CHAPTER

# 3.1

# Knitted Electronic Textiles

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## 1. FROM FIBERS TO TEXTILE SENSORS

From the time of birth, fabric is the first and the most natural interface for the body, a soft, warm, and reassuring material that protects our skin from the environment. Clothing usually covers more than 80% of the skin, which is why textile material can be seen as the most appropriate interface to implement new sensorial and interactive functions. Functions like sensing, transmission, and energy generation are implementable through textile technology. Functional yarns and fibers can be used to manufacture garments where electrical and computing properties are combined with the traditional mechanical characteristics of fabric, giving rise to electronic textile (e-textile) platforms that are mechanically comparable with the textiles that are normally used to produce our garments.

Electrical conductivity is the main physical property that is capable of transforming a textile material into a sensing material and that plays an important role in the development of e-textile apparels. Conductive fabrics can be used as bioelectrodes or (when combined with elastomers) as piezoresistive sensors that are capable of sensing biomechanical variables. Several different methods can be used to construct an electrically conductive textile, starting from the integration of metal monofilaments into the yarn, the enrichment of the fiber with conductive components, and the coating of man-made fibers with a conductive layer, to the printing of conductive pigments onto the fabric surface.

A yarn can be defined as a linear assembly of fibers or filaments arranged into a continuous strand, with textile characteristics such as tenacity and flexibility. Conductive yarns are generally made with conductive inorganic components combined with traditional textile fibers. Metals can be used in the form of fibers blended in the pre-spinning stage or in the form of filaments that can be mixed with other yarns at doubling, knitting, and weaving stages (Figure 1(a), (b)). Conductive bicomponent fibers composed by a matrix polymer and a conductive layer can be manufactured using the conjugate fiber spinning

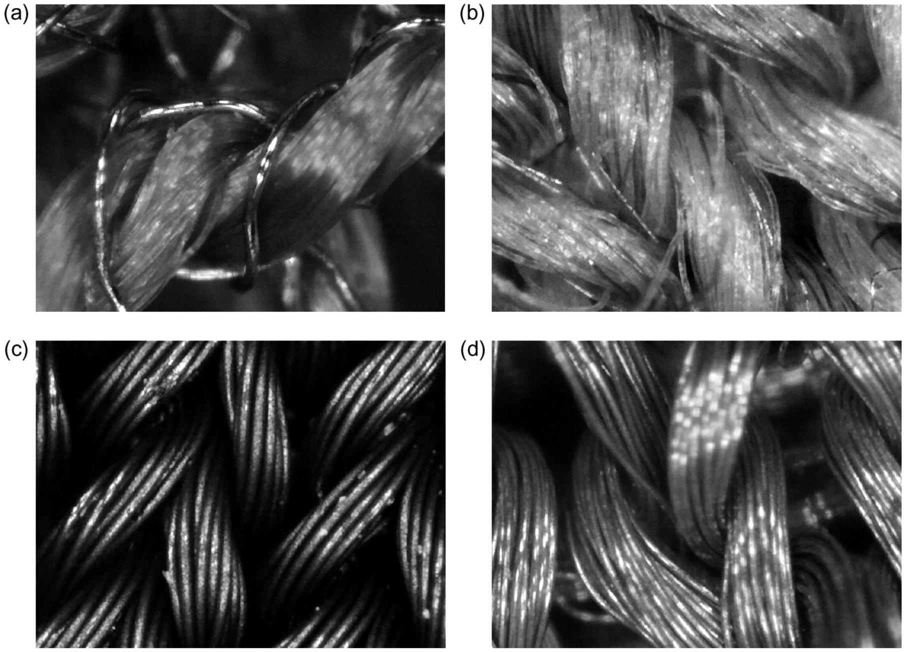


**FIGURE 1** Conductive yarns and fabrics: (a) enlarged view of a fabric made with a conductive yarn obtained with stainless steel wires twisted around cotton and elite fibers; (b) enlarged view of a fabric made with conductive stainless steel fibers blended with polyester fibers; (c) enlarged view of a fabric realized with conductive bicomponent fibers; and (d) enlarged view of a fabric realized with silver-coated fibers.

technology. In this case, the conductive layer contains a densely embedded carbon black or white metal compound as conductive particles. Such fibers can be blended with other fibers to make the resultant fabric sufficiently conductive. Conductive nylon and polyester are available in the form of filaments and staple fibers (short fibers that need to be spun into yarn, Figure 1(c)). Another possibility is the use of coated conductive fibers, as shown in Figure 1(d). The coating can be applied through various techniques. Highly conductive fibers can be produced by metallic or galvanic coating, but these methods have some limitations in terms of adhesion and corrosion resistance and suitability of the substrate, while metallic salt coatings have some limitations in terms of conductivity.

In conventional textile production, metal components in the form of fibers, filaments, or particles are typically used for technical applications such as shielding and antistatic protection (work clothes, dust-free garments, school uniforms, dress suits, sweaters, carpets, upholsteries, car seats, blankets, curtains, and static-free brushes for cleaning office equipment), in bacteriostatic applications (for apparel and furnishings), as well as for fashion since the presence of metal changes the mechanical properties of fabric, creating a wrinkle or shaping effect. Pure stainless steel slivers can be blended with fibers such as polyamide, polyester, and cotton at the spinning mill to obtain electrically conductive yarns in a wide range of yarn counts. Such yarns and fibers are corrosion-resistant, inert, and stable, and can guarantee long life.

From the perspective of conductivity, manufacturability, and textile handling, silver-plated fibers may be considered the best option for the production of conductive fabric sensors to be worn close to the body. However, poor washability and poor resistance to strain due to the development of stress cracks during cladding, as well as sweat oxidization problems, make their life span shorter than fibers and yarns made of stainless steel. Several more stable products based on silver have appeared on the market in the last decade (XStatic [1], Shieldex [2]) and have been used for sensing applications, such as the women's SuperNova Seamless Glide bra by Adidas and the H2 heart rate sensor by POLAR.

Another important property required for sensing applications is elastic recovery of the fabric, which is the result of combined use of elastic and functional fibers. The elastomeric fibers are those fibers that possess extremely high elongations at break and that recover fully and rapidly from high elongations. These fibers are normally used in applications where high elasticity is necessary within the textile structure. An elastic yarn may consist solely of a number of elastomeric fibers combined to make a "bare" elastomeric yarn, such as spandex, where each fiber is made up of many smaller individual fibers that adhere to one another due to the natural stickiness of their surface [3]. Alternatively, the yarn may use the elastic strand as a core in a composite yarn having inelastic staple fibers as an outer covering. A yarn such as this is said to be "core spun." The use of a core spun elastomeric yarn in the fabric improves appearance, handling characteristics, shrinkage control, color fastness, control over elongation, and the power of recovery. In addition, the outer covering, which may be composed of natural fibers such as cotton, man-made fibers such as polyester or polyamide or a combination of both, provides additional breathability to the fabric in which it is used.

Conductive elastic yarns can be manufactured through different processes. Usually, the final structure comprises at least one elastic core thread, at least one electrically conductive thread that is wound around the core thread or is core spun around the elastomer, and, if it is needed to insulate the conductive filament, one non-electrically conductive yarn that is wound around the whole core structure. Elastic components are used for the production of stretchable fabric that allows creating apparels or garments capable of fitting the body shape as a second skin. This is an important property as a textile sensing surface collects the information from the human body by fitting it in a comfortable and unobtrusive way. Stretch fabrics are manufactured by using elastic yarns or by mixing elastomeric filaments with other natural or man-made fibers at doubling, knitting, and weaving stages. The result is a fabric that is not only stretchable, but may also have the desirable characteristics of wrinkle-resistance, transpirability, and washability.

## 2. THE INTERLACED NETWORK

Fabrics can be manufactured with different technologies from knitting to weaving. The path followed by a single yarn in the fabric is different according to the production process. Knitting methodology consists of loops called stitches that are pulled through each other. The active stitches are held on a needle until another loop can be passed through them and a single yarn is building the whole structure. Weaving is a process where two distinct sets of yarns or threads, called the warp and the filling or weft, are interlaced with each other to form a fabric or cloth. The warp threads run lengthways on the piece of cloth, and the weft runs across from side to side.

When yarns are combined into the fabric, the resulting structure is a network where the contacts among the single filaments are random, and the fibers are untidily assembled. Compared to weaving that requires running the same yarn along the weft and the warp direction, knitting technologies are more suitable for smart fabrics. First, knitting provides the possibility to select a desired functional yarn and confine it in a specific region of the fabric according to a precise architecture. Second, knitted fabric is usually highly elastic (it stretches easily) and readily drapeable (it hangs and folds nicely), and it is porous. This means that it is breathable and comfortable when worn next to the skin. Finally, knitted fabric can be made in the required garment shape.

Machines and looms for knitting can be defined according to needle arrangement. Needles are fixed on metal structures that can have a circular or linear shape. In knitting machines the needles are free to move individually, while in knitting looms the movement of needles is done jointly (Figure 2). The set of needles and metal structures is known as the "bed." Both machines and looms can be equipped with single or double beds.

Computerized flat knitting and circular knitting machines allow operating each needle individually. Factors such as knitting speed, yarn tension, cone size, batch difference, and humidity can contribute to variations in loop formation. A flat knitting machine is very flexible, allowing complex stitch designs, such as jacquard, plated, and intarsia patterns, double jersey, shaped knitting, and precise width adjustment. This type of machine is,

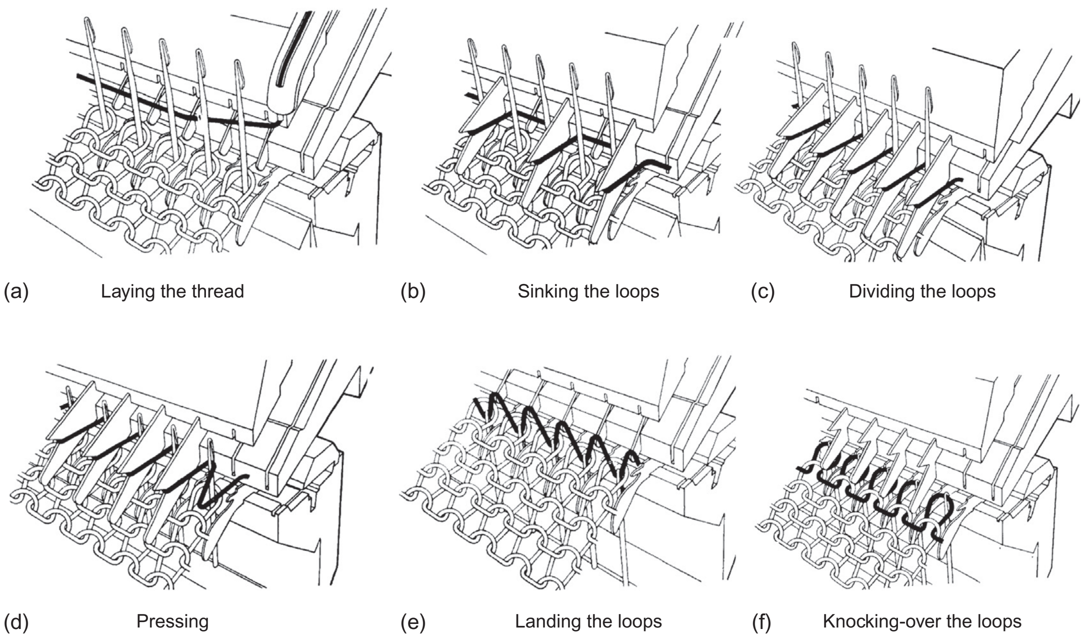


**FIGURE 2** Movement of knitting loom elements to produce a course of loop [4].

however, relatively slow when compared to circular machines. Machines are usually equipped with several independently motorized yarn carriers that allow the electronic selection of each single needle. For an e-textile application, intarsia is one of the most important knitting techniques used to create patterns with multiple diversified yarns. Double-bed knitting machines can also be used to realize the double-jersey technique where two layers of fabric are knit simultaneously. The fabrics may be inseparable, as in interlock knitted fabrics, or they can be knit as two unconnected fabrics, as in tubular knitted fabric.

Another interesting feature of knitting is the possibility of handling two different yarns concurrently with the same needle and overlapping, known as the *vanisè* technique or plated knitting. With this technique a metal yarn can be covered (plated) with another one (Figure 3(a)), and with the second series of needles, the external side of the fabric can also be protected. Using a combination of different knitting techniques offers the possibility of designing and implementing a sort of logic circuit. With conductive patterns that can be on a specific side of the fabric, these patterns can be connected through fabric tracks where the conductive elements are running inside the fabric, invisible from both the sides of the fabric, in a multi-layered structure (Figure 3(b)).

Circular knitting, and in particular the seamless technique, provides comfortable, stretchable, well-fitting, and adherent garments, which makes this technology suitable for sensing applications where adherence, elasticity, and comfort are the main requirements. The seamless technique comes from the fusion of two fields, hosiery, and knitting, and allows the production of tubular fabrics without seams, laid-in elastic yarns inserted in the welt bands and equipped with areas having gradual compression. Therefore, the garments knitted on seamless machines merge comfort with functional performance as they allow the creation of different stitches such as rib, net, jacquard, piquet, stripes, and laces, as well as pre-shaped structures, hidden supports, pockets, collars. and hoods. It is possible to implement seamless systems, where both electrodes and sensors are knitted in the same production step, through intarsia technology. The main difference between circular knitting and flat knitting is that the latter can combine intarsia and double knitting, while the

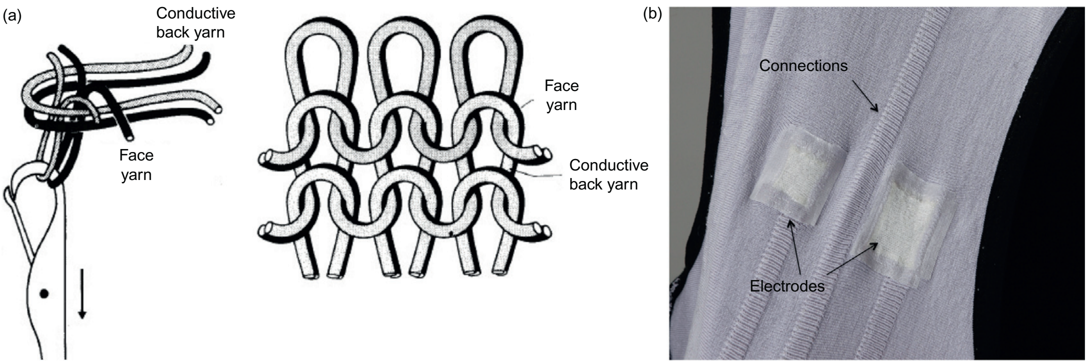


**FIGURE 3** Plated knitting: (a) plating in weft knitting [4] and (b) fabric connections. The conductive yarn is knitted inside the multi-layer structure and fabric electrodes with membrane.

former can only handle these two processes separately. However, seamless technology is unique in combining elasticity and comfort of the fabric with low production costs.

Different knitting techniques can be used to provide different sensing functionalities. In particular, piezoresistive fabric sensors can be implemented using knitting techniques by combining conductive and elastic yarns, and by an industrial serigraphy screenprinting process. Circular knitting machines (such as Santoni machines) can be used for the production of piezoresistive fabric sensors due to the high elastic recovery of fabric made with this technology. Conductive bicomponent fibers yarn, such as polyamide loaded with carbon particles, is one of the yarns typically used in combination with one or more elastomers to implement these sensors. Piezoresistive fabric sensors change the electrical resistance according to strain, and the variation of the electrical properties occurs due to the different path of the electrical current inside the fabric structure. Usually this property can be observed in stretchable fabric, where a mechanical solicitation affects the flow of carriers inside the structure. When the conductivity of the yarn is due to the presence of conductive particles (like in a bicomponent fiber), the elongation of the yarn produces different distribution of the conductive particles in the structure of the yarn and in the fabric, leading to a modification of charge transport mechanism. The interconnection among fibers and stitches is affected by the mechanical deformation, as can be seen in Figure 4. The elongation of the fabric modifies the distance among the stitches as well as the arrangement of the fibers in the yarn.

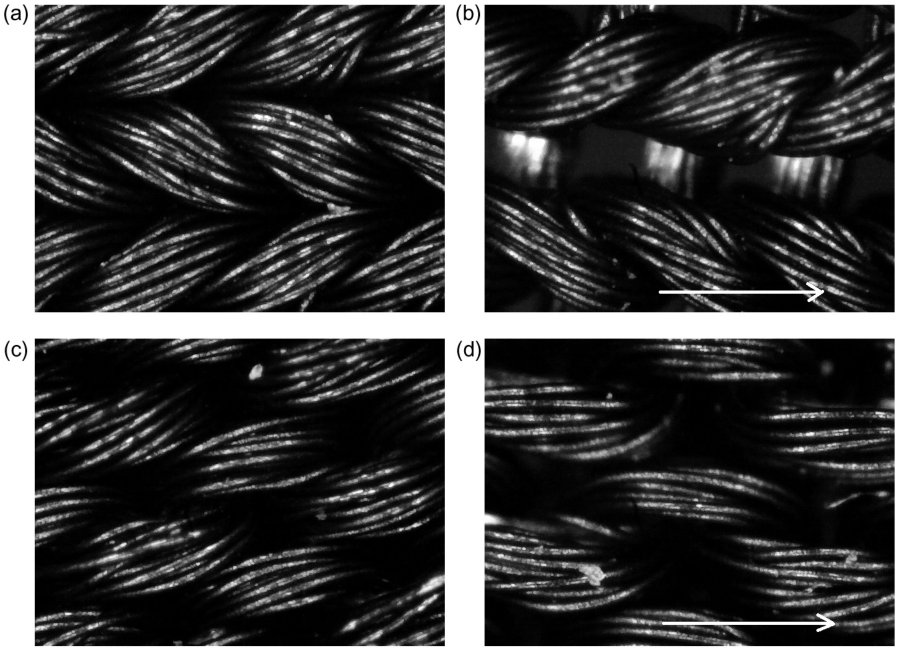


**FIGURE 4** Interconnections in fabric under mechanical deformation: (a) front view, rest condition; (b) front view, under mechanical strain along the arrow direction; (c) back view, rest condition; and (d) back view, under mechanical solicitation, fibers are elongated along the arrow direction.

The path of charges inside the textile structure is correlated to the distance among the conductive components of the network. When the structure of the yarn and the resulting fabric is modified by a mechanical solicitation, the fabric sensor behaves as a strain-gauge transducer within the limits of sensor elasticity such that it does not break or permanently deform. Modeling of the charge transport mechanisms inside the textile structure is not easy, but when the structure of the yarn and the resulting fabric is kept planar to the extent possible, the fabric sensor can be considered a strain-gauge transducer. The different architecture of the yarn and resulting fabric confers a different response to the whole sensor. A recent study [5] demonstrated that a small change in the structure of the fabric, due to a different geometry of the conductive yarn, results in a dramatic change of the functionality and sensitivity of the final fabric sensor.

The lithographic technique is normally used to implement piezoresistive-coated sensors. During the process, a rubber or silicone solution containing conductive particles is applied to the fabric, then, after the removal of excessive rubber material, the conductive elements are immobilized in the structure through treatment at high temperature. This technology provides both sensors and wiring by using the same elastic material and avoids the use of obtrusive metallic wires, which may hamper the movements of the kinematic chain. The mechanical properties of the final product are affected by the speed of the coating process, the viscosity of the solution, and the capability of the material used as substrate to adsorb it [6]. The viscoelastic properties of the fabric substrate affects the mechanical response of the textile sensor. The hysteresis effects can be reduced by acting at the level of the textile structure, as well as by increasing the elastic properties of fabric. These fabrics behave as strain-gauge sensors [7] and show piezoresistive properties similar to knitted fabric sensors. In both cases, the increase of elastic properties in the fabric is directly proportional to the increase in piezoresistive properties.

## 3. TEXTILE SENSORS FOR PHYSIOLOGICAL STATE MONITORING

A wearable sensor system based on textile technology, where sensors are implemented with fibers and yarns, requires construction of a fabric containing shaped regions where sensors are located on a specific part on the body. Thus, the textile-sensing interface has to be a garment tightly fitting the body, to avoid any possible mismatch between the body and the sensors. The fabric will act as a second skin, and it has to be elastic and comfortable. By using a flat knitting and seamless knitting technique it is possible to confine specific yarns in defined regions of the fabric and at the same time to process different yarns together according to a desired topology.

Sensors, electrodes, and connections can be fully integrated in the fabric and produced in one single step by combining conductive and non-conductive yarns [8]. A combination of intarsia and double-knitting techniques allows the production of double layers, using the external non-conductive part to isolate the electrode from the environment. Use of another yarn in vanisè configuration allows multi-layered structures, where the conductive surface is sandwiched between two insulated textile surfaces. The same conductive yarn can be used for the electrodes as well as for the implementation of the connections, as seen in Figure 3.

Conductive textile is the basic material for the detection of electrical signals. In standard clinical practice, electrodes located on specific parts of the body are used to measure electrical potentials of biological origin. Biopotentials occur due to electrochemical activity of the cells, where the electrical activity is caused by differences in ion concentrations within the body. There are several diagnostic applications of biopotentials: electrocardiography (ECG), electroencephalography (EEG), electromyography (EMG), and electrooculography (EOG). Biopotential electrodes convert ionic conduction to electronic conduction and are used for measuring electric potential of biological origin, or to transmit electrical energy to and from a human subject.

Textile electrodes can be used to detect a variety of biological signals [9] as well as to measure body impedance and skin conductance. Piezoresistive sensors can be used to monitor respiratory activity at the thorax and abdominal level as well as movement of joints. Other important parameters like temperature or $SpO_2$ can simultaneously be measured by means of transducers embedded into the fabric. The list of signals detectable with non- invasive textile or wearable sensors varies according to a specific application, but some of the most important information about the physiological state of the user is provided by cardio-pulmonary activity.

Electrodes can be classified according to conduction mechanism as “perfectly” polarizable, characterized by a capacitive effect, with no charges flowing between the electrode and tissue (i.e., stainless steel electrodes), and “perfectly” non-polarizable, characterized by a resistive effect, with free charges flowing between the electrode and tissue (i.e., Ag/AgCl). Electrodes are applied directly on the body surface; in order to get a good skin contact, a thin layer of electrolyte is usually applied between the skin and the electrode. Electrodes can be manufactured using different materials. In the standard practice, pregeled disposable electrodes are typically made using Ag/AgCl and are employed in long-term ambulatory monitoring and when high stability is needed. An alternative solution, compatible with textile technology, is represented by stainless steel polarizable electrodes; in this case, an electrolyte containing salts of low corrosion potential may be required.

If a pair of electrodes is in an electrolyte and one moves with respect to the other, a potential difference appears across the electrodes. This potential difference is known as a motion artifact and it acts as a source of noise and interference in biopotential measurements. Motion artifacts are greatly attenuated if the electrode is separated from the skin surface by a layer of conductive gel or paste. Any mechanical disturbances caused by relative motion between the electrode and the skin are damped by the gel layer, and their effect on the signal is limited. Moreover, artifacts can be reduced by reducing the skin impedance, such as removing the upper layers of the skin via abrasion. The standard practice foresees cleansing of the skin with solvents and use of a conductive paste for reducing skin impedance.

Conductive fabric electrodes can be used together with hydrogel membranes that act as electrolyte by reducing the contact resistance between the skin and the electrode. In order to improve the stability of the contact, the hydrogel membrane is used in the form of a patch that is adhesive on both sides as shown in Figure 5; this feature improves the quality of signal by reducing motion artifacts and the effects of the contact impedance [10,11].

Since the use of textile materials for wearable sensing culminates with the implementation of imperceptible monitoring systems, it should not be any different from traditional

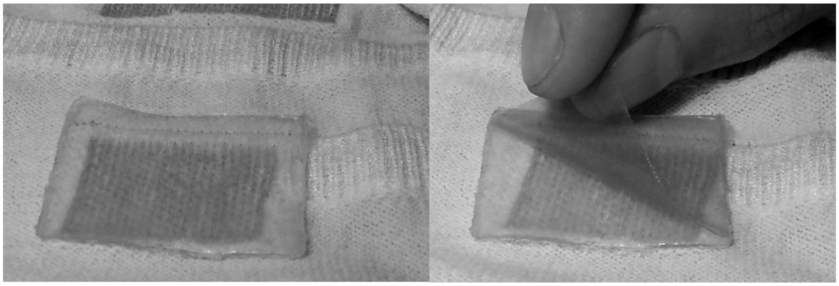

**FIGURE 5** Textile electrode with a hydrogel membrane.

garments or fabric, and the use of a material like gel is not desirable. Several factors have to be considered to implement electrodes able to sense without hydrogel. First, such an electrode cannot acquire information of clinical quality, instead it can give daily indication of health status and the physical behavior of the user in a context that cannot be monitored in another way, without interfering with the user's daily life. Second, the sensor locations should be selected according to ergonomic criteria at the locations that minimize motion artifacts. Third, contact with the skin should be improved by utilizing the elastic properties of the garment. Fourth, local environmental conditions in terms of temperature and humidity in the electrode region should be manipulated to increase sweat production, e.g., by decreasing the breathable properties of the fabric. Sweat can act as an electrolyte to decrease skin-electrode contact impedance. Finally, the fabric should be designed to fit tightly to the skin.

## 4. BIOMECHANICAL SENSING

Knitted piezoresistive fabric (KPF) sensors are based on the piezoresistive effect that is characterized by the change of the electrical resistance when the sensor experiences a mechanical strain. This effect occurs due to the change of current path in the fabric, due to the variation of the conductive contacts between the filaments inside the yarn, and due to the deformation of the fabric loops during the applied strain. Typically, these sensors are characterized by electro-dynamic tests where a pre-defined strain is applied with controlled amplitude. The electrical-resistance variation of the KPF sensors is monotonously correlated to the strain amplitude.

In order to define the electrical characteristic curve of KPF sensors, a set of data was acquired in static condition and analyzed in terms of electrical resistance, sensitivity, hysteresis effect, and repeatability [5]. Two kinds of piezoresistive sensors, called T_KPF and P_KPF sensors, were compared with the goal of demonstrating how the structure of conductive yarn influences the electrical response of the fabric sensor. The characteristic curve of the P_KPF sensor is shown in Figure 6.

The T_KPF sensor is made by using the conductive yarn after the texturization process with the aim of improving the mechanical properties. The P_KPF sensor is made with the same conductive yarn with an ordered structure in which each filament appears in a

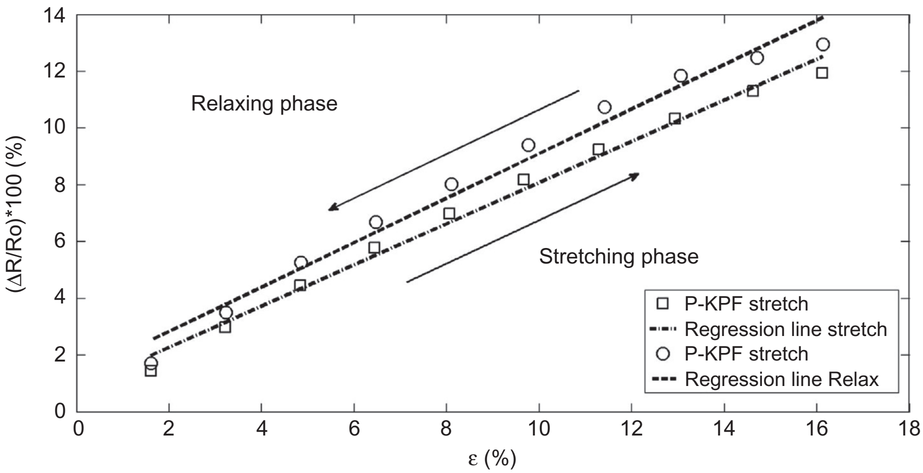


**FIGURE 6** The calibration curves of the P_KPF strain sensor; the experimental data are represented by the markers and the linear fit result is shown by the dotted lines.

parallel configuration. Figure 7 illustrates the hysteresis effects evaluated on both kinds of KPF sensors. Rectangular sensors with dimensions of 10 mm x 62 mm have been strained along the longer side to a maximum elongation of 1 mm at 0.25 Hz. The results show that the hysteresis effect is less pronounced in the P_KPF sensor. In fact, the maximum percentage of the measured hysteresis error of the P_KPF sensor is less than 10% compared with that of the T_KPF sensor, which is less than 40%.

Piezoresistive textile sensors have been used to detect the movements of the human body in a wide range of configurations, for monitoring of hand, wrist, elbow, and knee articulation [7]. To test the capability of fabric sensors to evaluate joints movements, they were compared with standard electrogoniometers during simultaneous motor acquisitions. KPF sensors have also been applied to detect the movement of the fingers in combination with other kinds of sensors with the aim of developing the first integrated prototype for the simultaneous acquisition of gesture and physiological signals as a new gestural interface [12].

A new generation of wearable goniometers is under study in the framework of the European project Interaction. The electrical response of the wearable goniometer is based on the different response in terms of piezoresistive effect of the two KPF layers during the flexion and extension phase. For example, in the flexion phase, the length of the KPF layer close to the joint is constant, while the external KPF layer elongates in response to the flexion of the joint. In such sensors the variation of electrical resistance doesn't depend on the elongation effect, but only on the angle of the joint [13].

KPF sensors can also be used to detect breathing activity, in particular the plethysmography signal, by measuring the mechanical variation of the rib cage in the inspiration and expiration phases of respiration. To verify the correlation between the change in electrical resistance of KPF sensors as a function of respiration, the electrical behavior of a KPF sensor has been compared with the signal obtained from a standard ambulatory

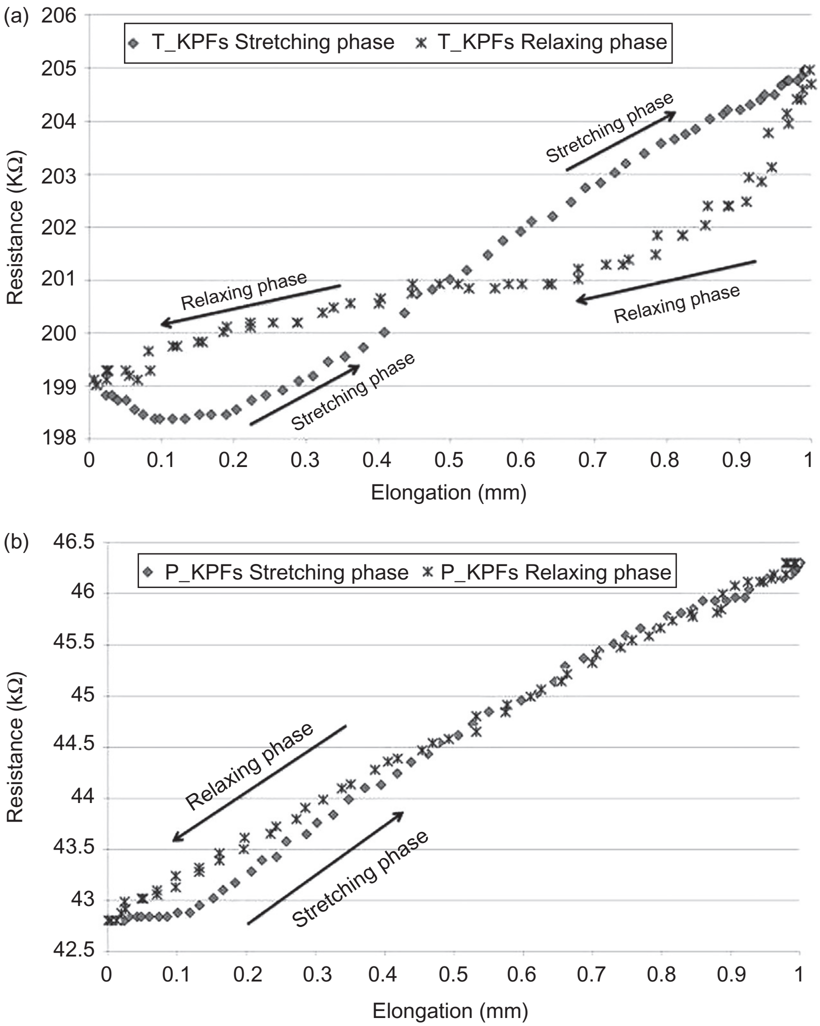


**FIGURE 7** The hysteresis behavior of the knitted piezoresistive fabric sensors: (a) T_KPF sensor and (b) P_KPF sensor.

plethysmography sensor (Biopac Lab System, Inc.) by placing both kinds of sensors in the thoracic position during normal respiratory activity. The breathing rate calculated from the breathing sensor signals produced comparable results [14].

## 5. NON-INVASIVE SWEAT MONITORING BY TEXTILE SENSORS

A sweat analysis system that can be easily integrated into fabric for real-time analysis of sweat during exercise has been developed in the framework of the Biotex project [15]. This project aimed at the development of sensing patches adapted to different targeted body fluids and biological specimens. Most of the sensors were developed for implementation in a textile material and easy integration in a patch or a garment. The remaining sensors were developed taking into account textile compatibility. Sensors based on optical, electrochemical, and electrical principles (impedance monitoring) were explored for sensing:

1. Sweat monitoring: relative quantity (i.e., perspiration rate), over salinity (i.e., conductivity), specific ions (K + , Na + , Cl-, Mg + ; Ca + ), pH, and organics
2. Infection detection through blood and body liquid monitoring for burnt persons
3. Oxygen saturation of blood for medical, sport and security applications

A number of textile sensors were distributed around the body. The sensor garment combined a multi-parametric patch for sweat analysis measures, such as pH, sodium, and conductivity with other sensing modules, including perspiration rate, ECG, respiration, and blood oximetry sensors.

A special textile-based platform with fluid-handling properties was designed and used to collect and analyze sweat samples. Sweat collection and flow were controlled by a passive textile pump based on capillary action, which has been implemented by combining hydrophilic and hydrophobic fabrics with highly absorbent material [16]. A textile channel was created using hydrophilic material, while the absorbent was placed at the end of the channel. The absorbent controls the fluid flow, drawing the sweat toward it along the length of the channel (Figure 8). In this way, a continuous flow of sweat enters the channel where it is analyzed by the sensors and then travels toward the absorbent where it is stored. A location on the lower back was chosen to collect sweat as this is an unobtrusive location for sensor placement during exercise.

The sensor garment also contained a humidity sensor and ionic concentration sensors, allowing the monitoring of the loss of liquid from subjects and to alarm in the case of dehydration. The skin is a complex structure, but for the modeling of perspiration, it can be approximated by a homogenous flat surface that continuously emits water vapor. With this assumption, Fick's first law of diffusion can be used to calculate sweat rate from the gradient of humidity measured by a pair of wearable humidity sensors located at two

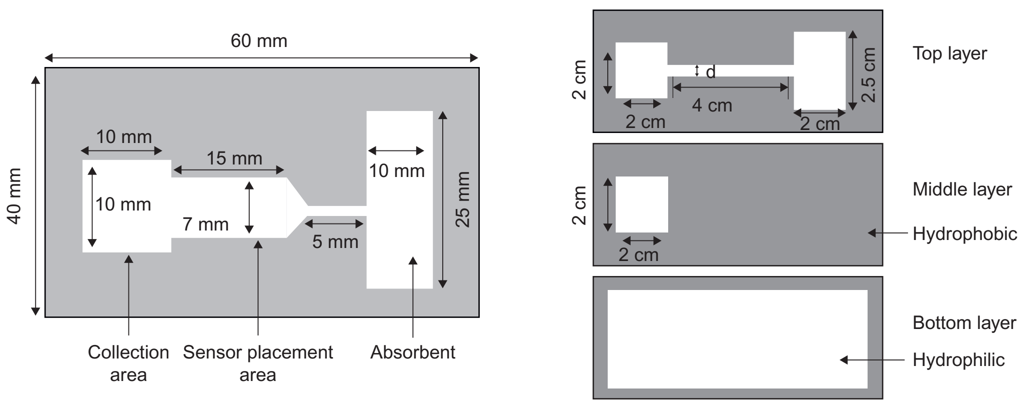


FIGURE 8 Textile passive fluid pump.

different distances from the skin. A textile humidity sensor can be implemented by sandwiching a hydrophilic insulating film between two conductive fabrics (capacitor plates). A conductive yarn (70% polyester/30% stainless steel, Bekintex) was used to manufacture the woven conductive fabric. An example of the sensor is shown in Figure 9. The capacitance change was sensed and used to derive the sweat-rate gradient over time.

The calibration curve of the textile humidity sensor is shown in Figure 9 with the corresponding curve of a commercial sensor (Philips H1). The commercial sensor is characterized by an almost linear calibration curve and a shorter response time, though the variation in capacitance is much more limited, with a dynamic range of about 30 pF. The textile sensor shows a much larger variation in capacitance, with a dynamic range that is more than one order of magnitude higher than the commercial sensor (3.5 nF), but most of this variation only occurs when the relative humidity is above 50%. The textile sensor performs reasonably well when the sensor is placed between a membrane and the skin (where high humidity was expected), but needs improvement in the low humidity region.

The sodium sensor was fabricated on a flexible kapton surface. This electrochemical sensor measures the open-circuit potential between a reference gold electrode and an ion-selective electrode (solid contact ion-selective electrode), which is a function of the sodium concentration. The sodium-selective electrode is made of a gold contact covered with a polymeric membrane that contains polypyrrole, plasticizer, ionophore, and ion exchanger. The polymeric ion-selective membrane requires a conditioning period before use (12 h in 1 mM NaCl) and must be calibrated before use using solutions with known concentrations of sodium (20, 40, 60, and 80 mM). The sensor is reusable and can be washed before reuse.

The conductivity sensor measures sweat conductivity, which is a function of the type and concentration of the ions, typically ranging from 2 to 15 mS/cm, with an average conductivity of 5 mS/cm for the sweat of healthy individuals [17]. For a single electrolyte solution, conductivity is an empirical function of concentration, often linear over a limited

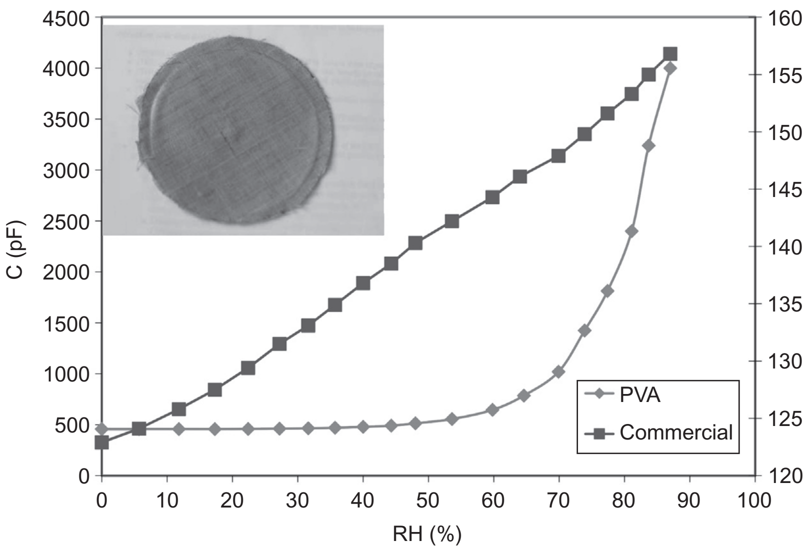


**FIGURE 9** Textile humidity sensor (upper left) and its calibration curve compared to a commercial humidity sensor [13].

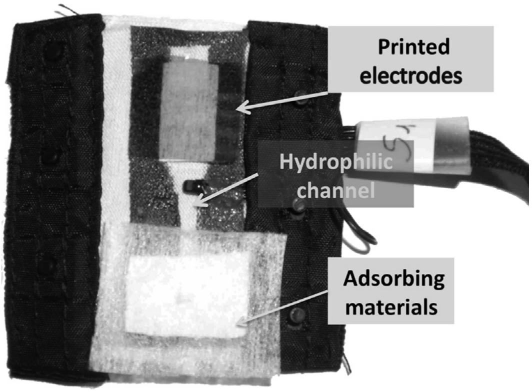


**FIGURE 10** A prototype of the passive pump and sweat sensor.

range if temperature and geometry of the electrodes are constant. In the case of a multi-component electrolyte solution, the conductivity is equal to the sum of the conductivities of its individual ions. The conductivity sensor consisted of electrodes fabricated on a flexible plastic patch that also contained the sodium sensor. As conductivity and sodium measurements are temperature dependent, a temperature sensor (Analog Device ADT7301) was included within the system to compensate for temperature changes. The Kapton patch was then placed across fluidic channel. An improved version of this sensor was implemented in the scope of Proetex project [18], where the electrodes for a sodium sensor were printed and manufactured directly on the fabric, as shown in Figure 10.

A special band worn around the abdomen was designed for the integration of the sweat sensors. Both the humidity and the ionic sensors were kept inside special pockets designed to allow easy recharge of the sensors.

## 6. SMART FABRICS AND INTERACTIVE TEXTILE PLATFORMS FOR REMOTE MONITORING

Textile sensors, such as those described earlier in this chapter, can be used for remote monitoring of physiological parameters. The first ever e-textile platform addressing clinical rehabilitation for cardiac patients was implemented in the scope of the Wealthy project [8]. The developed wearable-integrated system was able to acquire, simultaneously and in a natural environment, a set of physiological parameters such as electrocardiogram, respiratory activity, posture, temperature, and movement index. Figure 11 shows the core of the system, a knit fabric where sensors and connections are fully integrated and the conductive fibers are woven with stretchable yarns. The Wealthy sensing platform comprises: 1) six textile ECG electrodes, implemented as a double-fabric layer with the conductive part in contact with the skin; 2) three larger textile electrodes for the impedance measurement; 3) textile connections implemented as a multi-layered structure with the conductive fabric sandwiched between two insulating layers of textile; 4) two temperature transducers embedded in the garment; and 5) an integrated connector for the SpO2 sensor [19]. The fabric was made with a yarn containing stainless steel monofilament (Figure 1(a)).

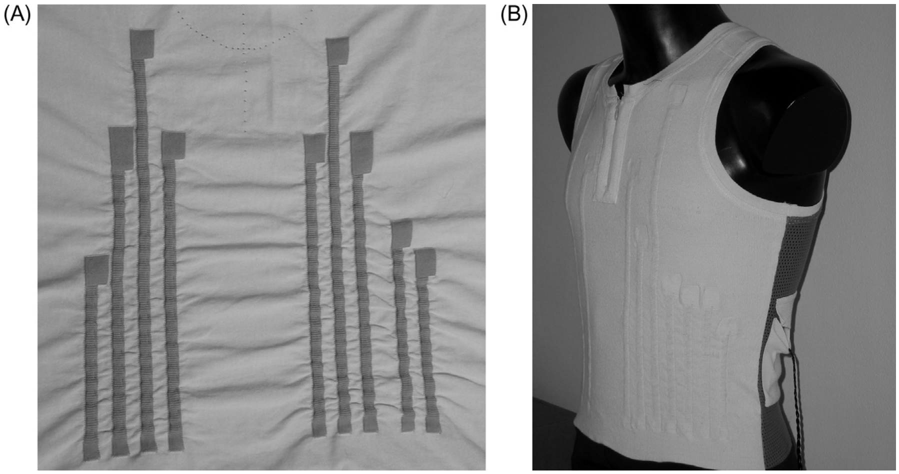


**FIGURE 11** Wealthy textile system: (a) knit fabric, the fabric connections, and the fabric electrodes; and (b) the final system.


**FIGURE 12** Typical set of signals acquired by the Wealthy platform.

Figure 3(b) illustrates particularities of the connections and the electrodes. Typical signals acquired by this system are shown in Figure 12.

Building on experience acquired within the Wealthy project and aiming at the implementation of a garment as unobtrusive as possible, a simplified platform, called the Wearable Wellness System (WWS), has been developed for long-term physiological monitoring and preventative healthcare. The WWS takes into account the limitations of wet electrodes used in the Wealthy platform that required the use of hydrogel for each measurement session. The WWS sensing shirt comprises two textile dry electrodes for ECG detection, and one textile piezoresistive sensor for detection of respiratory activity [20]. All textile sensors are fully integrated and manufactured as a one-step process. A fabric bus connects the shirt to a dedicated electronic device (developed in the frame of several projects, from MyHeart [21] to Psyche [22]) that is able to acquire and wirelessly transmit physiological signals. Posture (lying or standing) and level of activity can be monitored through a 3-D accelerometer integrated in this electronic device.

Other wearable textile platforms have been developed for applications in the field of security and protection. The latest generation of equipment and uniforms is characterized by functions like sensing, communication, and alerting. This is the paradigm of E-Sponder [23], a project that aims at the implantation of a service delivery platform based on a suite of real-time data-centric technologies and applications. The fusion of field-derived data within a central system provides information analysis, communication support, and decision support to first responders that act during crises occurring in critical infrastructure or elsewhere. Special equipment was developed to be worn by the first responders that is capable of monitoring the position and physiological parameters (ECG, breathing rate, and skin temperature) of rescuers in the field. The equipment comprises a shirt and a jacket, with sensing regions and support for embedded electronics. The garments have been designed according to the outcomes of the ergonomic trials in which rescue personnel tested the comfort and functionality of the wearable system. For the underwear shirt, the fabric and all the accessories exhibit fire-retardant characteristics. Four electrodes (two for ECG monitoring and two for respiratory monitoring by impedance measurement) are embedded in an elastic band to guarantee their adherence to the skin during action.

An instrumented jacket was produced by modifying a standard garment to hold the portable electronic devices developed during the project. A smartphone is used to receive the signals coming from the shirt and transmit them along with positioning information in real time. Special internal pockets, closed by fire-retardant zips and connected by a track of fabric integrating a flexible cable for device charging, are used to hold the smartphone and the positioning sensor. External pockets were added to hold external temperature and gas sensors.

## 7. SYSTEM FOR REMOTE REHABILITATION

Within the scope of the European project MyHeart [24], a neurological rehabilitation system (NRS) supporting stroke patients in the performance of speech and motor rehabilitation therapy was designed and manufactured. The NRS was designed for hospital and home use. When hospitalized, patients are trained to use the system. After hospital discharge, patients can use the rehabilitation device at home (or in a long-stay ward) and continue exercises with the help of a caregiver, relative, or nurse.

NRS is based on a textile-sensing platform capable of detecting body movement and a software package able to recognize if a set of movements performed during the rehabilitation exercise is correct. The system was designed to assist patients remotely during home rehabilitation or to support simultaneous sessions with several patients in a clinic. During rehabilitation, patients perform a set of exercises designed to recover functionality. Patients review a tutorial video about the exercise they are asked to perform. Next, they don the sensing garment with the help of a caregiver. After a calibration phase, the motion-recognition software starts to provide real-time feedback on the progress and accuracy of exercises by displaying symbols such as colored bars and a smiling or frowning face. The movements have to be repeated until the assigned exercise suit is performed correctly or time expires.

Motion recognition in NRS is based on printed piezoresistive fabric sensors [25]. Movement of the joints of the upper limb is monitored through 29 textile sensors placed on a shirt. The NRS garment monitors shoulder, elbow, and wrist joints, and can monitor the lateral abduction and adduction of the arm, the 90° flexion of the arm in the sagittal plane, external rotation of the arm with flexed elbow, forearm flex-extension, and prono-supination. Functions such as eating and combing were also recognized. To insulate the sensing material from the external environment and from the body, sensors were sandwiched between two layers of fabric, and shaped printed patterns were embedded in the elastic garment with the printed face in contact with the garment fabric, as shown in Figure 13.

Use of textile electrodes for EMG detection and functional electrical stimulation (FES) therapy has been explored in the Tremor project [26]. A sleeve integrating multi-electrode patches has been designed to help patients affected by movement disorders. The design of the sensing part was based on the results of ergonomic tests as well as on the functionality required to detect the presence of tremor and to control it through FES therapy. The platform combined two separate electrode patches with different distribution of the electrodes, corresponding to the different muscles of the arm [20]. The electrodes were round, 1 cm in

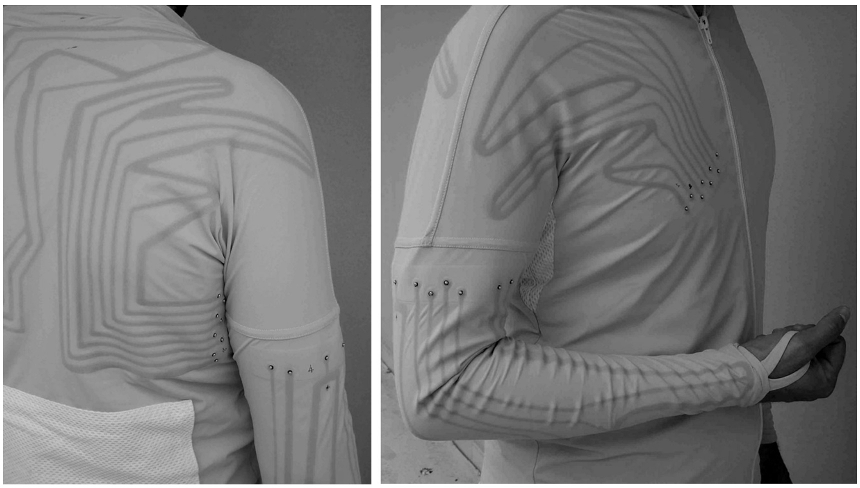

**FIGURE 13** The neuro-rehabilitation platform.

diameter and 1 cm of inter-electrode spacing. Size and shape were chosen to reduce the risks of edge effects that can increase pain during the FES.

The number of stainless steel electrodes was redundant to guarantee the maximum coverage of the region to stimulate, and it depended on the arm's size. For a size S of the upper arm, 28 sensing units were deployed in two matrices composed by four rows alternating with three and four electrodes. The larger muscle area of the lower arm was covered by 42 electrodes distributed in two matrices with six rows alternating with three and four electrodes. Three sleeve sizes were developed in order to permit the stimulation of larger arms, adding one electrode to each row to increase the sensing area. Figure 14 shows the largest sleeve size; it covers two-thirds of the proximal forearm. The chosen electrode structure allowed for easier placement of the hydrogel membranes on the electrodes. To improve coupling with the skin, use of an electrolyte in the form of hydrogel was necessary to reduce pain during FES therapy. For EMG measurements, the electrodes were moistened with a small amount of water before each measurement.

This sensing platform combines fabric electrodes and biomechanical textile sensors to perform EMG detection and recognition and to provide FES when needed [27]. The system demonstrated promising preliminary results as a possible alternative for remote rehabilitation therapy [28].

The latest generation of a wearable system for stroke rehabilitation (under development in the scope of the Interaction project) combines electronics devices (inertial sensors and portable electronics) with textile sensors (wearable goniometers and EMG sensors). The goal is development of a system able to sense the motor functions in stroke patients in a remote environment without any supervision. The wearable system is comprised of a shirt and trousers. The male and female versions of the garment have been tested on patients to evaluate their level of acceptability. The fabric that has been selected for the implementation of the garment is thermally comfortable as well as very soft and able to slip on the skin. Moreover, the use of zips and elastic strings linked to the slider facilitates the wearing of the shirts. The garments have been designed taking into account future implementation of the sensors: a more elastic and heavier fabric has been inserted in both shirt and trouser to hold the sensors and the electronics.

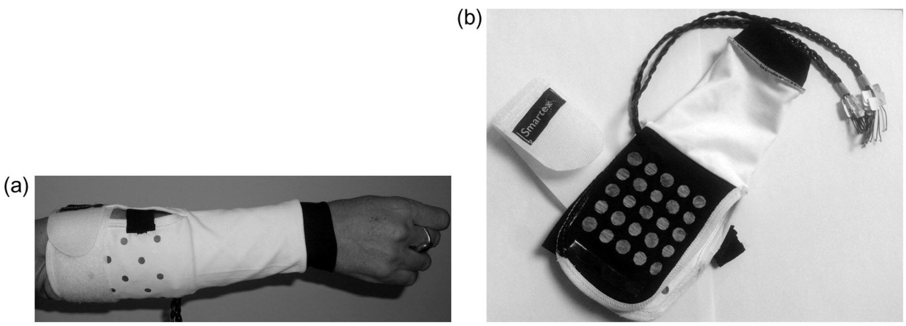


**FIGURE 14** System for the forearm designed to be tested for both EMG measurements and FES therapy: (a) outer side and (b) inner side.

# 8. SYSTEMS FOR EMOTIONAL STATE ASSESSMENT

The monitoring solutions presented in the previous sections can be used the whole day. Monitoring can be performed during the day, during the night, or during special selected time-frames. For instance, more detailed monitoring can be done in the morning and before sleep. Such monitoring can acquire data from the cardiopulmonary system as well as weight, blood pressure, and activity and information about therapy, pain, mood, and any other relevant data.

One of the more interesting applications concerns the use of smart fabric and interactive textile (SFIT) systems to push people toward a healthier lifestyle. It is important to help people acquire awareness of their overall health, to motivate them to become active in staying healthy and feeling well. SFIT technology may potentially provide innovative solutions to manage lack of sleep, stress, inactivity, metabolic syndrome, and mood disorders. The European project Psyche focused on the development of a personal, cost-effective, multi-parametric monitoring system based on textile platforms and portable sensing devices for the long-term and short-term acquisition of data from patients affected by mood disorders. The sensing platform developed during the project (Figure 15) allows identification of the triggers that precede every episode the patient has suffered.

The project relies on the previously described Wearable Wellness System (WWS) to collect physiological data that includes heart rate variability, respiratory rate, activity, and

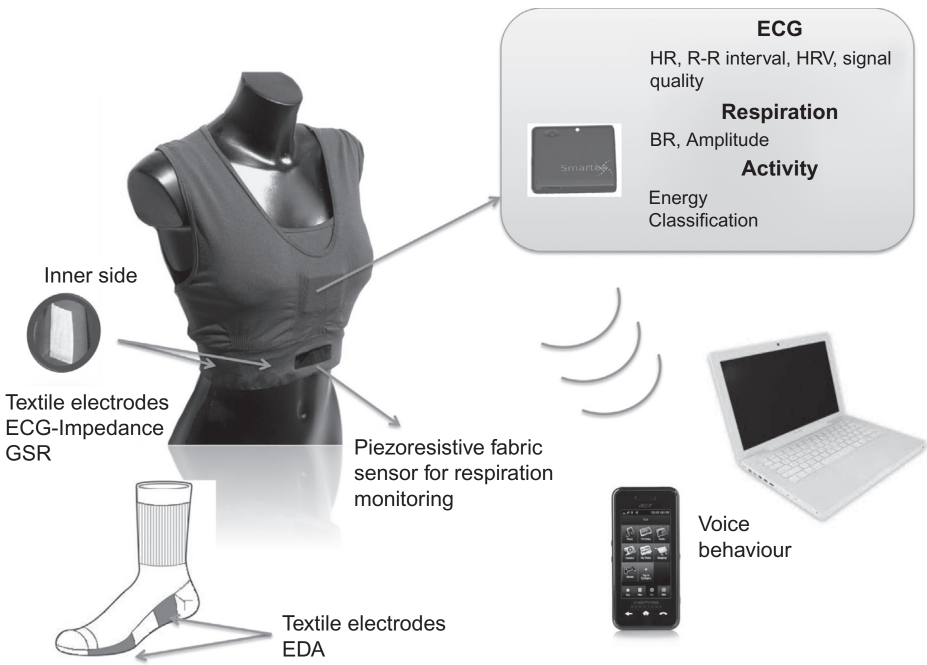


**FIGURE 15** Psyche multi-sensing platform.

movement. The WWSs have been distributed among healthy volunteers and individuals with mood disorders for the acquisition of data that will serve as reference for the database [22]. Biochemical measurements, voice analysis for emotional assessment [29], and the detection of attitudinal indicators (social interaction, daily activity, productivity, emotional perception) are also being considered as predictive metrics. The reference database also foresees inclusion of sleep-pattern data, measures of cardiovascular and respiratory functions, electro-dermal response, and stress-related hormones, including change in the diurnal variations of all these measurements [30]. The system is also able to gather subjective data through the smartphone where clinical questionnaires can be completed by patients.

The Psyche system also provides the chance for patients to establish communication with other patients in similar situations. The communication tool could be used to implement communities of users and to implement online group therapies mediated by a health professional involved in the long-term healthcare of bipolar patients. The use of the WWS sensing platform is voluntary: the smartphone allows the patient to activate and deactivate the sensors. The feedback from the patient is mediated by the physician, who can remotely assess symptoms and trends.

The final validation study of the Psyche system is ongoing in three different research centers. Fifteen complete platforms have been distributed, and for each platform a redundant number of WWSs have been manufactured to cover all sizes and models. So far more than 30 patients have been enrolled in the study. Each week the WWS is used for two nights to collect the full set of physiological parameters, while the smartphone is used during the whole week for the behavioral information. Data are automatically sent to a remote server through the smartphone. Preliminary results on data collected in the first phase of the project have already been published and are extremely encouraging [31,32].

## 9. CONCLUSIONS

Electronic textiles are a key enabling factor in the creation of SFIT-based systems, which are conceived as the integration of textile and non-textile sensors, computing capabilities, and an interactive communication network. SFIT systems rely on advances in such fields as material processing (fibers and polymers), microelectronics, signal processing, nanotechnologies, and telecommunication. Textile is the common platform where smart fibers are integrated, where the properties of the material are augmented through a combination of chemical processes, and where the structure of the fabric allows the use of redundant sensor configurations. The examples presented in this chapter illustrate applications based on textile platforms in which the fabric is enriched with new functionalities while maintaining the mechanical properties that make the material comfortable, conformable, and pleasant to the touch. The full potential of e-textile technology has not yet fully been explored and, in the future, novel flexible structures will be developed that will be sensitive to new electric, chemical, and biological variables. Within these structures properties such as toughness, elasticity, breathability, and appearance will merge with sensing capabilities. Meanwhile, the wearable e-textile platforms that have already been implemented are providing more and more data, generating information about people living normal lives. Future research will analyze these data and generate new strategies for the treatment of diseases and implement new tools for remote assistance. Knowledge in fields like medicine and physiology will also increase as well as the consciousness of people in maintaining healthy lifestyles and managing chronic diseases.

## Glossary

ECG Electrocardiograph
EMG Electromyograph
EOG Electrooculograph
EEG Electroencephalograph
G Gauge factor (G) or strain factor is the percent of change in resistance per unit strain.
SFIT Smart fabrics and interactive textile
FR First responder
KPF Knitted piezoresistive fabric
T_KPF Textured knitted piezoresistive fabric
P_KPF Parallel knitted piezoresistive fabric

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