-the-art flexible electronics are developed by fabricating (Ahn et al. 2006) or transfer-printing (Meitl et al. 2006) devices on flexible substrates or by thinning down silicon wafers with methods such as dry etching, wet chemical etching, grinding, chemical-mechanical polishing, and exfoliating (Feil et al. 2003; Gumus et al. 2017; Zhai et al. 2012). On the other hand, stretchable electronics are realized by developing intrinsically stretchable materials (Wang et al. 2017), designing serpentine interconnect architecture that allows stretching of rigid structures on an elastomeric substrate (Zhang et al. 2013a), or leveraging buckling mechanisms by pre-stretching an elastomeric substrate (Sun et al. 2006).

To bolster technological translation, using materials and substrates that are imperceptible, transparent, and have self-healing properties is also essential (Benight et al. 2013; Salvatore et al. 2014). Self-healing ability allows certain electronic devices to regain back their mechanical and electrical characteristics upon minor damage and is particularly useful for devices that experience accidental cuts or scratches. In addition, ultra-thin and transparent designs allow intimate integration, enhance comfortability, and support a widespread user acceptance.

This chapter will give an overview of existing flexible and stretchable sensors, actuators, and transducers systems with a focus on their applications for the next generation of HMI. The mechanically adaptive features of these artificial skin, neural implants, and sensory prostheses facilitate conformal and intimate integration to different regions of the human body. Depending on their modalities and locations, as listed in Table 18.1, these collective devices can give insights to various physiological and biomechanical changes of the human body. They can also further realize the ultimate goal of seamless bi-directional communication between machines and the human nervous system. This effort promises to restore and augment our physical and mental capabilities and will radically transform the way we interact with computers in the future.

18.2 On-body Interfaces

The rise of ubiquitous computing and the Internet of Thing has brought us into an era where our physical and digital worlds start to blend, seamlessly coupling without boundaries. However, current wearable devices, mobile phones, and remote displays are still limited in their active surface and accessibility in particular scenarios. Given the fact that human skin and its appendages are easily accessible, intimate to us, and they provide a vital sensory barrier between ourselves and the environments, visions start to emerge of using on-body electronics, or electronics that merge with the body as extensions of self (Leigh et al. 2017). Leveraging our skin as mobile input and output surfaces that are with us and available all the time can result in compelling human-computer interaction (HCI) applications (Steimle 2016). This vision is further supported by the current developments of flexible and stretchable devices, which are light-weight, seamless, aesthetically pleasing, and compatible with the skin, making them comfortable and socially acceptable to everyone.

18.2.1 On-body Inputs

Several techniques have been adapted for the development of touch panels, including capacitive, resistive, surface acoustic wave, and infrared-based sensing. Capacitive and resistive methods, in particular, are widely used for on-body applications. Compared to the others, these methods are more approachable
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One of the first explorations of on-body touch-sensing using stretchable sensors for user interface applications was conducted by Weigel et al. (2015). iSkin is a flexible and stretchable array of touch sensors fabricated by laser-patterning of carbon black and polydimethylsiloxane composite (PDMS). A spacing layer is made by making holes across a PDMS layer. This layer is then sandwiched between two conductive carbon PDMS layers to make touch-sensitive sensor. Possible applications of this device range from a headset control by laminating the stretchable touch sensing skin at the back of the ear, a one-handed slider or touch control by wrapping the sensor skin on the fingers, to a rollout and on-skin keyboard control by attaching the sensor skin on a smartwatch and surface of the forearm.

Another approach, as shown in Figure 18.3a, utilizes a multi-layer of graphene grown by chemical vapor deposition (CVD) and transferred onto a polyethylene (PET) substrate (Kang et al. 2017). The study, using similar approach as Weigel et al. (2015), leveraged the mutual capacitive coupling in between two electrodes to detect a presence of finger touch. As our fingers get close to the sensing layer, they disrupt the electromagnetic field due to the role of our body as a ground conductor and reduce the total capacitance of the sensor. Depending on the sensor design and active area, this method could also detect near-proximity gestures of up to 7 cm (Figure 18.3b). The row-column architecture enables multi-touch gesture capability (Figure 18.3c), and the addition of the third bottom ground layer eliminates the noise caused by the change in surface charge of the skin underneath due to sweat and other physical conditions. DuoSkin presents a similar concept of on-body touch sensing by using a low-cost material, which is gold leaf, to rapidly fabricate temporary tattoo (Kao et al. 2016). Instead of using a mutual capacitive method, a floating or self-capacitive touch sensing mechanism is applied in this work where a finger strike in contrast increases the total capacitance of the sensor. By experimenting with the electrode design, one can design not only

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a touch button, but also a continuous slider and an XY pad. The ease of DuoSkin fabrication and the use of low-cost material allow people to design and personalize the shape and functionality of their own on-skin touch sensor.

Other researchers have also explored the use of biocompatible, hyper-elastic, and transparent materials, such as hydrogels, to develop a mechanically invisible and stretchable ionic sensory skin (Sun et al. 2014). Figure 18.3d illustrates a pressure sensor array that consists of a stretchable dielectric layer sandwiched in between two stretchable ionic conductors. The hydrogel ionic conductors are fabricated by mixing polyacrylamide powder, sodium chloride salts, and cross-linker agents. The pressure exerted on the sensor will reduce the distance between the two conductors and result in the increase of capacitance (Figure 18.3e). In order to create a high resolution touchpad, Kim et al. (2016) developed a stretchable transparent ionic touch panel for a continuous, two-dimensional (2D) gesture by using

FIGURE 18.3 On-skin touchpad (a) image of a finger approaching the 3D conformable sensor on the palm. (b) Capacitance change versus time for finger touch and proximity of various distances between 0 and 7 cm. (c) Demonstration of graphene-based wearable capacitive touch sensor for multi-touch, spread, and scroll user interaction. (Reprinted with permission from Kang, Minpyo. et al., ACS Nano, 11, 7950–7957. Copyright 2017 American Chemical Society.) (d) Ionic pressure sensor laminated on the back of the hand with a pressing interaction. (e) Capacitance change versus time in one location as finger pressure gets stronger and back to initial condition. (Adapted with permission from Sun, Jeong Yun et al., Adv. Mater., 26, 7608–7614, 2014.)
similar materials. The processing method in this ionic touch panel is based on 4-wire sensing, where a signal source is applied to these 4-points. As a finger couples the active area to the ground, it closes the electrical network through the resistance of the active hydrogel. The ratio in between the flowing currents can then be analyzed to find the location of a single touch. The device works reliably on the skin, even though calibration algorithm would improve its accuracy and precision when the device is pre-stretched. They further demonstrated various gestural control applications using the touchpad, including tapping, holding, and dragging to perform certain tasks, such as writing words, performing music, and playing a game.

To extend our perception in interacting with physical and virtual objects in mixed reality, Bermúdez et al. (2018) explored the possibility of touchless manipulation through magnetic field sensing. They fabricated a conformal magnetosensitive electronic skin (e-skin), which consists of 2D giant magnetoresistive (GMR) spin valve sensors in a full-bridge configuration (Figure 18.4a). The full-bridge Wheatstone configuration eliminates the temperature dependence and intrinsic output offset of each meander spin valve stack sensor. The stack is comprised of alternating ferromagnetic and non-ferromagnetic thin-films. The GMR e-skin calculates the angle of a magnetic field input by comparing cosine and sine voltage outputs of the inner and outer bridge, respectively (Figure 18.4b, c). As shown in Figure 18.4d, laminating the GMR e-skin on a palm and interacting with a magnet enables a touchless 2D control of virtual objects. The GMR e-skin, therefore, could continuously monitor the body movement wirelessly with respect to an external magnetic field. It allows various novel applications in navigation and motion tracking for robotics, augmented, and virtual reality.

Textiles are soft and conformable materials, as opposed to conventional electronics built on rigid structures or flexible substrates that can only be deformed along one axis. Supported by recent advances in intelligent textiles, its pervasion in our daily life, particularly in clothing, presents many exciting applications for on-body interactions. Intelligent textiles are fabricated by fusing electronic materials with common textile fibers, threads, yarns, and fabrics. They allow fabrics to not only passively

**FIGURE 18.4** Magnetosensitive e-skin for augmented reality (a) 2D magnetic field sensor based on spin valve sensors in Wheatstone bridge configuration. (b) Magnetic field angle reconstruction based on flexible 2D magnetic field sensor on a flat surface in respect to the orientation of a permanent magnet. The experiment setup is shown in (c). (d) Demonstration of augmented reality-based, virtual object touchless manipulation using the on-skin magnetic field sensor. (Adapted from Bermúdez et al. (2018), Copyright American Association for the Advancement of Science.)
sense, but also to react and adapt its behavior to the user through computation (Post and Orth 1997). One example is FabricKeyboard, a multi-sensory textile-based interface that attempts to enrich physical interaction modalities by embedding a multi-layer of smart and common fabrics (Wicaksono and Paradiso 2017). By using a combination of Ag-plated conductive fabrics and polypyrrole (PPy)-coated piezoresistive fabrics, the textile sensors stack can simultaneously detect proximity, touch, pressure, stretch, and electromagnetic fields. This effort enables new tactile experiences and novel interactions with both physical and non-contact gestures, particularly for deformable interfaces that can conform to any 3D structure.

A functional garment, Levi’s Commuter X Jacket by Google Jacquard has a woven touchpad with gesture sensing capability on its sleeve (Poupyrev et al. 2016). The woven touchpad consists of an XY matrix of conductive threads. Similar to how a touchpad works, projected capacitive sensing is applied to the woven matrix to detect the capacitance in between the threads and calculate XY position of finger gestures. Conductive threads commonly consist of purely metal filaments or a combination of metal filaments and common yarns or fibers. Metal filaments are fabricated by wire-drawing and twisted with the core fibers using a spinning machine. Another approach to making conductive threads is by metal-coating common fibers. Google’s Project Jacquard, in particular, developed special conductive yarns that are customizable and can be easily soldered by twisting insulated copper core with ordinary threads. The project demonstrated a significant effort to large-scale manufacture and commercialize intelligent clothing: from customizing conductive yarns, weaving them into fabrics, to integrating all electronic components. Current research endeavors are directed to an emerging area of fiber electronics (fibertronics), which explores methods to fabricate electronic devices such as sensors, actuators, and transistors directly onto fibers for a high-density and seamless integration between electronics and textiles (Cherenack et al. 2010; Hamedi et al. 2007).

18.2.2 On-body Sensing and Display

Going further beyond sensing, the mechanical properties and the appeal of our skin presents several exciting tactile and visual interaction possibilities. SkinMarks explores the applications of conformal electronics for novel on-skin input and output devices (Weigel et al. 2017). Thin layers of sensing and display electrodes are fabricated by screen-printing poly(3,4-ethylenedioxythiophene)-poly(styrenesulfonate) (PEDOT:PSS). The electroluminescent (EL) display is implemented by printing thin layers of phosphor and dielectric resin paste composite in between the electrodes. The stretchability of PEDOT:PSS prevents structural damages to the devices as our skin deforms. Unique patterns of conductive traces can form input sliders, by leveraging interpolation of neighboring electrodes or squeeze sensors, by the serpentine design of strain gauges. Moreover, the placement of SkinMarks around body landmarks such as fingers could also enable novel input gestures, such as finger bending. Visual feedback via on-skin display further enhances the functionality and interactivity of the SkinMarks.

An exploration of coupling on-skin sensor and display is also demonstrated by Yokota et al. (2016). The ultra-flexible light-emitting diodes (LEDs) are fabricated using polymeric light-emitting materials, sandwiched in-between two thin electrodes. The integration of an ultra-flexible organic pulse oximeter and display allows simultaneous sensing and display of physiological signals in digital format directly from the skin, as illustrated in Figure 18.5a. Further efforts have also been conducted to develop 2D stretchable displays by incorporating serpentine interconnects to bridge inorganic LED-islands on an elastomeric substrate (Hu et al. 2011) or developing novel polymer-based organic LEDs that are intrinsically stretchable (Liang et al. 2013; Sekitani et al. 2009). The thin and deformable nature of these stretchable displays permits their integration onto the surface of the skin for on-body interface applications.

Another muscle-actuated modality besides touch, squeeze, bend, or strain sensing for on-skin interfaces is pressure sensing. The previous examples of on-skin interfaces, however, have limited input-output modalities and resolution. To enable a truly interactive, multi-gesture, and programmable on-skin
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interface, multi-array and multi-layer architecture of sensing and display electronics must be adopted (Wang et al. 2013). Figure 18.5c illustrates an interactive electronic skin that optically responds simultaneously to pressure. In the 2D array (Figure 18.5b), a single device stack comprises a pressure-sensitive rubber, an organic LED, and a carbon nanotube thin-film transistor (CNT-TFT) for matrix-addressing, are all fabricated, on a polyamide substrate. Each pressure-sensitive element is hard-wired to each of the LEDs, resulting in a higher flow of current and brightness as a stronger force is applied to the electronic skin (Figure 18.5d,e). The progress mentioned above shows how innovation in new materials and fabrication strategies can provide opportunities of functionalizing our skin as future sensing and display surfaces
for HMI applications. Further efforts in miniaturization of each sensing, display, and circuit component, as well as a large-scale integration of dense array, are still required for them to compete with current wearable technologies.

### 18.2.3 Tactile Stimulation

With the rapid growth of technologies for immersive sensory experience, there has also been an increasing interest in the development of wearable tactile feedback to provide users with a physical representation of information beyond visual and auditory. Tactile feedback can trigger several somatosensory receptors in the human skin. To them could relay various sensory information including light stroke, touch, vibration, stretch, texture, and pain (Delmas et al. 2011). Having a programmable tactile stimulation platform facilitates a more seamless, realistic, and meaningful interaction in an HMI system. Possible areas that can benefit from tactile feedback include sensory augmentation, motion training, rehabilitation therapy, telepresence, robotics, assistive technology, to virtual and augmented reality interaction.

The rigidity and lack of resolution of current tactile devices limit them from being highly wearable and practical. The ability of these devices to adapt to the curvilinear structure of our skin surface is paramount, as intimate contact improves the effectiveness of sensation. Any tactile sensation can be theoretically recreated with an adequate spatiotemporal resolution, which is around 1.5–3.0 mm and 0–1 kHz, respectively (Yem and Hiroyuki Kajimoto 2017). These challenges call for microfabricated, soft, and conformal tactile stimulators.

Figure 18.6 shows three types of fingertip soft tactile stimulators. The first approach (Figure 18.6a), which is based on electro-tactile stimulation, employs a flowing current to excite mechanoreceptors transcutaneously (Ying et al. 2012). The device consists of an array of gold (Au) circular electrodes on polyimide and is multiplexed by silicon (Si) nanomembrane (NM) diodes for programmable addressing. The entire structure, except active areas, is encapsulated with polyimide (PI) and transfer-printed to an elastomeric substrate (Ecoflex). Experimental results show that the voltage required to achieve sensation declines with an increasing actuation frequency, which agrees with the skin impedance model (Figure 18.6b).

Another approach, as shown in Figure 18.6c, leverages electroactive polymers (EAP) to develop a wearable soft-actuator tactile display (Koo et al. 2008). Commonly used as artificial muscles, capacitive-type EAPs deform with respect to a voltage potential, thus creating mechanical actuation force (Figure 18.6d). The fabrication process starts with spin-coating and curing a dielectric elastomer layer. A mask is then used to pattern electrodes deposited through spray printing of carbon powder solution on both sides of the elastomer layer. A multi-layer structure of electrodes can be applied to reduce the required actuation voltages. The simple fabrication process and low-cost material enable a fast production of large matrix arrays of EAP soft-actuators. The intrinsic mechanical properties of the dielectric and conductive composite layers also facilitate a realization of entirely soft and deformable actuators that can conform to any geometrical shapes. One possible application from this soft-actuator array is a wearable, finger-tip Braille device that can be used for visually impaired individuals.

A soft actuator-sensor skin for vibrotactile feedback has also been demonstrated by Sonar and Paik (2016) (Figure 18.6e,f). The tactile feedback relies on pneumatic actuation, where compressed air is converted into mechanical force. An array of hollow chambers for shape inflation is fabricated by embedding a masking layer in between two silicone layers. At the bottom of each pneumatic actuator, there is a piezoelectric ceramic encapsulated in silicone layers. These piezoelectric sensors allow precise control of actuation amplitude and vibration through a closed-loop system as well as detection of external interaction forces. The actuator is capable of providing 0.3 N force with a frequency of 5–100 Hz. This device is useful not only for wearable tactile feedback in virtual reality systems, but also for rehabilitation devices requiring a haptic feedback.
The advance of multimedia technologies, such as in mobile devices, virtual, and augmented reality, as well as in robotics necessitate a richer set of alternative control modalities that enable a human to intuitively and seamlessly interact with the digital and physical world. Our gesture is a universal body language that enables us to communicate and express our intentions and ideas naturally. The recognition, classification, and interpretation of body gestures, particularly the hand motions have, therefore, been extensively studied and used in the HCI and augmentative and alternative communication (AAC) field (Higginbotham et al. 2007; Sharma and Verma 2015). Continuously monitoring the body gesture and posture is also useful for gait analysis and activity recognition, particularly in prosthetic control, rehabilitation, elderly care, sports, surgical, and industrial applications (Tao et al. 2012).

Most of the current gesture recognition platforms used in consumer electronics or research prototypes rely on vision-sensing (Cheng et al. 2015). Even though vision-based sensors are less cumbersome, non-contact, and allow full-body motion tracking, they commonly require a bulky set-up, color-contrast

![Image of wearable tactile stimulators](image-url)
markers, and a high computational power. Moreover, they could not accurately sense micromotions and have a confined working space. These challenges restrain vision-based sensors from being used for long-term and mobile gesture recognition. Flexible and stretchable electronic devices that can be seamlessly laminated on our body or integrated into textiles for high-quality sensing and maximum comfort could solve these issues. In this section, we will separate these novel devices for human-motion detection into three main categories: inertial, electrical muscle activity, and strain sensing.

18.3.1 Inertial Sensing

Extensive studies have been conducted to perform activity recognition and gait analysis with micro-electro-mechanical systems (MEMS) inertial sensors attached to multiple body segments (Tao et al. 2012). To sufficiently classify body motion in normal daily activities, inertial measurement units (IMUs) must be able to sense accelerations with amplitude from −12 to +12 g and frequency of up to 20 Hz (Bouten et al. 1997). An IMU typically comprises single or multiple accelerometers, a gyroscope, and a magnetometer. It collectively measures 3-dimensional linear acceleration, angular rate, and magnetic field, respectively. Body-worn inertial sensors enable the calculation of angles around the body joints based on their orientations relative to one another. These combined joint angles and other markers from the IMUs can be collected and processed in real-time to characterize gestures. Ultimately, users activity can be recognized or even predicted from this subsequent set of motions (Seel et al. 2014, Yang and Hsu 2010).

One of the initial attempts in developing a flexible format of inertial sensors began by printing silver nano-ink on a paper substrate (Zhang et al. 2013b). A flexible paper accelerometer consists of two conductive layers, in which one of them is a floating membrane supported by two hinges. Due to gravity, there is a force exerted by the membrane mass on the floating top electrode towards or against the bottom electrode as the sensor is accelerated. The distance in between these two electrodes correlates to the capacitance of the accelerometer. As the acceleration towards the bottom electrode gets stronger, the gap between these two electrodes becomes smaller, resulting in a higher capacitance from the equilibrium. The plate size and structure of the hinges can be modified to control the sensitivity and resonant frequency of the paper accelerometer.

An alternative form of a deformable accelerometer is demonstrated by Yamamoto et al. (2016) in their multi-functional on-skin electronics for physiological and motion sensing. The device has reusable and disposable parts; in the disposable part, a printed 3-axis accelerometer is fabricated on a PET substrate with kirigami structure that allows sufficient strain limit against skin deformation (Figure 18.7a–c). The inertial sensor consists of three silver-based strain sensors as hinges and an acrylic plate as a centre membrane (Figure 18.7b). The whole structure is then supported by a silicone rubber. Due to the gravitational force, the centre membrane exerts a force and pulls the hinges. Z-direction acceleration exerts a strain to all of the sensors, while x-direction motion influences sensor #2, and y-direction influences sensor #1 and #3, respectively. The resistance change is affected by the conductive network in the silver-based strain sensors. As illustrated in Figure 18.7d, the acceleration in sensor #1 due to x, y, and z movements begins to increase in a relatively linear fashion from around 5 to 12 m/s², depending on the direction of movement and except for the x axis. Thus, the flexible accelerometer can detect various high-intensity activities when intimately attached to the chest (Figure 18.7e).

Instead of using a floating membrane, free-moving smart materials such as liquid metal can also be utilized to develop deformable inertial sensors (Varga et al. 2017). As presented in Figure 18.7f,g, the device consists of a glycerol chamber encapsulated in silicone, in which an embedded droplet of eutectic gallium indium (eGaIn) could tilt to modulate the capacitance in between two electrodes. A planar eGaIn coil as an extension of the two electrodes is spray-painted on the silicone substrate; the inductor-capacitor (LC) resonant frequency formed by the tilt sensor and the coil correlates to the movements of the eGaIn droplet and can be analyzed wirelessly. Figure 18.7h shows both experimental and theoretical capacitance response of the tilt sensor due to various tilting angles. The tilt sensors, for example, can be worn as a bracelet.
FIGURE 18.7 On-skin accelerometer based on strain sensors bridge. (a) Cross-sectional device structure and (b) digital image of the strain-based on-skin accelerometer. (c) Photograph of the acceleration sensor along with the others attached to a chest. (d) Resistance change and FEM simulation of stress of sensor #1 under a range of acceleration in x, y, and z-axis. (e) Motion sensing demonstration of the on-skin accelerometer during various physical activities. (With kind permission from Springer Science+Business Media: Science Advances, Printed Multifunctional Flexible Device with an Integrated Motion Sensor for Health Care Monitoring, 2, 2016, e1601473–e1601473, Yamamoto, Y. et al.) Wearable and flexible tilt sensor based on eGaIn droplet. (f) Cross-sectional view of device structure with its electrical model. (g) Digital image of the moving droplet inside a chamber with two electrodes. (h) Measured and simulated capacitance response of the tilt sensor over a range of tilt angles $\alpha$ and inclinations $\beta$. (i) Demonstration of output capacitance through different arm gestures as two tilt sensors are mounted on the wrist. (Varga, Matija et al., Lab Chip, 17, 3272–3278, 2017. Copyright 2017 Royal Society of Chemistry.)
to recognize many gestures by processing their distinct output capacitances (Figure 18.10i). Another approach by Persano et al. (2013) attempts to develop a poly(vinylidenefluoride-co-tri-fluoroethylene) or P(VDF-TrFe) nanofiber-based device and explore its piezoelectric behavior in order to measure acceleration and orientation. An electrospinning process is used to fabricate the free-standing, highly aligned, and pressure-sensitive (0.1 Pa) P(VDF-TrFe) fibers. By placing the fiber array as a diaphragm in a chamber inside a box, vibrations and accelerations on the box induce bending on the P(VDF-TrFe) device and generate a voltage. Adding a test mass in the structure allows the device to function as an orientation sensor, as tilting the device would exert lower. Therefore, it generates a small voltage output. Even though the flexible accelerometers discussed here possess lower sensitivity and are relatively larger in size than state-of-the-art MEMS inertial sensors, they demonstrate the future possibility of large-scale manufacturing of soft wearable IMUs for human motion detection.

18.3.2 Electromyography

The second principle of detecting body gestures is to read bio-potential signals generated from the human body. Electromyography (EMG), electrooculography (EOG), and electroencephalography (EEG) are the most commonly used modalities for human-machine interface applications. In this part, we will focus on EMG signals from the body as they are correlated to our physical gestures. EMG detects the change in potentials produced by the muscle cells when they relax and contract. The signal amplitude typically ranges from 100 μv to 90 mv with frequencies from DC to around 10 kHz (Ferreira et al. 2008).

The traditional approach of EMG is to attach wet Ag/AgCl surface electrodes on the skin and record the electric signals generated from muscle activities. However, this approach is cumbersome and uncomfortable, as it requires the application of electrolyte gels, attachment of straps or adhesives, and connection through bulky wires to the electronic systems. The developments of electronic packaging, system design, and signal processing techniques have mitigated the low electrical coupling of skin-electrode interfaces and the influence of motion artifacts, which result in an improved signal-to-noise performance of dry and non-contact electrodes for bio-potential sensing. A commercially-available wearable EMG device such as the Myo armband relies on an array of dry surface electrodes to detect bio-potential signals on the arm and are integrated with on-board IMU for wireless gesture recognition (Torres 2015).

Advances in flexible and stretchable electronics have also further transformed the bio-potential sensing to intimately conform to our skin, especially in curvy and dynamic regions of the human body. An initial example is the epidermal electronic system (EES) proposed by Kim et al. (2011). The EES consists of an assortment of electronic devices, including EMG electrodes integrated with circuitry for signal amplification and processing. The devices, as well as the interconnects, are fabricated in a serpentine structure and encapsulated in PI to withstand skin deformations. The EES is transferred to the skin with a temporary substrate such as polyester or water-soluble polyvinyl alcohol (PVA) and strongly adheres via van der Waals force. By mounting the EES on different parts of the body such as legs, the EES can detect EMG signals of a person walking and standing.

A comprehensive study of the materials, mechanics, and geometric designs of the EES for surface EMG (sEMG) is required for a high signal-to-noise ratio (SNR) output by minimizing contact impedance, motion artifacts, and crosstalk. For this purpose, electrode size and design, inter-electrode distance, and conformity to the skin are the main optimization parameters in the design of on-skin electrophysiological sensors (Figure 18.8a). A study by Jeong et al. (2013) concluded an optimum electrode spacing of 20 mm for minimum crosstalk contamination and larger electrode area for a maximum integrative signal and thus, optimum SNR. The filamentary serpentine (FS) mesh design provides intimate and adaptive contact to the skin, while avoiding interface stress and delamination. The contact resistance of FS electrodes can be reduced by compromising trace width with the curvature radius of the serpentine structure. The mechanics model of skin-surface interaction to estimate the required van der Waals forces for a conformal contact reveals a threshold thickness of 25 μm (Figure 18.8b), with Figure 18.8c showing an excellent contact of
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**FIGURE 18.8** Epidermal electronics for sensing electrical muscle activity through sEMG. (a) Two configurations (bar and disk-type) of sEMG device with measurement, ground, and reference electrodes. (b) SEM image showing a conformal contact between thin silicone layer and the surface of skin. (c) Analytical plot of interfacial contact energy ($U_{interface}$) through various substrate thicknesses. Conformal contact can be deduced for the thickness with positive $U_{interface}$. (d) Comparison of sEMG voltage output between epidermal sensor and conventional electrode with gels. (e–f) Quadrotor control by various arm gestures using epidermal sEMG device. (From Jeong, Jae Woong, et al. Materials and Optimized Designs for Human-Machine Interfaces via Epidermal Electronics. *Adv. Mater.* 2014, 25, 6839–6846. Copyright Wiley-VCH Verlag GmbH & Co. KGaA. Reproduced with permission.) Integrated with sensory feedback from stimulation electrodes for prosthetic control. (g) Multifunctional EES device for simultaneous muscle sensing and stimulation among others. (h–i) Controlling a force on a robotic gripper using sEMG with and without muscle stimulation from the EES. (j–k) sEMG signals on both triceps and biceps while controlling a robotic gripper through the flexion and extension of an arm. (Adapted from Xu et al. (2016), Copyright John Wiley & Sons.)
5 μm silicone membrane with the skin. A membrane thickness over this threshold value will lead to the formation of air gaps, thus low conformability and SNR.

Figure 18.8d compares epidermal and conventional sEMG signal response as the skin is mechanically deformed. The result proves that EES provides a more robust signal for sEMG from noise caused by skin deformations. As illustrated in Figure 18.8e,f, the FS electrodes are then attached to various locations of the body such as the forearm, cheek, forehead, neck, and finger to perform sEMG reading, and both forearms to control movements of a quadrotor via pattern recognition. Further work also explores the possibility of non-contact, capacitive sensing of bio-signals through the electrical coupling between tissues and electrodes (Jeong et al. 2014). In this work, the epidermal electrodes are separated from the skin, with an insulation barrier made out of silicone. Given that there is no direct contact between the metal electrodes and the skin, this technique enables a more durable and long-term reading of bio-signals.

A conformal bioelectronic device that combines muscle electrical activity sensing and stimulation in one platform is also practical in assistive and rehabilitative technology. Xu et al. (2016) presented an EES with closed-loop sensing and stimulation capability for lower back exertion and sensorimotor prosthetics. The skin-like, multi-functional device consists of EMG, strain, and temperature sensors as well as electro-muscle stimulation (EMS) electrodes (Figure 18.8g). The close proximity of EMS and EMG electrodes enables further studies of muscle response to varying stimulation voltages. Sensorimotor control of a robot arm through the EES in the case of a paralyzed limb is also demonstrated. Stable control of gripping force can be achieved by real-time monitoring of biceps and triceps EMG signals and applying a local EMS feedback in concert (Figure 18.8h,i). Another application is to provide a proprioceptive feedback, by laminating two EES around the biceps and triceps brachii muscles (Figure 18.8j). As shown in Figure 18.8k, EMG signals from the EES are classified by linear discriminant analysis (LDA) to give insights to the arm movements. Two stimulation electrodes create a tactile funneling illusion that makes the user to perceive a phantom sensation in a point between the electrodes.

### 18.3.3 Strain Sensing

The third principle of wearable gesture sensing is by measuring strain change across the body using active skin sensors. Stretchable strain sensors can be used to measure the flexion of the fingers, wrist, arms, legs, and other body joints. They can be separated into three main classes: resistive, capacitive, and piezoelectric-based. Resistive or capacitive strain sensors are mainly developed by employing conductive materials such as CNTs, graphene, and metal in nanowires or nanoparticles, as well as liquid-metal alloys. These materials are combined with a stretchable polymeric base or substrate such as Ecoflex or PDMS through various techniques including filtration, deposition, transferring, coating, printing, and solution mixing (Amjadi et al. 2016; J. Park et al. 2015). The selection of materials, structural design, and fabrication methods will influence the strain sensor characteristics, such as stretchability, sensitivity, linearity, response and recovery time, and durability.

Conventional resistive strain sensors, commonly known as strain gauges, rely on a change in geometrical structure and inherent piezoresistive effects. The resistance of a conductor is given by $R = \rho L/A$, where $\rho$ is the electrical resistivity, $L$ is the length, and $A$ is the cross-sectional area. Taking into account the intrinsic change of resistivity in the material due to piezoresistive effects, the relative change of the resistance can be written as $\Delta R/R = (1 + 2\nu)\varepsilon + \Delta\rho/\rho$, where $\nu$ is the material Poisson’s ratio (Mohammed et al. 2008). Semiconducting materials such as silicon and germanium with high piezo-resistivity can be used to fabricate highly sensitive strain gauges. Nevertheless, these devices can typically withstand a strain of up to 1%, unless buckling mechanism or a horseshoe pattern is incorporated into the design (Kim, Xiao et al. 2010; J. Park et al. 2015). As a result, alternative materials and fabrication techniques based on conductive percolation network, micro-crack propagation, or electron tunneling effect have been investigated to develop highly stretchable and sensitive piezoresistive strain sensors, particularly for human motion detection (Amjadi et al. 2016).
The dexterity of our hand motivates a large number of studies to demonstrate the use of strain sensors in monitoring wrist and finger gestures. Yamada et al. (2011), for example, developed CNT-PDMS strain sensors that can be integrated into a glove for the detection of multiple hand gestures (Figure 18.12b, c). The sensors are fabricated by growing thin-films of aligned single walled (SW) CNTs and transferring them to a PDMS substrate (Figure 18.9a). These sensors can detect and withstand a strain of up to 280%, are highly durable when tested in 150% strain for 10,000 cycles. Another example of material that can be transferred or deposited onto an elastomeric substrate for micro-crack propagation strain sensing is gold nanosheet (Lim et al. 2016). Gerratt et al. (2015) used direct deposition of gold thin-films onto elastomeric substrates in order to develop a strain sensor integrated data glove (Figure 18.9d, e). To create the smart glove, strain sensors are connected to the read-out circuit through printed elastic liquid metal (EGaIn) interconnects. The sensor placements on the...
metacarpophalangeal and proximal interphalangeal joints enable accurate measurements of finger flexions (Figure 18.9f). Combined with additional pressure sensors, the smart glove is able to sense the compressibility of an object and guide users to maintain their grasping strength for prostheses tactile feedback.

As shown in Figure 18.9g, instead of integrating strain sensors onto a textile-based material, Muth et al. (2014) explored the applications of 3D-printing of conductive ink (Figure 18.9g) to make a soft glove that detects various finger gestures (Figure 18.9h). The conductive ink consists of silicone oil filled with carbon black. A 3D-printing fluid nozzle, programmed to pattern a strain gauge, prints the conductive ink inside an uncured silicone elastomer and filler fluid layers. Due to the intrinsic conductive network, each strain gauge increases in resistance as it is being stretched (Figure 18.9i). The strain sensor has been tested to reasonably detect a strain of up to 400% and undergo percolation network breakdown at 700%.

A high-resolution spatiotemporal strain sensing by a miniaturized array of strain sensors has also been demonstrated by Kim Jaemin et al. (2014). Because of its high piezo-resistivity and fracture strength, p-type doped single crystalline silicon nanoribbon (SiNR) is employed as the active layer of the strain sensing component. The SiNR array is fabricated on a PI and then encapsulated in a PDMS substrate. A motion-capture system is used to study the strain distribution of multiple hand motions (Figure 18.9j). The results are used to respectively, design the mechanical structure of a strain sensor array based on the maximum local strain (Figure 18.9k, l). As shown in Figure 18.9m, the SiNR serpentine design allows higher deformations with the compromise of lower sensitivity. SiNR6 can withstand a strain of up to 30%, which is suitable for strain sensing on the area close to the wrist, whereas SiNR1 can only handle up to 10% strain. This design strategy permits a robust implementation of high-resolution strain sensing with site-specific sensitivity.

Mattmann et al. (2008) designed a sensor-integrated shirt that can measure the strain distribution across multiple locations of the torso to distinguish different body gestures. The strain sensor is fabricated by mixing thermoplastic elastomer (TPE) and carbon black particles before processing them into a fiber. The sensor’s linearity and sensitivity can be engineered by experimenting the composition of carbon black particles in the mix. As depicted in Figure 18.10a, 21 strain sensor fibers are attached to a shirt and connected to the data acquisition unit by sewing conductive threads. The strain sensor could work with a maximum strain of around 80%. Strain rate testing is conducted by subsequently increasing the strain velocity from 50 to 600 mm/min, and a slight mean error of 5.5% is observed (Figure 18.10b).

![FIGURE 18.10 Sensor-integrated shirt for upper body gesture and posture monitoring. (a) Position of 21 textile resistive strain sensors connected to a data acquisition unit through conductive thread interconnects. (b) Typical resistance change due to cyclic strain to 80% with an observed hysteresis. (c) Demonstration of pattern recognition training study where a participant performs 27 upper body gestures while wearing the sensor-rich shirt. (Reprinted from Mattmann, Corinne et al., Sensors, 8, 3719–3732. Copyright 2008 MDPI.)](image-url)
The strain sensor also retains its functionality under extreme long-term, ageing, and washability tests. With a pattern recognition algorithm and prior user-specific training, the shirt successfully classifies 27 upper-body gestures with 97% accuracy (Figure 18.10c).

For monitoring lower-body gestures, Menguc et al. (2013) demonstrated a soft wearable motion sensing suit for lower-limb biomechanics measurements. The sensors are fabricated by injecting EGaIn liquid metal into a silicone mould. They are then attached to the hip, knee, and ankle parts of the lower body. Experiment results reveal a non-linear resistance change consistent up to 200% and fracture at 364% strain. The wearable motion sensing suit is used to approximate joint angles on the lower-limb, allowing the analysis of a dynamic range of motions and gait of the user for rehabilitation training and sports applications. Using other elastomeric materials with self-healing properties can also significantly improve the elasticity of a strain sensor. A recent effort led by Cai et al. (2017) leveraged a self-healing property of hydrogels to develop a SWCNT/hydrogel strain sensor that can work with up to 1000% strain. The strain sensor is fabricated by dispersing SWCNT and adding PVA solution. The SWCNT/PVA solution is then cross-linked by the addition of an aqueous borax solution and stirred until a hydrogel is formed. This ultra-stretchable strain sensor retains stable electrical and mechanical performance under multiple stretching cycles. Applying the sensors to different parts of the body such as finger, knee, neck, and arm allows them to monitor a diverse set of human motions continuously.

A capacitive strain sensor employs conductive and stretchable materials as active layer and substrate. Its structure consists of a stretchable dielectric layer in between two stretchable conductive electrodes, resembling the parallel plate architecture of a capacitor. The capacitance can then be written as \( C = \varepsilon_0 \varepsilon_r A/d \), where \( \varepsilon_0 \) is the electric constant, \( \varepsilon_r \) is the dielectric constant of the material between the plates, A is the overlapping area of the two plates, and d is the separation distance between the plates. Based on this underlying equation, tension on the strain sensor changes the effective area of the two electrodes and reduces the dielectric thickness in between, with respect to Poisson’s ratio. This will increase the sensor’s capacitance and vice versa. In comparison to resistive strain sensors, capacitive strain sensors exhibit better linearity, stretchability, recovery, and hysteresis performance, but with a trade-off of low sensitivity (J. Park et al. 2015; Amjadi et al. 2016). Thus, developing a strain sensor with an excellent performance on all of these parameters is still a challenge.

Some studies have been conducted to develop stretchable capacitive strain sensors for human motion detection. One approach is by depositing conductive films on an elastomeric substrate (L. Cai et al. 2013). Thin carbon nanotube films are initially grown using a CVD process. They are then transferred to the front and back surface of a stretchable PDMS base through PET frames. Because the CNT film is thin, the strain sensor exhibits a high optical transparency of 80%. It can also detect strain as high as 300% with an excellent durability. As an alternative approach, Atalay et al. (2017) investigated metal deposition and laser rastering to develop a capacitive strain sensor. Before depositing metal layers (Al and Ag) on a silicone substrate, surface modification techniques are executed. The surface modification through rapid laser rastering forms microgrooves. Pre-straining of the silicone substrate avoids cracks on the metal electrodes and ensures high repeatability of the sensor. Figure 18.11a shows the working mechanism of the laser-treated capacitive strain sensor and its cross-sectional view. The strain sensor remains linear and repeatable up to 85% strain and can withstand a maximum strain of up to 250%. The sensor successfully detects arm movements when attached to the elbow joint (Figure 18.11b, c).

The aforementioned capacitive strain sensors are customized to only allow single axis detection. To finely detect strain distributions on a large surface of the skin, an array of strain sensors must be developed. Zhao et al. (2015) reported a stretchable multi-functional electronic skin capable of mapping static and dynamic strain for HMI application (Figure 18.11d). This stretchable sensor consists of a silicone (Ecoflex) dielectric in between two layers of 7 by 7 islands or arrays of silver metal electrodes on a PET substrate and encapsulated in a PDMS layer. As a specific area of the sensor is touched, bent, stretched, or pressed, the strain distribution across this electronic skin can be analyzed through the change of
capacitance in individual islands. Bending tests on the sensor reveal a linear response of relative capacitance in respect to various bending angles (Figure 18.11e). Figure 18.11f demonstrates spatiotemporal strain mapping around the elbow joint with respect to multiple flexion angles.

Piezoelectric nanogenerators (PENGs) convert mechanical energy into electrical charges and vice versa. Since they can produce a relatively large output with a minimal strain, piezoelectric materials
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exhibit a high sensitivity in response to strain. They also generate electrical energy, making them consume little or no power and requiring a relatively simple read-out circuitry to digitize the sensor signals. The fact that PENGs and other nanogenerators intrinsically generate power also opens up the possibility of a self-powered system, where the output from the sensor is used not only to detect motion, but also to output power to the read-out and communication interface circuit simultaneously. However, PENGs also have their limitations; it can only be used to measure changing or dynamic strain instead of static change, limiting its application for continuous strain sensing (Kon et al. 2007). As a consequence, piezoelectric-based sensors can only be used in certain applications that do not require static load sensing.

An example of a piezoelectric-based gesture recognition device is a ZnO conformal homojunction nanowire film that can be laminated directly to the skin, as shown in Figure 18.12a (Pradel et al. 2014). The nanogenerator device fabrication starts with sputtered Al-doped ZnO (AZO) on a PET substrate as the bottom electrode. The hydrothermally grown, undoped n-type ZnO and Sb-doped p-type ZnO are then spin-coated with polymethylmethacrylate (PMMA) for current leakage intervention and finally sputtered with another AZO layer as the top electrode. Experimental results prove that the bilayer, homojunction structure of pn-ZnO gives the best piezoelectric performance compared to single-layer or other homojunction pairs. The piezoelectric sensor is then encapsulated in silicone and mounted on a wrist to detect multi-finger gestures. By monitoring the PENGs voltage output due to the movement of flexor tendon, this device can distinguish multiple complex finger gestures as demonstrated in Figure 18.12b. Another work by Lim et al. (2015) utilized transparent and ultrathin nanomaterials to develop an invisible and interactive human-machine interface based on a piezoelectric motion sensor and electrotactile stimulator or EMS. The piezoelectric sensor consists of a polylactic acid (PLA)/SWNT composite film as the active element and is sandwiched in between transparent graphene/PMMA films. The EMS electrodes are composed of a graphene (GP)/silver nanowire (AgNW)/GP layers on top of a PDMS substrate (Figure 18.12d). Material selection, thin structure, and serpentine design allow the device to be transparent and stretchable. The addition of an electro-tactile stimulator enables the demonstration of interactive HMI systems such as in a prosthetic arm, where the sensor and stimulator are linked to control an external robot arm in a closed-loop sensory feedback mechanism (Figure 18.12e). As shown in Figure 18.12f, bending and pressure tests confirm the electrical property of the sensor in detecting dynamic motions of the wrist and providing real-time tactile stimulation.

Another type of nanogenerator that can be used for human motion sensing is triboelectric nanogenerator (TENG). TENG produces electricity from mechanical energy through a combination of electrostatic induction and contact electrification due to frictions between different materials. TENGs are particularly attractive, as they can be fabricated with a wide option of materials, are relatively low-cost, light-weight, and easily scalable (Wang 2014). Lai et al. (2017), as shown in Figure 18.12g, developed a TENG single-thread that can be sewn into textiles. The device consists of a single stainless-steel conductive thread coated with silicone rubber. Another thread configuration, such as the helix-belt structure of an inner electrode with an outer electrode can be explored to maximize the output energy of thread-based TENGs (Wang et al. 2016).

The silicone rubber is used to attract electrons since it has a high electron affinity. Sewing the thread in a serpentine pattern on an elastic textile allows stretchability of the cloth-based TENG up to 100%. The entire cloth can be folded, twisted, and crumpled without any degradation in structural mechanics and performance of the TENG. The active, self-powered sensing device operates by the triboelectrification and electrostatic induction during its regular contact with the skin. As the TENG thread is compressed and separated to and from the skin, electrons travel from the skin to the silicone rubber surface and vice versa (Figure 18.12h). The recurrent induction of charges results in a successive flow of positive and negative currents. By sewing the TENG-thread on a textile glove, one can demonstrate an application of self-powered gesture sensing (Figure 18.12i) to detect specific finger actions based on its unique signal patterns, as well as self-powered wearable touch sensing (Figure 18.12j).
FIGURE 18.12 Conformal and transparent piezoelectric device on a wrist to detect finger gestures. (a) Digital image of the conformal ZnO nanowire device laminated and deformed on a wrist. The device is also highly transparent. A pn junction characteristic can be observed through current-voltage (IV) measurements. (b–c) Different wrist movements and finger gestures result in unique voltage outputs from the device. The unique patterns are consistent, and therefore, can be used as an input for hand gesture-based human-machine interfaces. (From Pradel, Ken C. et al., Nano Lett., 14, 6897–6905, 2014, Copyright 2014 American Chemical Society.) Thin and transparent piezoelectric sensor integrated with electrotactile stimulator for interactive HMI. (d) Exploded view of patterned GP heterostructures with PLA/SWNT as a sensing (left) and AgNW as a stimulation (right) layer. (e) The devices are then attached to the wrist for robotic arm control with closed-loop sensory feedback. (f) Relaxing, bending, and pressing the piezoelectric sensing element actuates the robotic arm to bend, grasp with stimulation feedback, and lift correspondingly. (Adapted from Lim et al. (2015), Copyright John Wiley & Sons.) Thread-based triboelectric nanogenerator for self-powered interactive HMI. (g) Fabrication process of single-thread triboelectric nanogenerator and its integration into textile. (h) Electricity generation principle of the triboelectric thread. Demonstration of triboelectric-integrated glove for (i) self-powered finger gesture sensing and (j) interactive touchpad showing signals from tapping Morse code patterns. (Adapted from Lai et al. (2017), Copyright John Wiley & Sons.)
18.4 Speech and Voice Recognition

Human vocal folds are integral media for human interaction through communication. They oscillate to generate audible waves, enabling us to express ourselves and interact with one another effectively (Traunmüller and Eriksson 1994). With the vast advances in hardware systems, big data, and machine learning algorithms, it is now possible for computers to efficiently process audio signals for applications in voice recognition, particularly for speaker identification, authentication, and speech recognition. Typically, we interface with these audio technologies through a microphone. However, the distance, indirect contact, and environmental noise sometimes misinterpret our commands and force us to speak louder than usual. Additionally, these technologies are also ineffective for people with speech disorders. In this respect, we will discuss the role of flexible and stretchable devices that can be intimately attached to sense vibrations or muscle movements and recognize the sound generated by the vocal cords for the next generation of personal virtual assistants. These devices can be as a media for people with speech disorders to communicate or as novel voice-based HCI devices such as for silent speech interfaces. Several techniques can be used to detect human voice and speech with on-skin electronics: sensing the electrical activity around the thyroarytenoid muscle or monitoring the physical change of skin surface around the neck with vibration, strain, and pressure sensors.

18.4.1 Electromyography and Vibration Sensing

A wireless, on-skin stethoscope by Liu et al. (2016) offers both EMG and vibration sensing for listening to the internal sounds of the human body. As shown in Figure 18.13a, the mechano-acoustic system consists of a collection of electronic devices with their read-out chips embedded in an elastomeric shell. The elastomeric shell, for the structural protection of the electronic chips, is made out of outer and inner silicone that can withstand a biaxial strain of up to 25%. The stretchable system is laminated on the neck (Figure 18.13b) to allow simultaneous sensing of EMG signals through serpentine conductive electrodes and acoustic vibrations through the accelerometers for sensor fusion (Figure 18.13c). As shown in Figure 18.13d, compared to a standard microphone system, the on-skin stethoscope performs better in capturing sound signals in noisy conditions due to the intimate contact between the sensor and the skin. Hence, this device is particularly useful for first responders, security agents, or ground controllers to communicate in noisy environments. They demonstrated an HMI application of the system for speech recognition, which is to control a game by detecting in real-time particular words (left, right, up, down) based on the sensor outputs. With preprocessing of signals to reduce noise and classification techniques based on linear discriminant analysis, a recognition accuracy of 90% can be achieved (Figure 18.13e). This speech recognition system can also be used in many other applications, such as in unmanned aerial vehicles and robotics remote control.

EMG signals generated from swallowing by submental muscles have also been utilized as an HMI input (Lee et al. 2017). Thin gold electrodes, resting on a PI and elastomer substrate, are designed in an island-bridge configuration with serpentine bridge structures. The device can withstand a biaxial strain of up to 150%, which is sufficient to accommodate maximum deformation of the human skin of up to 70% (Pawlaczyk et al. 2013). The swallowing detection is done by initially employing a 30–150 Hz bandpass filter to the raw EMG signals from swallowing, then smoothing the filtered signals by taking root-mean-square values of 250 data points. Finally, the processed signals are compared to a pre-calibrated baseline and threshold value for classification. A biofeedback game, customized for dysphagia rehabilitation exercises is also demonstrated. The biofeedback game takes binary data from the EMG sensor to trigger a ball that jumps through obstacles. Evaluation of the results concludes that the skin-like electrodes have a comparable and better performance than rigid electrodes, with a false positive rate of 3%, which is 2% lower than using rigid electrodes.
Inspired by the architecture of human eardrum, a flexible bionic membrane sensor (BMS) that measures dynamic pressure of cardiovascular and vocal cords activities was developed (Yang et al. 2015). The BMS, as shown in Figure 18.14a, leverages TENG effects from polytetrafluoroethylene (PTFE) nanowires to sense a high range of vibration frequency from 0.1 to 3.2 kHz with a pressure detection limit of 2.5 Pa and sensitivity of 51 mV Pa\(^{-1}\). By attaching the BMS on the throat, the system can simultaneously and sensitively detect the low-frequency vibration components from the arterial pulse and the high-frequency components from muscle movements stimulated by the vocal cords; applying a band-pass filter (BPF) in the range of 45–1.5 kHz reveals the vocal signals without the low-frequency arterial pulse component (Figure 18.14b,c). Given this capability, a novel multi-modal biometric authentication can be realized by correlating both arterial pulse and throat sound to find a unique signal pattern among users.

Dagdeviren, Su, et al. (2014) presented a PENG-based conformal sensor with transistor amplification for cutaneous pressure monitoring. An audio speaker system is used to observe the piezoelectric sensor response to audible tones under high frequencies and low pressures. The flexible device can detect
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Pressure as low as \( \sim 0.005 \text{ Pa} \) with a sensitivity of \( \sim 1 \mu \text{APa}^{-1} \) thanks to the 2D array configuration of sensing elements and SiNM n-channel metal-oxide semiconductor field effect transistor (n-MOSFET). The IV response of the SiNM n-MOSFET enables amplification of the piezoelectric voltage output from the array to the gate of the n-MOSFET, converted in the form of drain-source current. The device can sensitively detect pressure variations of blood flow in the near-surface arteries when attached around the wrist, arm, and throat.

Figure 18.14f presents a bi-directional, flexible artificial throat based on laser-induced graphene (LIG) (Tao et al. 2017). Instead of using the principle of piezoelectricity or triboelectricity, the LIG harnesses the thermo-acoustic effect to both detect and generate sound in the audible range. A direct laser writing for rapid and low-cost prototyping is used to transform a PI substrate into a graphene layer (Figure 18.14d). A periodic joule heating, induced by the applied alternating current (AC) signals, results in an expansion of air and thus, generation of sound. On the other hand, vibrations on the LIG surface change its electrical behavior. The LIG could sensitively detect human-generated sound, including a cough, hum, scream, swallow, and nod when placed on the throat. Simultaneous sound...
sensing and generation are also demonstrated by training specific sound inputs via a machine learning model and correlating it with a particular, higher-volume sound output (Figure 18.14e–g). The LIG artificial throat, therefore, enables people particularly with speech disorders to express themselves through this bidirectional technology.

In a related application, Wang et al. (2014) explored the possibility of monitoring muscle movements during speech by developing an ultra-sensitive pressure sensor. Silk-based textiles are used as a mould to create a PDMS layer with a microstructure pattern on its surface. Free-standing, thin-films of SWCNT are then transferred to this PDMS layer and annealed. Two layers of the micro-structured PDMS are then sandwiched to serve as a resistive pressure sensor (Figure 18.15a). The density of

![FIGURE 18.15](image)

**FIGURE 18.15** Flexible sound recognition devices based on pressure sensor. (a) Device structure of a conductive micro-structured PDMS pressure sensor. (b) Comparison of sensitivities between high-density polydimethylsiloxane (H-PDMS) and low-density polydimethylsiloxane (L-PDMS) sensor patterns. (c) A user wearing the pressure sensor on his neck to recognize muscle movements during speech activity. (d) IV curves of the H-PDMS pressure sensor as the user speaks different words. (From Wang, Xuewen et al. Silk-Molded Flexible, Ultrasensitive, and Highly Stable Electronic Skin for Monitoring Human Physiological Signals. *Adv. Mater.* 2014. 26. 1336–1342. Copyright Wiley-VCH Verlag GmbH & Co. KGaA. Reproduced with permission.) Based on strain sensor. (e) Device structure of graphene-woven fabric on a PDMS substrate. (f) Resistance response of graphene strain sensor during vocalization and non-vocalization as the user performs a similar speech action. (g) The graphene strain sensor attached to a loudspeaker’s membrane and the neck. (h) Resistance change of the strain sensor attached to the throat (red) and the loudspeaker (black) to the same speech action with the recording data (blue). (With kind permission from Springer Science+Business Media: *Nano Res.*, Ultra-Sensitive Graphene Strain Sensor for Sound Signal Acquisition and Recognition, 8, 2015, 1627–1636, Wang, Yan et al.)
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micro-structure patterns ultimately gives a boost of sensitivity to the device (Figure 18.15b). Under 300 Pa, the sensitivity of the high-performance high-density PDMS sensor is as high as 1.8 kPa⁻¹. As demonstrated in Figure 18.15c,d, attaching this pressure-sensitive skin on the neck reveals unique resistance patterns when the user speaks a different set of words and allows the application of speech recognition.

18.4.3 Strain Sensing

The last approach of an on-skin device for wearable speech recognition is based on strain-sensing (Wang et al. 2015). The device incorporates a graphene woven fabric (GWF), embedded in a PDMS, and acting as a microstructured conducting film. The GWF is developed by a CVD process on a seed copper mesh. The resistance of the GWF-based sensor changes when the user speaks, which corresponds to the muscle motions around the neck and strain across the sensor. The signal peaks of the GWF-based sensor when laminated on a throat and a vibrating membrane of a loudspeaker confirm with the original audio waveform, as shown in Figure 18.15h. In addition, testing data show a similar change in resistance as a user performs the same speech action with and without vocalization, revealing the usability of the sensor to speech-impaired individuals or for a silent-speech interface (Figure 18.15f).

18.5 Facial Gesture Recognition

Muscular movements of the face are triggered verbally by our speech and non-verbally by our facial expression and gestures. A facial gesture conveys valuable information of our emotional states and is a key parameter in the study of social interactions. Monitoring facial gestures is, therefore, essential in human-machine interfaces. It enables machines to understand human behaviors and to proactively respond by providing contextual and emotionally aware decisions. It has also extended its applications to AAC and medical rehabilitation.

The most commonly used technique to detect a facial gesture is vision-based sensing. This method requires high complexity processing, is not robust to environmental noise, and is relatively bulky. Researchers have also explored other modalities such as EMG, EEG, EOG, capacitive, and electromagnetic sensing for facial gesture recognition (Matthies et al. 2017). However, current research mostly used off-the-shelf components or commercial gel electrodes, restraining the system from giving high-quality signals while also being wearable and comfortable. As a solution, flexible and stretchable devices that can be easily worn, seamless, and conform to the skin should be developed.

18.5.1 Electromyography and Electrooculography

Paul et al. (2014) proposed a reusable fabric patch, which consists of EMG and EOG electrode arrays for facial electrical muscle activity monitoring. Twenty silver polymer electrodes are fabricated by using a screen-printing technique. A textile-based substrate, treated with an interfacial layer is used in this work. The interfacial layer ensures a smooth surface on the textile and prevents discontinuities when printing the conductive materials. The fabric patch, in the form of a headband, is then tested on a human subject by corresponding facial movements, such as raising eyebrows (frontalis muscle), clenching jaw (temporalis muscle), and horizontal EOG (lateral rectus muscle) to a cursor or controller. Testing procedures showed a significant change of voltage amplitude in four of the electrodes around the top of user’s forehead.

Amplification and filtering system is constructed to process the data from these four electrodes to three output channels. 25–125 Hz filter is used to process EMG signal from the jaw and eyebrow while 10 Hz filter is used to process EOG signal from the eyes. With feature extractions and baseline
calibrations, the user interface system can classify eyebrows and jaw movements to control ‘up’ and ‘down’ commands and eyeball movements to control ‘right’ and ‘left’ commands. This work is particularly useful for AAC, since the facial gesture recognition system can support speech and writing for those with dysarthria or other severe neuromotor disorders, such as cerebral palsy, amyotrophic lateral sclerosis, traumatic brain injury, and Parkinson’s disease.

### 18.5.2 Strain Sensing

On-skin electronics that are attachable to the human body especially around the face or the neck should also be designed to look seamless and natural in order to gain widespread user acceptance. As a result, optical transparency or invisibility is a critical feature in the development of wearable electronics on the facial skin. As shown in Figure 18.16a,b, a transparent and stretchable piezoresistive strain sensor has been developed to monitor facial gestures for emotion monitoring during daily activities.

![Resistive strain sensors for facial gesture monitoring](image)

**FIGURE 18.16** Resistive strain sensors for facial gesture monitoring. (a) Placement of transparent, stretchable, and ultra-sensitive around the forehead, near the mouth, under the eye, and on the neck to measure skin micro-motion induced by movements of the facial muscle movements during emotional response and daily activities. (b) A cross-sectional structure of the strain sensor comprising a three-layer stack of PU-PEDOT:PSS/SWCNT/PU-PEDOT:PSS on a PDMS substrate. The relative resistance change of the strain sensor when attached to the (a,e) forehead and (b,f) near the mouth as the subject was laughing and crying, respectively. (Roh, Eun et al., ACS Nano, 9, 6252–6261, 2015. Copyright 2015 American Chemical Society.) Facial expression recognition. (g) Using the AgNP curved array strain sensors previously shown in Figure 18.16g and placing six of them around the facial skin basic muscle groups allow feature extraction for classification. (h) PCA and (i) HCA results show a significant separation between each cluster when the strain sensors are placed at the corner of the lips, canthus, and in between the eyebrows. (From Su, Meng et al., Nanoparticle Based Curve Arrays for Multirecognition Flexible Electronics. Adv. Mater., 2016. 28. 1369–1374. Copyright Wiley-VCH Verlag GmbH & Co. KGaA. Reproduced with permission.)
Flexible and Stretchable Devices for Human-Machine Interfaces (Roh et al. 2015). The strain sensor has a three-layer stack structure consisting of SWCNTs sandwiched in between two polyurethane (PU)-PEDOT:PSS layers on a PDMS substrate. The PU-PEDOT:PSS and SWCNT solutions are deposited and developed onto the PDMS substrate by spin-coating, followed by thermal annealing and a chemical functionalization process. The sensor gauge factor and transmittance are dependent on the SWCNT concentrations. Mechanical testing confirmed the sensor can operate with a strain of up to 100%. Placing the sensor around various regions of the face reveal a maximum of 0.6% and 40% strain on the forehead and skin near the mouth, respectively. Figure 18.16c–f shows the sensor response as the subject was laughing and crying.

Another research of on-skin devices for facial skin micro-motion detection was conducted by Su et al. (2016). The sensing device is designed by the self-assembly of Ag nanoparticles (NPs)-containing liquid in a curve-patterned micropillar (Figure 18.18g). The AgNP array is then printed on a PDMS film and sintered. Cr/Au interdigitated electrodes are then deposited on the PDMS film and provide the base connections to the AgNP curves array. The strain sensor gives consistent results when stretched to 5% its length in 1000 cycles. Six of these sensors are then mounted on basis muscle groups around the face to recognize eight facial expressions, which are relaxed or default, angry, disgusted, scared, laugh, sad, smile, and surprised (Figure 18.16g). Jack-knifed classification procedures were performed on each of the six sensors and gave insights and validations on the top three sensor locations, which are on the corner of the lip, canthus, and in between the brows. As illustrated in Figure 18.16h i, principal component analysis (PCA) and hierarchical cluster analysis (HCA) results show a clear separation between each cluster of the facial expressions. It can be concluded that most of the negative emotions correlate with the movements of the brows, while the contraction of muscles around the canthus and lips exemplify delightful and strong expressions, such as smiling, laughter, fear, and surprise.

18.6 Eye-motion and Gaze Detection

Eye motions and gaze contain a natural, high-bandwidth source of voluntary and involuntary inputs for human-machine interfaces. They are one of the main markers that correspond to our attention. Eye tracking devices that measure eye movements and positions have been used in a broad range of applications. These include visual behavior monitoring in psychology (Gidlöf et al. 2013), oculomotor rehabilitation (Sharma 2011), activity recognition (Bulling et al. 2011), gaze-based human-computer interaction (Bulling and Gellersen 2010), sleep monitoring (Keenan and Hirshkowitz 2010), and augmentative and alternative communication (Al-Rahayfeh and Faezipour 2013).

There are three well-known technologies for eye tracking: optical, electric potential, and eye-attached tracking. Most efforts on eye tracking are predominantly vision-based, with a typical spatial resolution of 0.5–1 (Whitmire et al. 2016). Some of these vision-based systems used optical instruments that can be relatively worn as a light head-mounted gear. High-resolution optical instruments are currently expensive and computationally demanding. They also require a complex processing setup. Moreover, movements of the user and varying environmental conditions could heavily add noise and reduce the accuracy of the tracking. The recent advances in mobile systems, particularly for the emerging augmented reality (AR) and virtual reality (VR) technologies, hence, demand a novel solution for highly accurate, low-power, and wearable eye tracking.

18.6.1 Electrooculography

EOG measures electrical potentials caused by the movements of cornea in between multiple electrodes around the eye. Compared to the optical tracking, this method works under various lighting conditions, requires a low computational power, and can be worn non-obtrusively (Bulling et al. 2011). EOG systems can even be implemented to record eye motions with the eyes closed, which is useful during sleep.
study for the detection of rapid eye movements. Although EOG can be effectively used to monitor saccadic eye movement and blinks, due to its low SNR and potential drifts, it is challenging to use it for detecting gaze direction and monitoring slow eye motions with high accuracy.

Mishra et al. (2017) demonstrated soft, bio-signal electrodes to perform wireless EOG that can be attached intimately to the skin (Figure 18.17a). The electrode fabrication is similar to the EES.
previously mentioned, with a fractal structure of electrodes transferred to an elastomeric substrate (Kim et al. 2011). They compared the performance between fractal and conventional electrodes in measuring EOG potentials, as shown in Figure 18.17b. Based on the filtered and processed derivative peaks, five features are defined: amplitude, velocity, mean, wavelet energy, and definite integral. A machine learning algorithm is then used to classify the EOG signals in real-time. In the end, they showed an application of human-wheelchair interface for individuals with a disability. The soft bio-electronic sensor with the integrated system can detect four eyeball directions (up, down, right, and left) with 94% of accuracy in order to control the movements of a wheelchair.

18.6.2 Scleral Search Coils

A more high-resolution method to detect eye motions and gaze direction that is comparable to video recognition is magnetic field tracking with scleral search coils (van der Geest and Frens 2002) (Figure 18.17c,d). This semi-invasive technique provides a high temporal (>1 kHz) tracking with a spatial resolution of under 0.1° and is still functional with the eyes closed (Sprenger et al. 2008; Whitmire et al. 2016). The setup consists of torsion coil(s) embedded in a silicone rubber as a contact lens. The lens adheres to the sclera of the eye around the iris. Multiple Helmholtz coils that uniformly produce a magnetic field are placed around the head. This magnetic field induces a voltage in the torsion coil(s) according to their 2-/3D orientations. The voltage change given by these scleral coils in respect to the transmitters, therefore, corresponds to the eye motions (Robinson 1963). Since the torsion coil is semi-invasive and uncomfortable to the eye, this approach currently cannot be worn for long-term applications and is mostly used in medical settings.

18.6.3 Pressure and Strain Sensing

Several investigations of on-skin eye tracking involved novel materials and device designs. Lee et al. (2014), for example, presented a flexible, light-weight, ultra-thin PENG for detecting small skin deformations. The PENG is developed by depositing a ZnO seed layer onto Al foil with Al₂O₃ and then growing ZnO NW through a hydrothermal process. The high sensitivity of this thin PENG to bending enables the application of eyeball motion tracking when the device is laminated around the eyelid. The output voltage and current results generated by the piezoelectric device during the strain influenced by the eye movements showed the functionality of the PENG to detect slow to rapid (0.4–1.6 Hz) eye movements. The proposed system can be utilized to monitor emotion, alertness, and sleeping pattern. A TENG patch can also be attached near the canthus to sensitively detect mechanical micromotion of the skin due to the eye motions and blinks (Pu et al. 2017). This flexible and transparent TENG patch consists of fluorinated ethylene propylene (FEP) coated with indium tin oxide and laminated to a PET substrate. A cavity wall formed by PET then separates a natural latex layer with the FEP. As the users’ eye is closed, the muscle contraction pushes the latex towards the FEP and electrons flow from the external circuit to the indium tin oxide (ITO) and vice versa. The voltage generation by the patch due to blinking (~750 mV) surpasses the typical voltage generated by EOG (1 mV). This performance reduces the possibility of false-positive detections due to low signal level and high noise. The interface system could also be self-powered and does not require complex circuitry and signal processing.

Another approach explores the use of piezoresistive strain sensors as previously discussed in detail in the previous section of this chapter (Roh et al. 2015; Su et al. 2016). Attaching these micro-motion strain sensors around the upper eyelid, lower eyelid, and canthus of the eye allows tracking of four eyeball directions through the resistance change of the sensors (Figure 18.18h). Further work of incorporating more nodes of micro-fabricated strain sensor with high-level signal processing could improve the eye-tracking resolution and enable a higher accuracy, point-of-gaze detection.
Emotions play a vital role in our daily life as they enable us to express and understand each other’s feelings. They are represented by external physical expressions and internal mental processes that may be imperceptible to us. The ability to recognize human emotions and simulate empathy has become an important aspect in human-machine interaction systems, prompting the field of affective computing or artificial emotional intelligence (emotion AI) (Picard 1997). Recognizing emotions enables machines to adapt and
react depending on the user’s behaviors, allowing a more natural and efficacious mutual relationship between human and computers. Multiple methods have been explored in the past years to monitor and classify human emotions. The most widely used approach involves the detection of facial expressions, speech, body gestures, and physiological signals (Castellano et al. 2008). Except for physiological monitoring, which uses wearable sensors, current approaches of emotion recognition mainly use an external camera in order to recognize facial gestures or microphone to process voice signals. As we have covered recent developments of flexible and stretchable devices for body gesture, speech, and facial expression recognition (Sections 3 through 5), in this section, we will mainly discuss the development of these devices for physiological sensing. The fact that individuals cannot easily control their physiological signals makes sensing them extremely useful, as manipulating these signals to hide our emotions is challenging.

18.7.1 Electrodermal Activity

Several physiological parameters can give insights into human emotions. These include skin conductance, temperature, respiration rate, cardiac function, and electrical activity of the muscle, heart, and brain (Wioleta 2013). Electro-dermal activity (EDA) relates to the conductance of the skin, commonly known as galvanic skin response (GSR) that can be influenced by psychological arousal (Boucsein 2012). This skin conductance change is primarily influenced by sweat from stimulation of sweat glands through the autonomic nervous system. The water and electrolytes in the sweat provide conductive pathways through the skin, thus increasing the overall conductance. The EDA can be measured by measuring the change in impedance or potential in between two electrodes or using a combination of them. Several researchers designed EDA monitoring systems that are wearable and conformable. This prevents the use of wet or gel electrodes, which are uncomfortable and not reusable, or dry electrodes, which have low performance and are not adaptive to the curvature of the skin.

Figure 18.19a shows a flexible GSR sensor with conductive polymer foam, designed by Kim, Jeehon et al. (2014). The use of the soft conductive foam for the electrode enhances the sensor’s comfort, reusability, and contact to the skin upon pressure. To test the system, they strapped the wearable GSR sensing system on the back of a body and compared it to a conventional GSR system. The results showed a high correlation between the two GSR signals when three auditory stimuli, which are screaming, glass breaking, and gun blasting were triggered without the subject’s knowledge (Figure 18.19b). Further tests to 19 participants revealed a 94.7% success rate of GSR reading.

18.7.2 Multi-modal Physiological Sensing

The use of more than one modality of physiological parameters has been proven to improve classification number and accuracy of emotion recognition (Lisetti and Nasoz 2004). Yoon et al. (2016) proposed a multi-modal flexible human stress monitoring patch that incorporates not only GSR, but also skin temperature and arterial pulse-wave sensors in a multi-layer structure (Figure 18.19c). Two silver electrodes (A and D) serve as the GSR sensor while a meander-line structure of interconnect (B and C) acts as the skin temperature sensor. A Parylene C layer separates the skin contact layer with the pulse wave sensing layer. The arterial pulse is sensed by a thin flexible piezoelectric membrane in between two silver electrodes (E and F). Figure 18.19d–f shows the experimental and theoretical response of each sensor to its corresponding stimulus. They concluded that with a machine learning algorithm such as support vector machine (SVM), the physiological parameters given by this patch could be potentially used to classify four types of human emotion, such as surprised, angry, stressed, and sad.

A study to observe the electrophysiological response given by an individual in an unforeseen situation was conducted by Jang et al. (2014). For this purpose, a stretchable electronic system that includes electrophysiological (EP) sensing electrodes and an optical blood oximeter is fabricated (Figure 18.20a). Filamentary serpentine mesh geometry serves as the three EP electrodes for EOG, EMG, or ECG. A micro-scale inorganic light-emitting diode (µILED) composed of AlInGaP is used as the blood
The oximetry actuation element. The whole system including the interconnects is supported by a PI layer and transferred to a breathable, silicone-coated textile substrate. Based on the intensity of the scattered lights, they showed a comparison between a human subject performing meditation and a stressful task and concluded an inversely proportional relationship between the intensity of task and the light absorption due to the blood flow rate (Figure 18.20b,c). Further tests in a virtual driving simulator also revealed an abrupt change in EP signals (Figure 18.20d) around the eye, chest, and arm during sudden braking and merging events, as shown in the spectrograms of Figure 18.20e. There is a sudden increase of eye activity level through the EOG signals after the event. EMG signals collected from the EP sensor on the arm showed a brake reflex of the subject, as the subject reacted to avoid the collision by turning the wheel. Regional activations in two different frequencies can be observed on the EP sensor attached to the chest. These activations correlate to the surprised and nervous state of the subject, represented by an increase of ECG and EMG response evoked by the unexpected event.

The physiological parameters discussed in this chapter can also be used not only to monitor stress and cognitive states, but also detect pattern abnormalities that may lead to mental disorders, such as schizophrenia and depression. Given the fact that flexible and stretchable microfabricated sensors allow long-term, comfortable, high-sensitivity monitoring of physiological signals, as these devices are
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becoming more accessible to researchers in artificial emotional intelligence, we are transitioning to an era where machines will start to gain a sense of empathy as they process affective information from their users in ambient.

18.8 Brain-Computer Interfaces and Neuroprosthetics

The brain is a primary coordinator of the nervous system. It receives and sends sensory and motoric signals from and to the rest of the body through the spinal cord. It is a powerful tool that allows us to perceive, process, and store information. Studying the brain dynamics will help us to understand how the brain responds to particular stimuli or how changes in brain structure can influence a person's personality, cognitive behavior, or well-being. It could also potentially enable future computers
to precisely identify our thoughts, predict our intentions, and recognize our emotions. Neuroimaging
tools such as functional magnetic resonance imaging (fMRI), functional near-infrared spectroscopy (fNIRS),
positron emission tomography, functional transcranial Doppler sonography (FTCD),
magnetoencephalography (MEG), and EEG have consequently been used to map and monitor brain
activity by measuring localized changes in hemodynamic response or recording electrical currents
and magnetic fields due to neuronal activations (Duschek and Schandry 2003; Shibasaki 2008). Brain
stimulation techniques, such as transcranial magnetic stimulation (TMS), deep brain stimulation
(DBS), electroconvulsive therapy (ECT), and transcranial direct current stimulation (tCDS) have also
been used in research and medical settings, mainly for the treatment of mental disorders (George et al.
2002; Fregni et al. 2006).

Most of these tools are currently operator-dependent and require a large setup. These challenges limit
their usage for wearable, real-time, and long-term applications. Researchers have recently developed
non-invasive systems such as screen-printed flexible receiver coils for magnetic resonance imaging
(Corea et al. 2016) and wearable fNIRS with tCDS (McKendrick et al., 2015). Commercial and personal
EEG headsets such as NeuroSky and Emotiv also exist and have been widely used in brain-computer
interface (BCI) research and clinical trials (Swan 2012). However, to achieve a high SNR and direct
contact with the brain, spinal cord, peripheral nerves, or even specific neurons, device miniaturiza-
tion, modification efforts, and its translation into implants are required. Multi-modal, neuromodula-
tion capability also needs to be incorporated to facilitate bi-directional communication. This section
will cover recent development of various flexible and stretchable devices for neural reading, stimulation,
and neuroprosthetics. The mechanically compliant structure and intrinsic property of these devices
bridge the biological, mechanical, and geometrical mismatch between electronics and our body tissue,
facilitating seamless and intimate coupling between the two.

18.8.1 Electroencephalography

In EEG, the electrical signals induced from the neural activity are spatially read by a set of electrodes,
attached around the scalp. The electrical signals, which can be classified into several frequency bands,
could give insights to human cognition, emotion, and attention (Ray and Cole 1985). Even though cur-
tent EEG systems are wearable, they are still bulky and uncomfortable, restraining them from daily
use for long-term monitoring of brain signals. Norton et al. (2015) explored the applications of an EES
attached to the auricle for BCI applications. The Au electrodes are designed with stretchable filamen-
tary serpentine traces, deposited on a PI, and then transferred to an elastomer. Placing these electrodes
around the curvilinear region of the ear (Figure 18.21b) presents a novel and comfortable alternative
than attaching them around the scalp, which is enabled by the conformal design of the electrodes. Bio-
compatibility and long-term wearing tests of the device both show no adverse effects on the skin. A sec-
ond structure, with tripolar concentric ring electrodes in a capacitive configuration, is also designed for
an improved spatial resolution and durability.

The EEG system, as shown in Figure 18.21a, consists of three electrodes: recording (REC), refer-
ence, (REF), and ground (GND) with stretchable interconnects to the read-out and processing system.
The signal acquisition system will then amplify the EEG signals, and the computer will process these
signals to extract suitable features for classification (Figure 18.21c). To demonstrate its application for
BCI, steady-state visually evoked potential (SSVEP) approach is tested on several subjects. The SSVEP
test triggers optical stimulation that flicker around 6–10 Hz to the retina. The electrical signals gener-
ated by the brain during the visual stimulus can be used to monitor eye gaze. The SSVEP BCI ultimately
allows the subjects to interact with a text speller software and populate a word from a group of letters.
Other applications of EES are also demonstrated by Kim et al. (2011). By applying an EES on the fore-
head, alpha brain waves of around 10 Hz can be observed to reflect opening, closing, and blinking of the
subject’s eyes. A Stroop test to study human cognition is also investigated, by performing demanding
tasks and analyzing reaction time of the subjects based on their EEG signals.
Due to noise signals and attenuation caused by the skin and bone barriers, research has also been conducted to fabricate an array of electrodes that can be laminated directly on the surface of the brain to perform electrocorticography (ECoG). Conventional invasive neural recordings and stimulations have mainly relied on rigid materials, such as silicon or metals which have a mechanical, geometrical, and biological mismatch to the brain tissues (Jog et al. 2002; Kim et al. 2009). These incompatibility issues could
result in hemorrhage and inflammatory response. They could damage target tissues and influence the clarity of neural reading (Polikov et al. 2005; Saxena et al. 2013). Flexible and stretchable, minimally invasive implantable electronics could solve these challenges with a greater precision, sensitivity, as well as spatial and temporal resolution. Several researchers used a polymer layer as a substrate for micro-fabricated sensors and circuits. One example is a flexible multi-channel electrode array to simultaneously monitor ECoG and local field potential (LFP) in the visual cortex of a rat (Toda et al. 2011). The device consists of gold electrodes encapsulated in a Parylene C substrate. Visual stimulus from a monitor evoked a neural response from the rat, and recording results showed reliable data throughout the 2 weeks implantation.

Another work by Viventi et al. (2011) produced a high-density, flexible electrode array integrated with multiplexing circuitry for high resolution (micro-ECoG or μECoG) neural reading. Transistors are fabricated from silicon nanomembranes and transferred into a PI substrate. Interconnects and vias are then deposited layer by layer with PI encapsulation that connects to the outermost platinum (Pt) electrodes. The multiplexer enables addressing mechanisms that minimize connection numbers to access a total of 360 active electrodes. As illustrated in Figure 18.21e, this compliant, foldable, and flexible device allows high-resolution spatiotemporal neural activity reading even in a commonly inaccessible region of the brain. Other researchers have also incorporated bioresorbable and transient materials such as silicon and silk in their design strategies. The usage of these materials enable ECoG systems that can dissolve with the cerebrospinal fluid (Kim, Viventi et al. 2010; Kang et al. 2016).

18.8.3 Multi-modal Implantable Interfaces

The previous examples of flexible 2D devices provide non- to minimally invasive approaches to neural reading. Yet, there are some cases when it is necessary to penetrate further to target specific regions for simultaneous neural reading and modulation. New materials and fabrication strategies have motivated researchers to develop a 3D out-of-plane structure of probes or injectrodes that are bio-compatible and compliant with the mechanical properties of the nerves. Figure 18.22 shows many existing platforms and techniques of multi-modal neural interfacing that can penetrate to delicate regions or conform to the curvilinear surface of the brain, spinal cord, and peripheral nerves while alleviating tissue damage and foreign-body reaction (Lacour et al. 2016). Besides conventional electrical stimulation that uses an electric current to excite surrounding neurons, and delivery of biochemical agents to affect neurotransmission, a new modality for neuronal activations through the flow of photons has emerged. Optical stimulation through optogenetics provides a means to activate specific neurons that exhibit light-responsive ion channels by genetically modifying them with reagents such as channelrhodopsin (Grill et al. 2009; Fenno et al. 2011).

One example of multi-modal neural interfaces is a flexible injectrode for stimulation, sensing, and actuation of the brain soft tissues developed by Kim et al. (2013). The injectrode consists of several layers of microdevices such as an electrode, inorganic photo-diode, and LEDs, as well as a bi-functional temperature sensor/heater on a plastic strip. They are then wrapped in a removable, bio-degradable substrate. The needle-type devices discussed here are commonly used for deep brain recording and stimulation; therefore, they should be sharp and pointed to be able to move freely around the cerebrospinal fluid. The micro-scale size and structure of these devices proved to improve spatial targeting and reduce gliosis after implantation. An injectable micro-LED with near-field radio-frequency (RF) coil also serves as a stand-alone, wireless optoelectronic subdermal implant (Shin et al. 2017). The system consists of several devices with Au-interconnects on PI that are encapsulated in Parylene or PDMS. A soft and stretchable version of this optoelectronic implant is demonstrated in Figure 18.23a,b (S.I. Park et al. 2015). The deformable feature of the device allows it to be intimately attached to the epidural space of the spinal cord or peripheral nerves (Figure 18.23c). An external coil induces an electromagnetic field that couples with the receiver coil to power and communicate with the implantable system. Through wireless capacitive or inductive coupling, these devices bypass the use of batteries or wired connections and correspondingly facilitate free-moving and long-term in vivo trials.
Instead of using an LED with patterned metal interconnects on a long strip of polymer, research performed by Lu et al. (2017) and Canales et al. (2015) exploited thermal drawing techniques to develop flexible and stretchable fiber-based neural probes. The thermal drawing of a polycarbonate and cyclic olefin copolymer (COC) with the introduction of a conductive polymer composite resulted in a flexible multi-modal all-polymer fiber probe (Figure 18.23d,e). The implantable, fiber-based neural probe facilitates *in vivo* optical stimulation, drug delivery, and electrophysiological recording through the delicate regions of the brain with pinpoint accuracy. Figure 18.23f shows a long-term trial of the fiber probes in freely moving Thy1-Chr2-YFP mice. They retain both recording and stimulation capabilities for 2 months. These multi-modal injectrodes have the potentials to advance tools for manipulation and analysis of brain circuits, particularly for the treatment of neurodegenerative disorders. For instance, they could be integrated with wirelessly controlled pumps that could deliver micro-liters dose of drugs on-demand (Dagdeviren et al. 2018).

As an alternative approach to high spatiotemporal monitoring of neural activity in the deep brain, Liu et al. (2015) introduced a novel, syringe-injectable electronic device. This device consists of an array of Pt-electrodes with metal interconnects, encapsulated in a polymer material, and fabricated in a mesh structure. The size and flexibility of the device facilitate compact storing of a dense, large-volume mesh
network in a syringe (Figure 18.23h), and after injection, enables the mesh structure to unfold and three-dimensionally conform to the surrounding tissue (Figure 18.23g). Injection of the mesh into the hippocampus of anaesthetized mice confirmed the ability of 16 channels to record brain electrical activity (Figure 18.23i). The mesh structure of this device provides seamless, unobtrusive interface with brain tissue and results in a better chronic immune response compared to the other flexible thin-film devices (Zhou et al. 2017).
18.8.4 Soft Neuroprosthetics

Soft implantable neuroprostheses aim to substitute or restore sensory, motor, or cognitive function that could have been damaged due to injury, ageing, or neurodegenerative diseases. Due to their adaptive structures, they function while alleviating foreign body reaction and maintaining long-term stability. One of the most recent applications of soft neural implants is to restore motor or sensory pathways due to spinal cord injury. Electronic dura mater (E-dura) is a multi-functional subdural device that can simultaneously perform electrical sensing, stimulation, and biochemical drug delivery (Minev et al. 2015). This soft implant has been clinically tested in a spinal cord injury case of a mouse. By precisely delivering electrical stimulation around the lumbosacral segment of the spine and injecting serotonergic agents, the E-dura can restore paralyzed rats and control their locomotion behavior.

The progression of soft electronics for the human body has also promoted a new generation of artificial organs integrated with a neural feedback. One example is an organic skin prosthesis that mimics how a biological mechanoreceptor functions (Tee et al. 2015). The skin-inspired digital mechanoreceptor consists of a micro-structured resistive pressure sensor with sensitivity approaching the human skin capability, a printed organic voltage-controlled oscillator circuit, and an optogenetic neural interface system. The CNT-based, piezoresistive pressure sensor converts physical inputs to electrical signals that oscillate and change in frequency through a three-stage ring oscillator. To model similar action potentials response during human tactile stimulations, the sensor and circuitry are engineered to output a maximum frequency of 200 Hz. The pressure-dependent frequency signals given by the digital mechanoreceptor successfully stimulate neurons in the primary somatosensory cortex region of a mouse either electrically or optically.

Another research, as shown in Figure 18.24a, presents a multi-sensory prosthetic skin integrated with platinum(Pt)-NWs on a stretchable multi-electrode array (MEA) to interface with the peripheral nerves (Kim Jaemin et al. 2014). The stretchable MEA allows the relay of sensation from each sensor to its corresponding peripheral nerve through electrical stimulation. A microcontroller unit processes pressure sensor signals and as a feedback, controls stimulation voltage of an electrode on a sciatic nerve (Figure 18.24b,c). Amplified and filtered EEG signals from the ventral posterolateral nucleus (VPL) in the thalamus are simultaneously read to confirm a successful afferent signal transmission from the sciatic nerve to the brain.

For intimate, effective, and robust long-term interfacing with the peripheral nerves, bio-compatible compliant electrodes with low impedance must be developed. Metals such as tungsten, gold, platinum, and iridium have been adopted as the electrodes, typically deposited or embedded in an insulating substrate such as PET or PI (Geddes and Roeder 2003; HajjHassan et al. 2008). Figure 18.24d shows conformal lamination of the PtNWs/Au MEA on peripheral nerve and muscle tissues of a rat model. To reduce the impedance of the electrodes, PtNWs are grown on the MEA using an electrochemical method. Ceria nanoparticles are also introduced on the PtNWs to suppress and prevent inflammation caused by reactive oxygen species. The PtNWs/Au MEA is then transfer-printed onto a stretchable and thin PDMS substrate. The prosthetic skin has a potential for burnt victims or individuals with a prosthesis to restore their skin sense of touch, pressure, and temperature (Figure 18.24e).

Inspired by the structure of the human eye, Choi et al. (2017) proposed a soft retinal implant. As illustrated in Figure 18.24a, the working mechanism of a human eye starts with a lens that captures incoming lights. The retina then converts these lights into action potentials and transmits them to the brain through optic nerves. To overcome the mechanical mismatch between soft retina and rigid electronic devices, an array of optoelectronic devices is configured in such a way to mimic the hemispherical structures of the human eye. The system induces minimal stress and deformation to an artificial eye model (0.61 MPa), which is orders-of-magnitude lower than the other approach (Figure 18.24g). The image sensing component consists of hemispherical curved arrays of MoS$_2$−graphene phototransistors. The property and high-density design of the phototransistors eliminate infrared (IR) noises and enable high-quality imaging.
FIGURE 18.24  Silicon nanoribbon electronics for skin prosthesis. (a) System illustration of the multi-sensory prosthetic skin interfacing to the peripheral nerves through MEA. Inset shows the stretchable MEA, which consists of PtNWs with adsorbed ceria NPs grown on Au electrodes. (b) Flowchart of peripheral nerve simulations based on pressure sensor signals processed by microprocessor with cross-validation by EEG recordings of the ventral posterolateral nucleus (VPL) of the thalamus in the right hemisphere (inset). (c) Signals from pressure sensor (top) trigger and affect the stimulation amplitude to the peripheral nerve (centre, red, and blue). Correspondingly, neural activations can be observed from the VPL through EEG signals (bottom). (d) Optical image of the conformal stretchable MEA on peripheral nerves (blue arrows) in a rat model. (e) Several applications demonstrated with the multi-sensory prosthetic limb including typing on a keyboard, grasping a ball, and grabbing a cold mug. (With kind permission from Springer Science+Business Media: Nat. Commun., Stretchable Silicon Nanoribbon Electronics for Skin Prosthesis, 5, 2014, 5747, Kim et al.) Eye-inspired soft optoelectronic system as retinal implant. (f) Illustration of the ocular structure of human integrated with the soft optoelectronic device. (g) Stress induced by the soft optoelectronic device compared to other technologies. (h) Optical image of CurvIS array and UNE connected by soft flexible printed circuit board (FPCB). Magnified image shows three pixels array. (i) Layer configuration of phototransistor (bottom) and stimulation electrode (top) separated by a PI isolation and connected to the soft FPCB. (j) The soft FPCB consists of a system that receives optical signals from CurvIS array, amplifies them, and electrically stimulate through the UNE to the optic nerves (k) of a rat correspondingly. Digital ECoG system is used as a cross-validation of the neural stimulation. Transient voltage graph of the evoked spike (l) and LFP changes (m) in the visual cortex by electrostimulation. (With kind permission from Springer Science+Business Media: Nat. Commun., Human Eye-Inspired Soft Optoelectronic Device Using High-Density MoS2-Graphene Curved Image Sensor Array, 8, 2017, 1664, Choi, C. et al.)
A system to interface with the curved image sensor (CurvIS) array as a soft implantable optoelectronics is shown in Figure 18.24h. A soft and flexible printed circuit board (PCB) processes optical information from the LED to the image sensor and correspondingly applies electrical pulses to a set of Au neural-interfacing electrodes (UNE). The UNE is located on the back side of the image sensor array and facing towards the retina (Figure 18.24i). The soft PCB consists of an amplifier and microprocessor unit that processes the signals from the CurvIS array, makes decisions, and actuates the UNE (Figure 18.24j). ECoG reading is conducted to validate the functionality of this soft optoelectronic implant by simultaneously reading the LFP changes at the visual cortex of the rat as optical stimulation is pulsed to the CurvIS array (Figure 18.24k). From these discoveries, it can be observed that soft implantable devices could pave the way towards the future of electronic implants, digital nervous systems, and thought-controlled prostheses for human augmentation and assistive technology.

### 18.9 Conclusion

In the last few decades, we have witnessed the emergence of new technologies that transform personal electronics and reconfigure the way we understand ourselves and interact with the world. With the rapid advances in flexible and stretchable bio-integrated devices, we have come to a new age where electronics are becoming much closer to the body, breaking boundaries between the synthetic and biological systems. This chapter highlights the role of various novel sensors, actuators, and transducers that can be conformably laminated to the curvilinear surface of the skin, embedded into textiles, or implanted into the body for seamless body augmentation, as well as intimate physical activity recognition and physiological monitoring. New materials, device designs, and microfabrication techniques allow these electronic devices to be flexible, stretchable, imperceptible, bio-compatible, self-healable, and transparent which improves their sensitivity, comfortability, robustness, long-term usage, and social acceptability. Integrating these devices into different regions of the human body internally and externally enables various applications of human-machine interfaces from on-skin sensing and display, tactile feedback, body gesture monitoring, speech recognition, emotion and stress recognition, brain-computer interfaces, to neuroprosthetics.

Future challenges include system-level efforts of heterogeneous integration of sensors, actuators, and transducers with other electronic modules, such as wireless communication, circuits, memory, energy storage, and energy harvesters. Large-scale fabrication will enable high-coverage of active areas that could entirely cover skin or brain surface, for example. Multi-modal, dense array of sensing and actuation elements will provide a high-resolution spatiotemporal mapping and actuations with high recognition rate and pinpoint accuracy. In addition to these, extensive clinical studies must be conducted, particularly for implantable devices to gain regulatory approval. It is, therefore, an exciting time for materials scientists, engineers, medical practitioners, computer scientist, and interaction designers to work together and push forward wearable and implantable electronics through flexible and stretchable devices and revolutionize the future of human-machine interfaces.

### References


Flexible and Stretchable Devices for Human-Machine Interfaces


