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Electroactive fabrics and wearable man–machine interfaces

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4.1 Introduction

Multifunctional electroactive fibres and fabrics will give to the traditional textile industry a new added value represented by the possibility of making daily life healthier, safer and more comfortable. They will bring technological advances closer to the public through the realisation of easy-to-use interfaces between humans and devices. This can be achieved by combining advanced microfabrication technologies, material science, textile and electronic engineering in the production of smart clothing, realised by innovative and high knowledge-content textiles that integrate sensing, actuating, electronic and power functions.

The fabrication of such multifunctional interactive fabrics represents a potentially important method for promoting progress, sustainable development and competitiveness in several disciplines, including:

- health monitoring: detecting and preventing diseases
- rehabilitation: restoring lost functions
- health assistance: compensating for disabilities to achieve a higher quality of life
- sports medicine: assessing performance to prevent risks and improve training techniques
- telemedicine and teleoperations: supporting health professionals.

Multifunctional interactive fabrics can also be employed in emerging technology markets, such as:

- wearable wireless communication systems

- localisation and tracking of people
- ergonomics: comfort and safety
- virtual reality: simulation for professional training and entertainment.

There have been a few attempts to design and build prototypes of wearable functional devices. Most of them have taken a limited approach, consisting of attaching off-the-shelf electrical components such as microcontrollers, surface mounted light-emitting diodes (LEDs), piezoelectric transducers, and so on, to traditional clothing material, transforming the cloth into a breadboard of sorts. In fabrics containing conductive strands, these may be used to provide power to devices, as well as to facilitate communication between them. More recently, attempts to construct chip packages directly by a textile process have been reported (Post and Orth, 1997). Other research lines progressed towards the possibility of routing electrical power and communication through suspenders made of a fabric with embedded conductive strands (Gorlick, 1999).

Promising recent developments in material processing, device design and system configuration are enabling the scientific and industrial community to concentrate efforts on the realisation of smart textiles. In fact, all components of interactive electromechanical systems (sensors, actuators, electronics and power sources) can be made of polymeric materials, to be woven directly in textile structures (sensing and actuating micro-fibres) or to be printed or sewn onto fabrics (flexible electronics). In particular, intrinsic sensing, actuating, dielectric or conductive properties, elasticity, lightness, flexibility and the relatively low cost of many electroactive polymers make them potentially suitable materials for the realisation of such systems.

Accordingly, Table 4.1 presents a non-exhaustive list of different polymers used at present for sensing applications and their inorganic conventional counterparts. The list is divided into passive sensors (those that directly convert or amplify the input without a power source) and active sensors (those that require an external power source to convert or amplify the input into a usable output). These two classes of materials should be considered as complementing rather than competing with each other; however, polymers still have to be considered as materials that are at a stage of development. Most of these materials are currently not available in fibre form, and much work has to be done to reach this result.

Table 4.2 reports different polymers currently under investigation for actuating applications and their inorganic counterparts largely used in industrial applications. All of these materials are currently not available in forms compatible with textile technology and much effort is necessary before this goal can be reached.

The aim of this chapter is to give a picture of the potential use of organic materials in the realisation of sensing strain fabrics and of actuating systems. In particular, the early stage of implementation and the preliminary testing of fabric-based wearable interfaces will be illustrated with reference to wearable motion

Table 4.1 Polymers for sensor design and conventional inorganic counterparts

Physical effect	Materials	
	Polymers	Inorganics
Passive sensors		
Piezoelectricity	Polyvinylidene fluoride Polyvinylidene fluoride trifluoroethylene Polyhydroxybutyrate Liquid crystalline polymers (flexoelectricity)	Piezoelectric zirconate titanate Zinc oxide Quartz
Pyroelectricity	Polyvinylidene fluoride Ferroelectric superlattices	Triglycine sulfate Lead-based lanthanum-doped zirconate titanate Lithium tantalate
Thermoelectricity (Seebeck effect)	Nitrile-based polymers Polyphthalocyanines	$Cu_{100}/Cu_{57}Ni_{43}$ Lead telluride Bismuth selenide
Photoelectricity	Polyacetylene/n-zinc sulfide Poly(*N*-vinyl carbazole)+ merocyanine dyes Polyaniline Poly(*p*-phenylenevinylene) Polythiophene	Silicon Gallium arsenide Indium antimonide
Electrokinetic	Polyelectrolyte gel Porous ionic polymers	Sintered ionic glasses
Magnetostriction	Molecular ferromagnets	Nickel Nickel–iron alloys
Active sensors		
Piezoresistivity	Polyacetylene Pyrolysed polyacrylonitrile Polyacequinones Polyaniline Polypyrrole Polythiophene	Metals Semi-conductors
Thermoresistivity	Poly(*p*-phenylenevinylene)	Metals Metal oxides Titanate ceramics Semi-conductors
Magnetoresistivity	Polyacetylene Pyrolysed polyvinylacetate	Nickel–iron alloys Nickel–cobalt alloys
Chemoresistivity	Polypyrrole Polythiophene Ionic conducting polymers Charge transfer complexes	Palladium Metal oxides Titanates Zirconia
Photoconductivity	Copper phthalocyanines Polythiophene complexes	Intrinsic and extrinsic (doped) semi-conductors

Table 4.2 Polymers for actuator design and conventional inorganic counterparts

Physical effect	Materials	
	Polymers	Inorganics
Electronic activation		
Piezoelectricity	Polyvinylidene fluoride Polyvinylidene fluoride trifluoroethylene Polyhydroxybutyrate Liquid crystalline polymers (flexoelectricity)	Piezoelectric zirconate titanate Zinc oxide Quartz
Electrostriction	Dielectric elastomers (acrylic or silicone rubbers) Polyvinylidene fluoride Polyvinylidene fluoride trifluoroethylene copolymers Polyvinylidene fluoride hexafluoropropylene copolymers	Barium and lead titanate single crystals Polycrystalline $BaTiO_3$
Electrostatics	Dielectric elastomers	Silicon
Ionic activation		
Electromechano-chemical	Polypyrrole Polyaniline Polyelectrolyte gels Polymer–metal composites (IPMC) Carbon nanotubes	–

capture systems and a functionalised shirt capable of recording several human vital signs. Although the realisation of a wearable kinaesthetic interface is one of our main aims, it can appear somewhat futuristic. Nevertheless, we have focused our efforts on this application and progressed towards preliminary prototypes.

4.2 Sensing fabrics

Different sensing strategies based on piezoresistive materials are under study and development to realise sensors that can be integrated into elastic fabrics.

4.2.1 Materials and fabric preparation

Different fabrication methods have been used to confer piezoresistive properties to garments. The first approach involves coating conventional fabrics with a thin layer of polypyrrole (PPy, a Π-electron conjugated conducting polymer). PPy is a conducting polymer that combines the good properties of elasticity with mechanical and thermal transduction. PPy-coated Lycra/cotton fabrics that work as strain sensors are prepared using the method reported by Della Santa *et al.* (1999).

Another technique is based on the coating of yarns and fabrics with a mixture of rubber and carbon. The treatment is realised by immersing the material in a solution of rubber and microdispersed phases of carbon. After removal of the excess materials, the conductive elements are immobilised in the structure through a treatment at a temperature of 130°C. The mechanical properties of the final product are affected by the speed of the coating process, the viscosity of the solution and the mutual permeability of materials. Sensors based on carbon loaded rubbers (CLR) realised in this way work as strain sensors.

4.2.2 Characterisation of sensors

The PPy and CLR sensors were characterised in terms of quasistatic and dynamic electromechanical transduction (piezoresistive) properties. Thermal and ageing properties of the sensing fabrics were also preliminarily assessed.

Sensor electromechanical characterisation was performed by exerting uniaxial stretching through rigid links connected to a DC motor and by reading the corresponding variation in electrical resistance. The motor was driven and controlled by an encoder connected to a PC. Quasistatic characterisation was executed by applying small increments of uniaxial stretching, while dynamic characterisation was performed with stepwise stretching. In order to estimate the bandwidth of the sensors, sinusoidal mechanical stimulation at increasing frequencies was applied and the frequency components of the response were investigated.

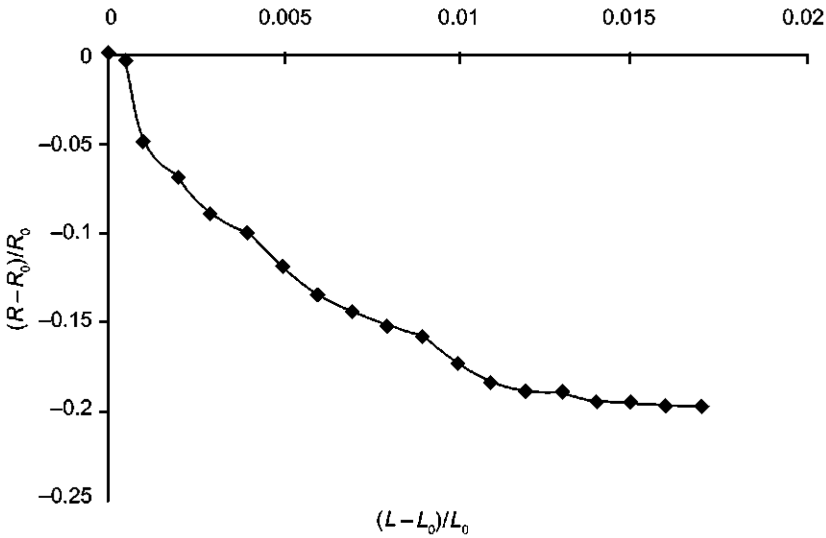
A simple thermal characterisation procedure was performed to determine how temperature influences the piezoresistive properties of the sensors. The electrical resistance of samples at different temperatures was measured by putting them into an electronically controlled thermostatic cell. To evaluate the ageing behaviour of sensors, the time dependence of the electrical resistance of unstrained threads was evaluated by daily measurements with a digital tester, over a period of one month.

4.2.3 Results and data analysis

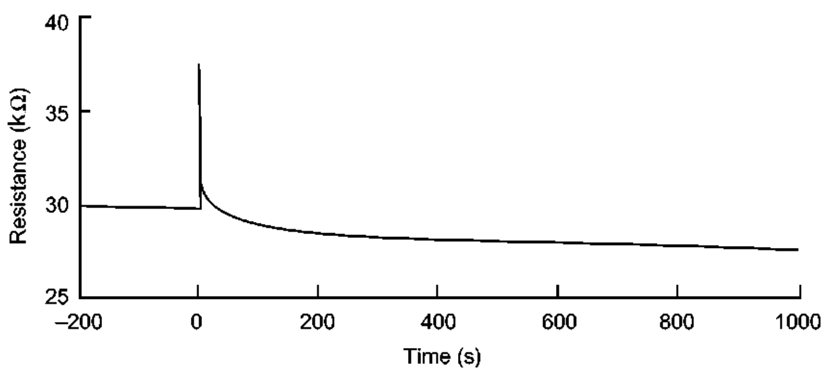
PPy-coated Lycra/cotton fabrics

The quasistatic characterisation on PPy-coated fabrics indicates an average gauge factor (GF = $(R-R_0)L_0/(R_0(L-L_0))$, where R and L are the sensor resistance and length, respectively, while R_0 and L_0 are their rest values) of about –13 (negative and similar to that shown by nickel). The numerical value of GF was calculated from a linear interpolation of the data (before saturation) reported in Fig. 4.1.

Despite the fact that the high GF value is suitable for strain gauge implementation, two serious problems affect PPy-coated fabric sensors. The first problem resides in the strong variation with time of the sensor resistance. The second



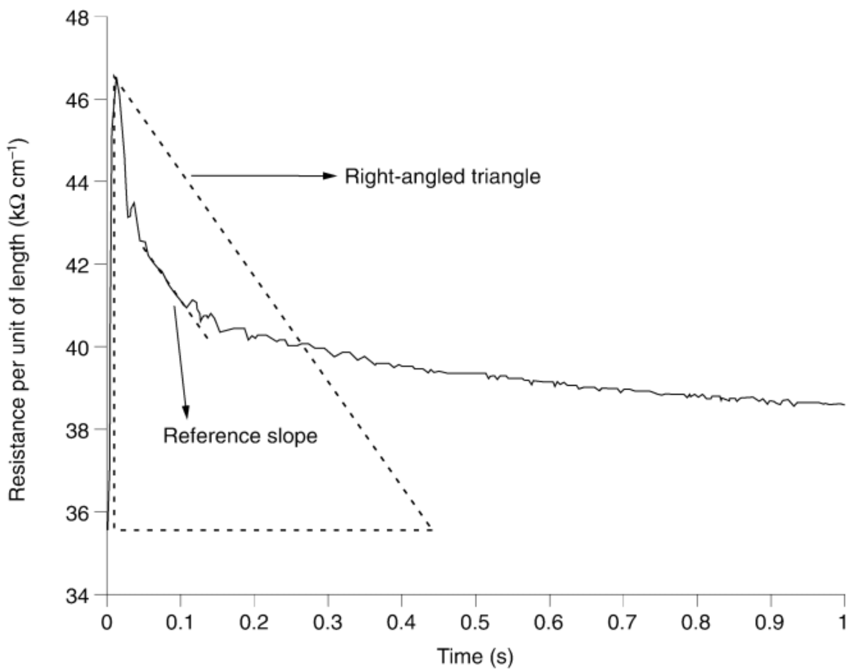
4.1 Typical quasistatic response in terms of percent change in electrical resistance versus uniaxial strain for a PPy-based sensor.



4.2 Response in terms of change in electrical resistance for a PPy-based sensor under a stepwise stretching (t = 0s).

problem is the high response time of the sensors; in fact, after the sudden application of a mechanical stimulus the resistance will reach a steady state in a few minutes (see Fig. 4.2); this could restrict the fields of application.

Nevertheless, both limitations have been partially overcome by the following 'ad hoc' coding procedure. Analysing the resistance response in the range of 1 s after the imposition of a stepwise deformation, it is possible to derive the applied strain in an ageing invariant way. We consider a right-angled triangle (Fig. 4.3) where the cathetus height is equal to the excursion of the response peak and the



4.3 PPy sensor resistance per unit of length versus time after the application of a stepwise stretching (t = 0s). The triangle considered to be useful in data treatment is also traced. The reference slope is given by the slope at the middle point between peak and final value.

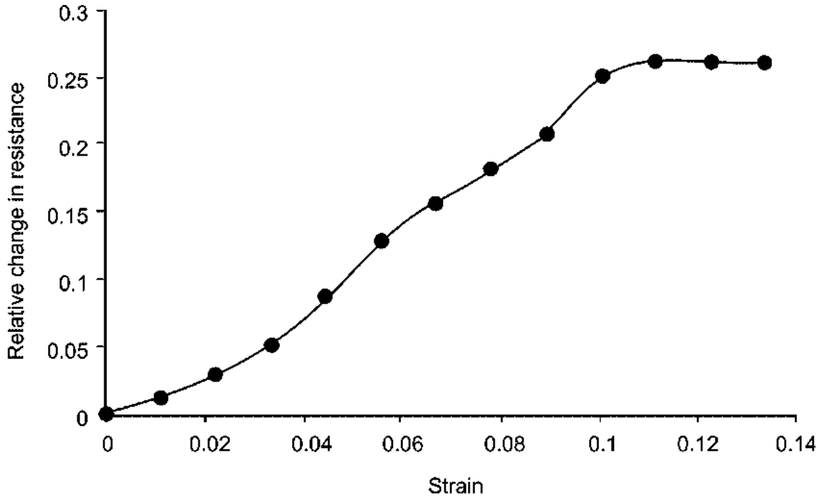
slope of the hypotenuse is equal to the time derivative of the resistance, calculated at the middle point between the peak and the final value of the range. It has been demonstrated (Scilingo *et al.*, 2003) that the area of this triangle codifies for the strain independently of the sensor resistance ageing.

PPy-coated fabrics showed a temperature coefficient of resistance (TCR) of about 0.018°C^{-1}.

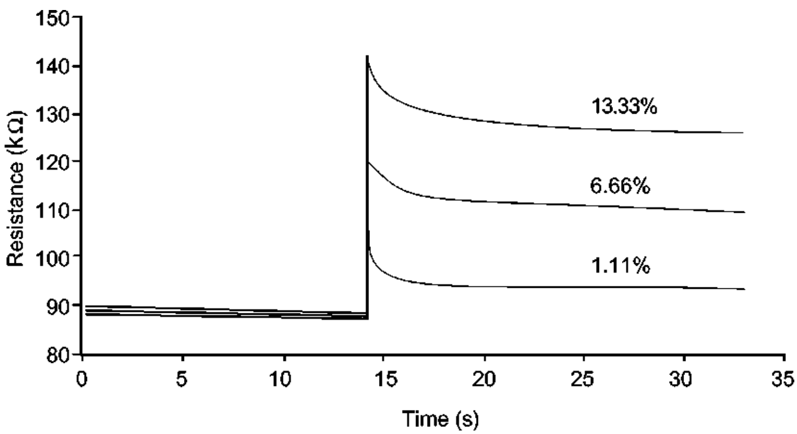
Lycra/cotton fabrics coated by carbon-loaded rubber

A GF of about 2.5 was measured for CLR-coated fabrics. The numerical value of GF was calculated from a linear interpolation of the data (before saturation) reported in Fig. 4.4. These values are quite similar to those of metals and are suitable for such sensors to be used in wearable applications.

The behaviour of the fabric sensor when subject to stepwise stimulations at increasing amplitudes was investigated. Different levels of strain were tested (ranging from 1.1 to 13.3%, with incremental variations of 1%). The resistance versus time for three step strains in stretching is reported in Fig. 4.5.



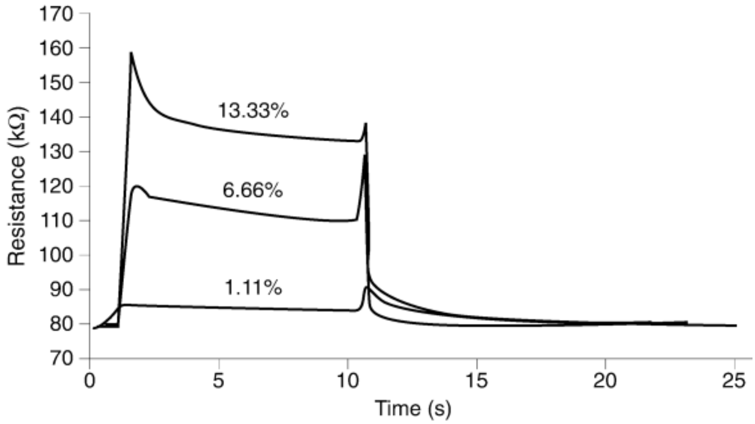
4.4 A typical quasistatic response in terms of percentage change in electrical resistance versus strain for a CLR-based sensor.



4.5 Resistance response to step strains (1.11, 6.66 and 13.33%) in stretching.

These fabric sensors work well in the range of 1 to 13%. Aside from this range, the response is not univocal and the sensor cannot be used for monitoring kinematic variables. A dual behaviour can be verified with step strains in shortening (see Fig. 4.6).

In this case the fabric was previously stretched to a certain strain, held at that strain for a few seconds in order to attain the steady value of resistance, and then

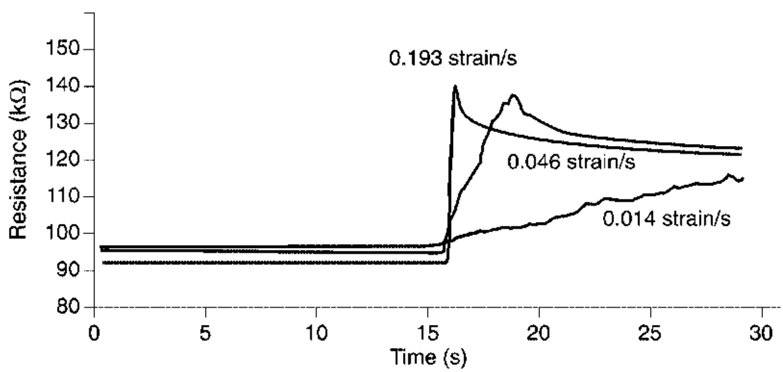


4.6 Resistance response to different step strains (1.11, 6.66 and 13.33%) in shortening.

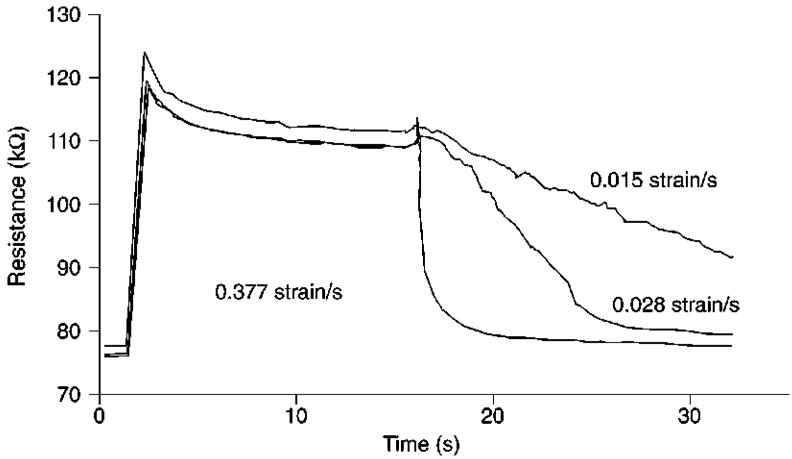
suddenly relaxed. To establish the sensitivity of the fabric sensor to the imposed strain rates, a set of ramps in stretching as well as in shortening with increasing slope was applied. The results are reported in Figs 4.7 and 4.8. It can be observed that the resistance of the fabric follows the time dependence of the applied strain. Tests on sensor bandwidth showed that once the frequency exceeds 8 Hz, the resistance response is nearly flat. CLR-coated fabrics showed a TCR of about 0.08°C^{-1}.

4.3 Actuating fabrics

Electroactive polymer actuators are being studied and developed to be embedded into fabrics, to endow them with motorized functions. Three kinds of electroactive materials (dielectric elastomers, conducting polymers and carbon nanotubes) are under investigation in our laboratory.



4.7 Resistance response to ramps with increasing slopes in stretching (strain 8.89%).



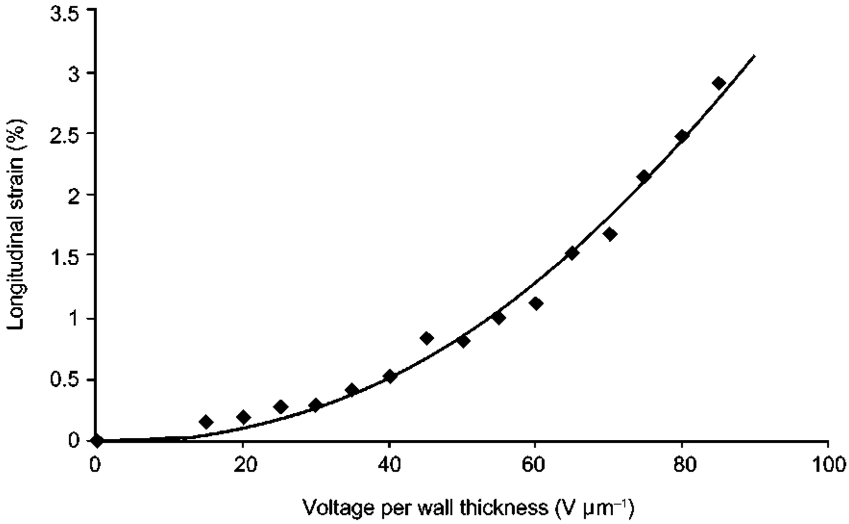
4.8 Resistance response to ramps with different slopes in shortening (strain 8.89%).

4.3.1 Dielectric elastomer wearable actuators

The realisation of actuating devices with fibre geometry implies the need to overcome several difficulties, such as the identification of efficient principles of operation and suitable configurations, the selection of high-performance materials and implementation of custom fabrication processes. Silicone rubbers are being tested as dielectric elastomers to realise high-strain wearable actuators. Dielectric elastomers possess several advantages: linear actuation strains of up to 60%, fast response times (down to tens of milliseconds) and generated stresses of the order of MPa (Pelrine *et al.*, 2000). The price for achieving such performances is represented by the very high driving electric fields (order of 100 V μm^{-1}).

A prototype actuator was realised, made of a silicone hollow cylinder with two compliant electrodes (carbon conductive grease) applied to the internal and external surfaces. Imposing a voltage difference between the electrodes, the polymer sustains an electric field-induced deformation. Preliminary tests showed a longitudinal strain of 3% with a voltage per wall thickness of 85 V μm^{-1} (see Fig. 4.9). The need for high driving electric fields is the actual limitation to the unconditioned use of these actuators.

To improve the performance of actuators made of dielectric elastomers, the possibility of implementing a mechanical amplification of strain is under evaluation. This solution would reduce operation voltages. The system under study is thought to be made of a bundle of cylindrical actuating units, covered by a cylindrical braid mesh (made with flexible but not extensible threads) able to contract when the internal elements impose a radial expansion. The system



4.9 Longitudinal strain versus applied voltage per wall thickness for a silicone cylindrical actuator.

changes shape, increasing its diameter, decreasing its length and changing the angle α between the axes of the cylinder and the threads.

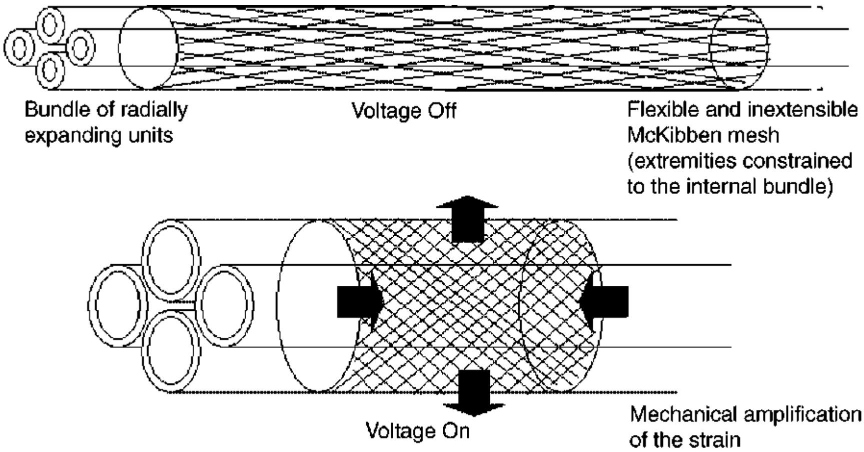
Two actuating configurations are under study. The first is based on the 'direct' McKibben effect (Chou and Hannaford, 1996). Under electrical stimulation each internal unit undergoes a radial expansion and the mesh produces a strain amplification from the radial direction to the longitudinal one, with tangible shortening and useful contractile forces (see Fig. 4.10).

The amplification factor depends on the mesh structure and its work conditions. It has been shown (De Rossi *et al.*, 2001) that, if the initial value of α is larger than $\pi/4$, the radial expansion is transduced into a linear contraction with an amplification factor larger than 1, as expressed by the relation:

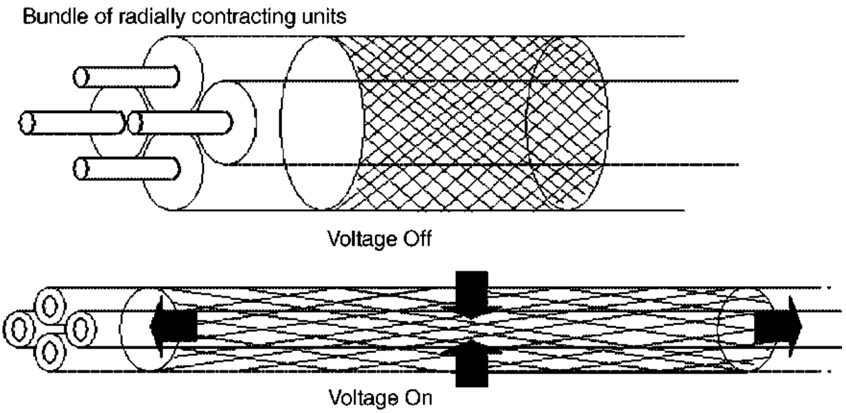
$$\left(\frac{\Delta L_m}{L_m}\right) \approx -tg^2(\alpha)\left(\frac{\Delta R_m}{R_m}\right) \quad \text{for} \quad \left(\frac{\Delta R_m}{R_m}\right) << 1,$$

where R_m and L_m are the radius and the length of the mesh, respectively.

The second configuration is based on the 'inverse' McKibben effect. Using radially contracting units and realising a bundle of them with a total diameter higher than the resting mesh diameter, each unit undergoes a radial contraction under electrical stimulation. The mesh produces a strain amplification from the radial direction to the longitudinal one, with lengthening and expanding forces (see Fig. 4.11).



4.10 First actuating configuration: 'direct' McKibben effect.



4.11 Second actuating configuration: 'inverse' McKibben effect.

4.3.2 Conducting polymer fibre actuators

Conducting polymer (polyaniline) fibres have been made and tested as actuators (Mazzoldi *et al.*, 2000). They exhibit sizeable active strains (of the order of 1% and more), large active stresses (up to tens of MPa), low driving electrical potential differences (a few Volts) and built-in tunable compliance. Continuum and lumped parameter models for such actuators have been formulated and validated (Mazzoldi *et al.*, 2000), providing a necessary tool for implementing biomimetic control strategies and algorithms. However, at present the use of such actuators is limited by the high value of their response time constants and their short lifetime, both factors being determined by the need for an electrochemical driving force. Attempts are underway to find suitable conductive polymer fibre actuators in order to overcome these limits and to integrate the actuators into active fabrics.

4.3.3 Carbon nanotube fibre actuators

Carbon nanotube fibres have been made and preliminarily characterised as actuators. Carbon nanotubes (the most recent addition to the class of electroactive materials suitable for actuation purposes (Baughman *et al.*, 1999)) are sheets of carbon atoms rolled up into tubes with diameters of around tens of nanometres. Their projected superior mechanical and electrical properties (high actuating stresses, low driving voltages and high energy densities) suggest that superior actuating performances can be expected (Baughman *et al.*, 1999). However, at present the preparation of carbon nanotube fibres has to be much improved to produce fibres able to show all their actuating potential.

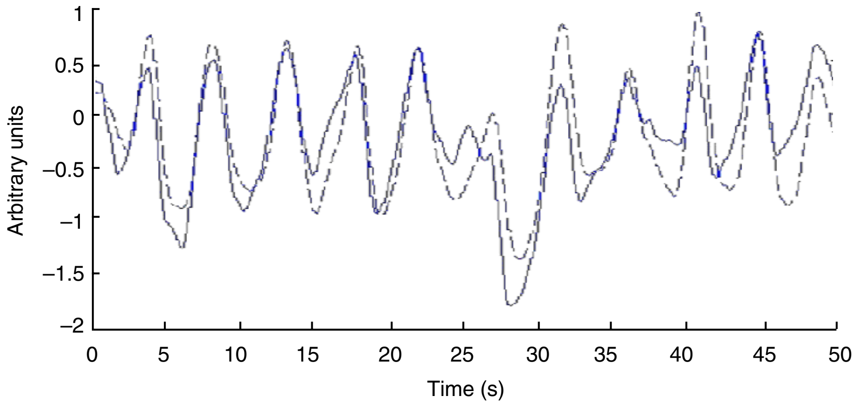
4.4 Smart fabrics for health care

An emerging concept of health care, the continuous monitoring of vital signs to provide assistance to patients, is gaining wide acceptance. Wearable non-invasive sensing systems will allow the user to perform everyday activities with minimal training and discomfort.

The development of an integrated sensorised shirt able to record vital signs was described in a previous work (Pacelli *et al.*, 2001). This truly wearable system is conceived to provide continuous remote monitoring of the health status of the patient. The simultaneous recording of vital signs would allow parameter extrapolation and inter-signal elaboration, contributing to the generation of alert messages and synoptic patient tables. Vital signs like electrocardiograms (ECGs) and electromyograms (EMGs) were detected by conductive fabrics, made of steel threads wound round acrylic yarns, while respitrace was acquired (for breathing rate monitoring) by using CLR fabric strips positioned around the trunk, one located at the abdominal level and the other one at the thoracic level. The response of the fabrics was compared with that of commonly used piezoelectric sensors and results were very satisfactory (Fig. 4.12).

4.5 Smart fabrics for motion capture

Truly wearable instrumented garments, capable of recording body kinematic maps with no discomfort for the subject and negligible motion artefacts caused by sensor–body mechanical mismatch, are crucial for several fields of application. The wearable devices described here meet the requirements of comfort and accuracy of motion capture systems. In most applications nowadays the bottleneck is given by devices too cumbersome and invasive for the subject, hence a well-



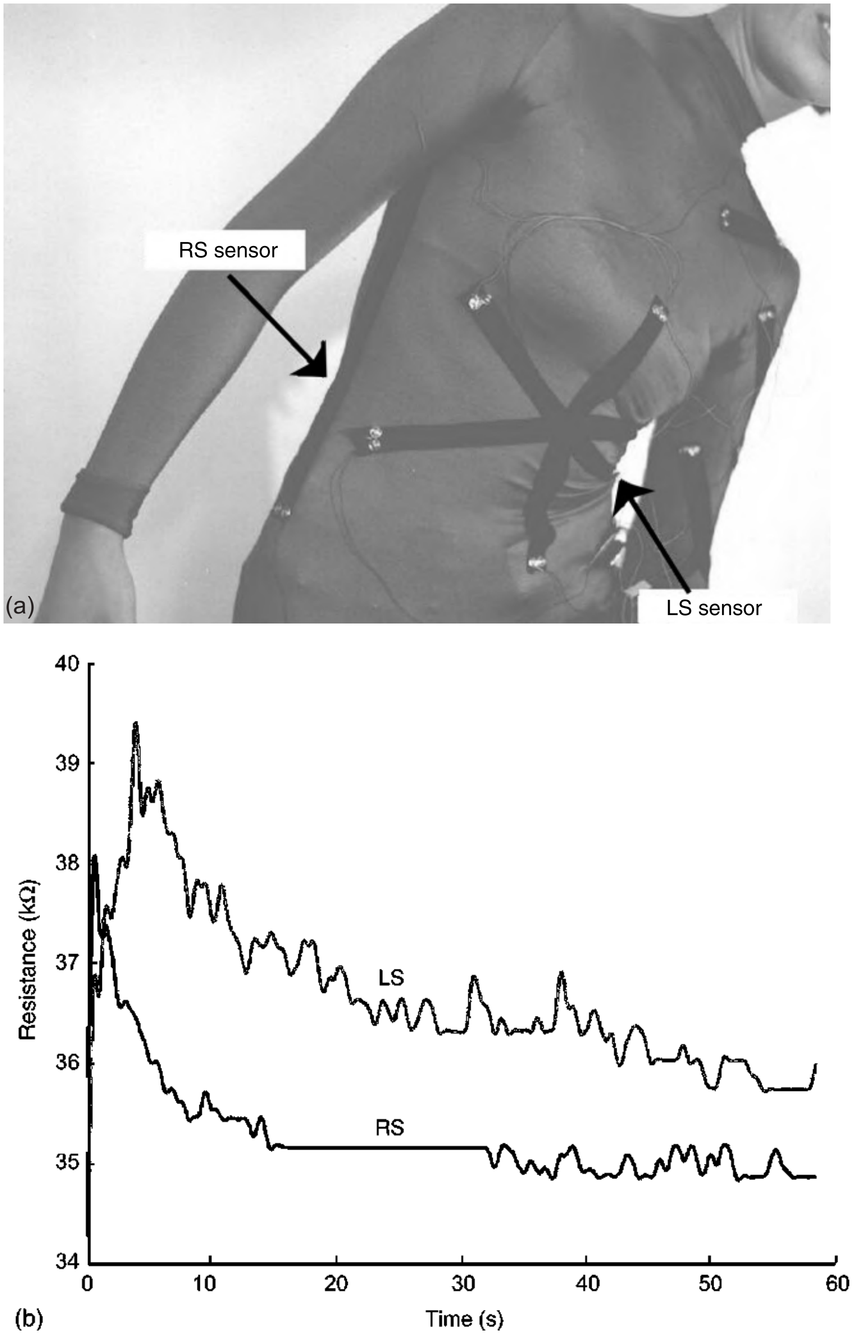
4.12 Respitrace recorded by using the CLR fabric sensor compared with respitrace obtained by a standard piezoelectric sensor (dashed line) currently in use in clinics.

fitting functionalised garment would provide strong advantages. In this context a sensorised leotard and a sensorised glove were fabricated. They are able to permit movements of the arms and upper trunk, and movements of the fingers, respectively. Fields of application range from rehabilitation to sports, from virtual reality to multimedia. The sensing properties of the garment are due to the CLR strain sensors previously described.

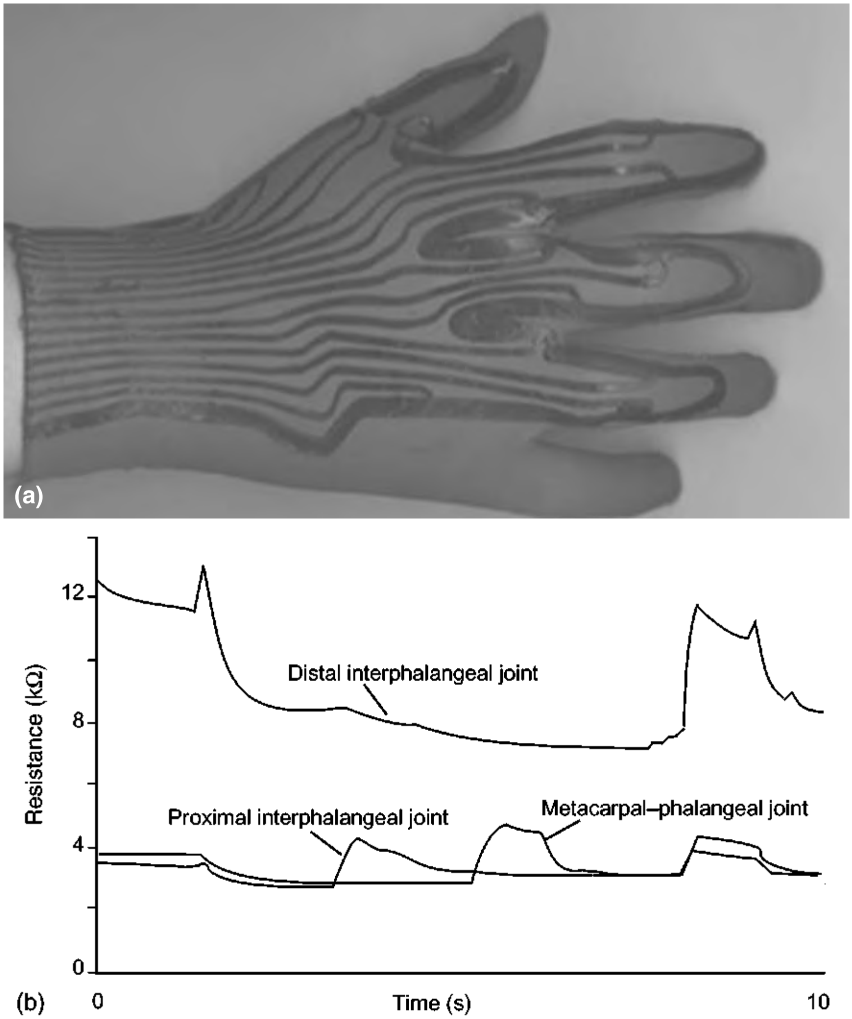
4.5.1 Detection of body movements

A few prototypes already realised (De Rossi *et al.*, 2003) have shown a reasonable ability to detect and monitor the position of body segments by reading the mutual angles between the bones. In Fig. 4.13(a) a left flexion of the thorax of a subject wearing a sensorised leotard is shown. Signals detected by sensors placed on the left side (LS) and right side (RS) of the leotard (see Fig. 4.13(a)) are reported in Fig. 4.13(b).

A prototype of a sensorised glove is presented in Fig. 4.14(a) (sensors are visible as black regions on the glove). Each interphalangeal joint is covered by a sensor, while at least two sensors are necessary to detect the position of each metacarpal–phalangeal joint and the carpal–metacarpal joint of the thumb. One degree of freedom was attributed to each interphalangeal joint, two to each metacarpal–phalangeal joint and two degrees of freedom to the trapezium–metacarpal joint. Moreover, relative movements between metacarpal bones have been considered (see the sensor disposition on the glove in Fig. 4.14(a)). The sensors modify their resistance to correspond to the flexions or extensions of each finger. Figure 4.14(b) shows three signals detected by three sensors placed



4.13 (a) Left flexion of the thorax of a subject wearing a sensorised leotard and (b) related time change in resistance of the fabric sensors located on the left side (LS) and right side (RS) of the leotard.



4.14 (a) Sensorised glove. (b) Time changes in resistance of the fabric sensors located on three joints (distal interphalangeal, proximal interphalangeal and metacarpal–phalangeal joint) of the index finger, caused by movements of the finger.

in correspondence to the distal interphalangeal joint, the proximal interphalangeal joint and the metacarpal–phalangeal joint of the index finger.

In the early prototypes sensors had been intuitively located to correspond to each joint in a number equal to the degrees of freedom. In the new generation of prototypes, a large set of sensors is distributed over the garment and a particular strategy of identification and inversion is adopted, as will be discussed in Section 4.6.

4.5.2 Multimedia applications

An application of sensorised fabrics for motion capture is represented by the possibility of triggering music and images by body movements. Strips of sensing fabrics were positioned at joints of the elbow, wrist and shoulder, and a multimedia event was associated with each sensor. The signal gathered from each sensor exhibits an initial peak when it is rapidly stretched. Therefore, its behaviour has to be interpreted and, if necessary, conditioned. As mentioned previously, a preliminary study has confirmed that, when the fabric sensor is submitted to stretching and shortening, its resistance changes. If the mechanical stimulus remains constant, the signal settles at a steady value. This behaviour is suitable for a quantification of the signal. By exploiting all of the dynamic range of variations, a threshold was fixed at the middle point. When this threshold is crossed, a given event starts. Let us focus, for example, on the sensor placed on the elbow. When the arm is completely bent, the sensor is entirely stretched. Corresponding to this position the maximum value of resistance is read. When the arm is stretched, the sensor is at maximum shortening conditions corresponding to the lowest value of resistance. Signals were acquired by a microcontroller-based electronic card via a serial port. During the movement of the arm, images and sounds were switched. The switching frequency depended on both the bandwidth of the sensor signal and the communication speed. Figure 4.15 shows a use of a sensorised glove as a man–computer interface.

4.15 Use of the sensorised glove as a human interface device.

Another application concerns the world of video games and interfaces (mouse) to control computers. In this case a sensorised glove was fabricated, where each interphalangeal and metacarpal–phalangeal joint was covered by a sensor. Starting from the flat hand position, each sensor modified its resistance in response to a movement of the finger or thumb. To read adduction–abduction of the finger, sensors were placed on the side of the fingers, in correspondence to the interdigital webs. Thumb opposition was detected by reading the output of a sensor crossing the carpal–metacarpal joint on the radial side. For joystick application, four sensors were placed on the index finger, working with one coupled on the first interdigital space and on the opposite side, and the other on the metacarpal–phalangeal volar and dorsal aspect of the index. This geometry was adopted to provide a differential reading of the sensors. Signals were acquired by the joystick port of a PC. This port was designed as an interface with two analogue joysticks. Each joystick had two buttons. The connector of the port enabled the two joysticks to be controlled at the same time. The stick was attached to two 100 kΩ potentiometers. One of the resistors changes its value with a change in the position of the stick along the x axis. The other potentiometer does the same with the y axis. By connecting the fabric sensors to pins relative to the movements along the x and y axes it was possible, through finger movements, to control the mouse pointer on the screen.

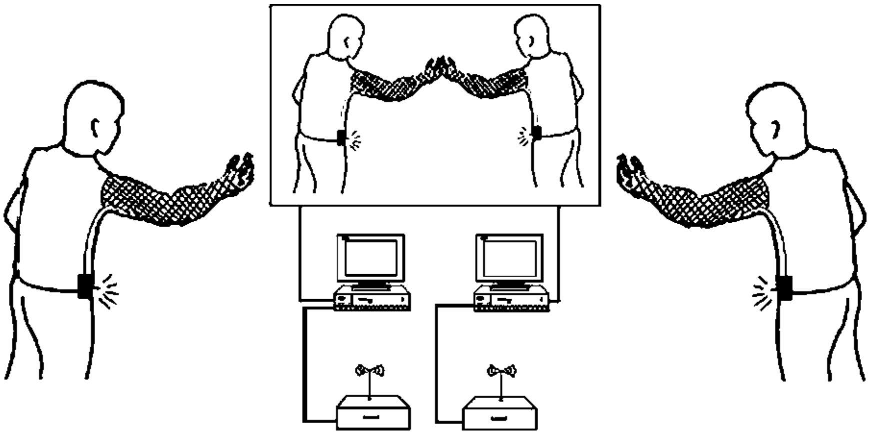
A prototype system for interpreting and translating American Sign Language was also developed. Basically, different static configurations of fingers are correlated to specific phrases of language. By means of a database it is possible to establish a small dictionary. In this case sensors are also positioned on the thumb, to allow the recognition of the alphabetical letters.

4.6 Smart textiles as kinaesthetic interfaces

The long-term goal of our research is to develop a family of wearable, bidirectional (sensing and display) man–machine interfaces to be used in surgery and rehabilitation (see Fig. 4.16). To achieve this distant goal, several methodologies and techniques need to be developed in terms of sensing (kinaesthetic), actuation and control. Figure 4.16 refers to a scheme of telerehabilitation, where a bilateral active interface is worn by the patient and is telemetrically controlled and monitored by a medical specialist from a remote position.

Telesurgery is another domain of interest, where an active interface could be worn by the surgeon in a master–slave system to provide better manoeuvrability, dexterity and ergonomic coupling: the surgeon could manoeuvre the system as if he were directly manipulating the remote object itself. Finally, a bilateral interface could be used as a wearable active orthoses for a paralysed arm. In this case, the impaired subject could perform the physical therapy by him- or herself, and the interface could also provide assistance in arm movements.

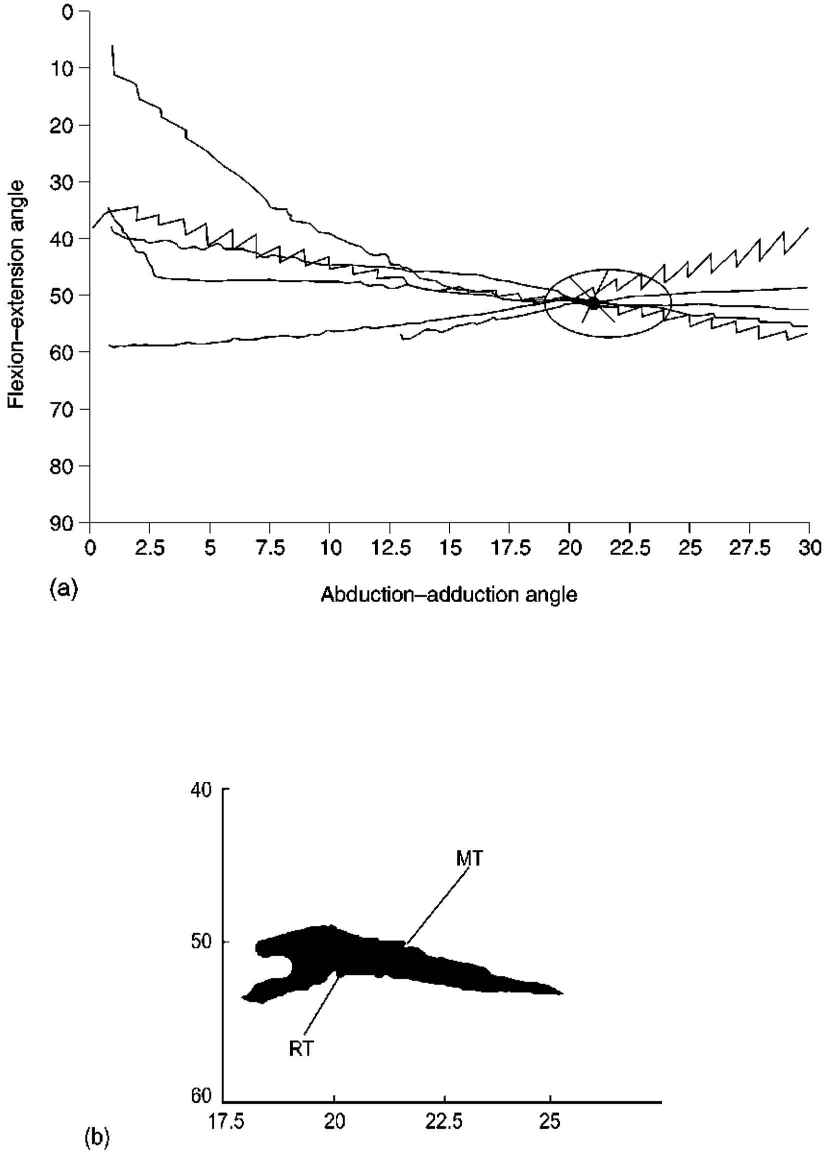
A reliable kinaesthetic interface should include a system for identifying every



4.16 Scheme of telerehabilitation.

single posture held by the subject wearing the interface itself (De Rossi *et al.*, 2003). With regard to this, a particular strategy of identification and reconstruction ('inversion') was developed, based on a redundant allocation of sensors on the overall interface and an opportune management of the information collected by them, as described below. The adopted technique is, in a certain sense, functional: the final aim is to know which gesture the subject holds and not which individual sensor has modified its status. The technique is based on a calibration phase of the entire system, regardless the single sensor. In fact, when the subject wears the interface for the first time, the information collected by the overall sensor net, following a definite set of known movements, is identified, although the location of the applied deformation is ignored. Data registered during this calibration are then interpolated by a piecewise linear function. From this point of view, redundant sets of sensing fabric patches, linked in different topological networks, can be regarded as a spatially distributed sensing field. Each possible posture and gesture can be reconstructed by simultaneously comparing the present sensing field with the values of the joint variables registered in the calibration phase. As an example, in Fig. 4.17(a) the reconstruction of a target related to the position of an arm is reported.

The abduction–adduction angle and flexion–extension angle of the glenohumeral joint are reported in the abscissa and the ordinate, respectively. Each piecewise line is the solution of the equations holding for significant sensors (responses from sensors which are not influential for the detection of this particular posture degenerate into the entire plane), projected from the entire space of the joint variables onto the plane of the coordinates of the shoulder. The estimated position is given by the intersection of these solutions. In Fig. 4.17(b) the intersection zone is enlarged. Owing to the redundant allocation of sensors, the solution was calculated in the 'least square sense', i.e. by considering the entire



4.17 (a) Traces of the values held by different sensors corresponding to a fixed position. The estimated position is given by the intersection of the traces. (b) High-density zone of probability for the estimated position. The measured target (MT) position differs from the real target position (RT) by less than 4%.

zone with a high density of possible solutions. The position of the target was calculated by the average value of all of the points contained on this zone. The distance between the calculated position of the target and the real one is about 10 cm in a range of about 2.5 m, an error of less than 4%. In other cases larger errors, always less than 8%, were obtained.

4.7 Conclusions

The advanced state of sensing technology can, at present, be considered a realistic scenario for the realisation, in a few years, of truly ready-to-use wearable systems. Much more work has to be done before efficient, reliable and small-size actuators that perform the integration of actuating functions into wearable interactive interfaces can be realised.

4.8 Acknowledgements

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