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# Electrically responsive polymers as materials for “artificial muscles”

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## 1. Introduction

Despite the advanced state of thermochemical, electromagnetic and pneumatic conventional motor technologies, there is a growing demand in fields such as mechatronics, robotics and bioengineering for polymer actuators with high power-to-weight ratio, high efficiency and large degree of compliance. These devices would provide a major advance in several applications, involving critical and delicate tasks; in particular, such kind of actuators could be advantageously used as “*artificial muscles*”.

Aimed at the satisfaction of these and similar needs, in recent years new types of actuators have been proposed. They are based on polymer materials able to change dimensions and/or shapes in response to a specific external stimulus (thermal, chemical, electrical, magnetic, electro-chemical, electro-magnetic, optical). A relevant class of such materials is represented by the so-called ElectroActive Polymers (EAP) [1]. These polymers exhibit interesting properties, such as sizable active strains and/or stresses in response to an electrical stimulus, low specific gravity, high grade of processability and down-scalability and, in most cases, low costs.

EAP materials can be classified in two major categories:

- 1) Ionic EAP (activated by an electrically-induced transport of ions or molecules):
  - polyelectrolyte gels [2]
  - ion polymer metal composites [3];
  - conducting polymers [4]
  - carbon nanotubes (they are currently classified as EAP, even though they are just non-polymeric macromolecular materials) [5, 6]

Such kind of materials can be activated advantageously by very low voltages (1÷5V), but can be operated only within a surrounding electrolyte medium.

- 2) Electronic EAP (activated by an external electric field and by Coulombian forces):

- piezoelectric polymers [7];
- electrostrictive polymers [8, 9]
- dielectric elastomers [10, 11].

These polymers require typically high driving electric field (order of 10-100 V/ $\mu\text{m}$  for electrostrictive polymers and dielectric elastomers), even though progresses concerning the reduction of the driving field are currently occurring, as it will be shown in Section 3. Furthermore, electronic EAP share interesting actuating performances, in terms of active strains and/or stresses, efficiency, response times, stability, reliability and durability.

Our research on EAP and on EAP-based systems is presently moving towards three main directions: the first is material-oriented, in order to develop materials with improved actuation capabilities; the second is device-oriented, aimed at the realization of actuating configurations able to get the most of the material potentialities; the last is system-oriented, to pursue the ambitious aim of the realisation of truly bioinspired systems. We underline here that the extreme complexity of the muscle molecular structure makes today realistic only a functional not-structural biomimesis of natural systems of actuation, at least in terms of practical engineering devices.

Concerning the materials used for the realisation of “*artificial muscles*”, currently our research activities are focused on two particular classes of EAP, requiring two complementary stimulation conditions: conducting polymers (requiring low driving voltages) and dielectric elastomers (requiring high voltages).

The first part of the chapter is devoted to conducting polymers. After a brief description of their principal characteristics, the attention is focused on the most recent results reported in literature, generally finalized to increase the actuator performances. Our latest specific contribution is described as well: it consists of a collection of some of the latest proposed approaches, in order to combine their effects and the recent observation that the radial expansion of the samples is several times higher than the axial one. However, being the axial deformation more useful, from a practical point of view, than the radial one, we propose a device able to transform radial in axial deformation, with an high amplification factor too.

The second part of the chapter concerns actuators made of dielectric elastomers. Following the description of principle of operation and state of the art, our latest results in order to gain a reduction of the driving electric field are shown. Furthermore, a new actuating configuration able to generate axial contractions (instead of more common axial active elongation) is presented. Our latest research avenues for bio-inspired muscular control, aimed at the realization of *truly* “*artificial muscles*”, complete the chapter.

## 2. Conducting polymers

Electron conducting polymers (CP) belong to the class of materials for direct drive actuators [12,13,14] and they may lead to useful realizations because of:

-sizeable active stress (various MPa)

-low electrical potential difference required to elicit their response (a few Volts).

In conducting polymers the electromechanical energy transduction originates from the doping and de-doping processes in the polymeric chain. Indeed, oxidation of the polymer chain changes its bound charge and intercalation (or de-intercalation) of doping ions occurs, causing structural and dimensional

changes of the material to restore electroneutrality [12].

Noticeable changes in mechanical properties are associated to these phenomena and the control variable of a CP actuator is the exchanged ionic charge between the polymer and the surrounding. An example of the relation of strain vs. exchanged density charge is reported in Fig. 2.1.

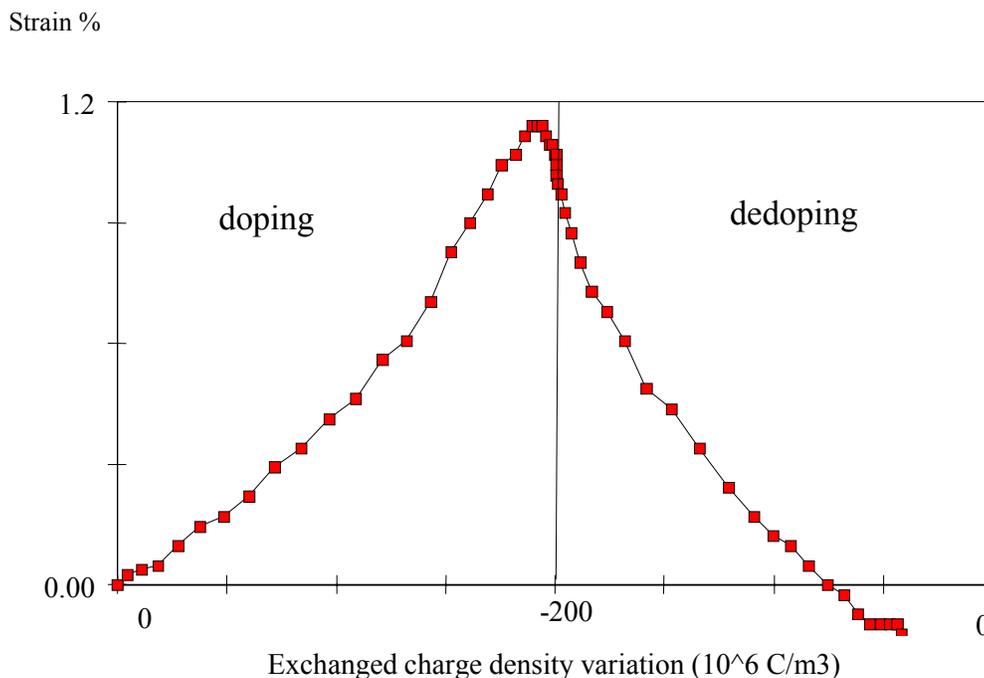


Fig. 2.1 Strain vs. exchanged density charge during cyclic voltammetry of polypyrrole doped with benzen sulfonate.

Unfortunately, several aspects limit up to date an extensive use of conducting polymers as actuators, such as low axial strain, low response time and low ciclability.

Although encouraging results have been obtained in the last years, much work has yet to be done in order to have a usable and effective system. In the following pages we present the more recent approaches developed to improve actuator performance.

In particular, attention will be given to radial strain, which shows higher values of the axial strain and for this reason should deserve attentions.

Along this line, the final part of this Section reports on a new device able to amplify and transform the radial strain in axial strain, leading to an amplification factor up to 50.

### 2.1. Recent results on conducting polymers from literature

The most important problems related to the actuation properties of conducting polymers are connected to their electrochemical nature. The applied voltage is not entirely utilized to change the degree of doping of the polymer, but it is in part spent in the cell resistance and, in part, lost in parasite reactions. These processes are generally irreversible, so there is a progressive decreasing of the actuator performances and of lifetime. In response to these problems different strategies have been proposed.

Lu et al. [15] addressed their research toward new surrounding media, in order to find proper stimulation conditions to increase the conducting polymer performances. In this context they found new solvents (liquid electrolytes as namely 1-butyl-3-methyl imidazolium cation (BMIM<sup>+</sup>), tetrafluoroborate (BF<sub>4</sub><sup>-</sup>) or hexafluorophosphate (PF<sub>6</sub><sup>-</sup>) anions) able to limit parasite reactions. This property is maintained within large stimulation voltage ranges, causing higher active strain.

Madden et al. [16] and Barisci et al. [17] applied short duration pulses of unusually high potentials, resulting in increased strain rate and power-to-mass ratio achievable from polypyrrole actuators of a given thickness. Most of the potential drops initially across the solution and not at the polymer/electrolyte interface, explaining why there is no rapid polymer degradation at short times, despite the high potentials used.

Hara et al. [18] reported very interesting strain and stress results (strain up to 12%, stress up to 22 MPa) for polypyrrole prepared electrochemically on Ti electrode from pyrrole 0.25 M and tetra-n-butylammonium tetrafluoroborate, both in methyl benzoate. Electrochemical tests were performed in NaPF<sub>6</sub> aqueous solution.

Similarly, high strains in aqueous solution were also obtained by Bay et al [19], by holding up the polypyrrole film with a very thin corrugated gold electrode. The gold electrode does not constrain the film and consents uniform distribution of the potential all along the sample.

In parallel to these approaches, an important observation has been made by Smela et al. [20]. They focused their attention on small samples on polypyrrole doped with dodecylbenzenesulfonate. They observed that the volume change was strongly anisotropic. In fact, in response to an electrochemical stimulation film thickness can increase over 35%, while the strain in the orthogonal directions can be considerably lower.

## 2.2. *From radial to axial strain*

Starting from the previous reported observations, our work has been devoted to combine in a single system some of these solutions and to design a system able to transform radial strain in a more useful axial strain.

We evaluated the radial strain in the case of pyrrole polymerized on a stainless steel wire [21, 22].

Electrochemical tests were performed in the same kind of liquid electrolytes suggested by Lu et al. [15]. An example of the experimental set-up is reported in Fig. 2.2.

The measured radial strain was over 50%, up to 80%, with an exchange density of about 8 mC. The variation of the strain is opposite to that of the charge, and that suggests a prevalent cationic flux.

Since in this kind of experiment no load is applied to the sample, it is not possible to infer the stress induced by the electrochemical stimulus.

Concerning the radial stain it is important to report an observation of Chiarelli and De Rossi [23] on free swelling of gels, which, if confirmed for conducting polymers too, could have important consequences. They noted that for a gel material in a cylindrical configuration the response time in the radial direction is shorter than the one in the axial direction, even though both times have the same dependence on the radial dimension of the sample.

This can be intuitively explained if one thinks that the in/outward diffusive flow in the direction orthogonal to the surface mainly exposed to the electrochemical reservoir is mechanically less limited and, therefore, the swelling process is faster.

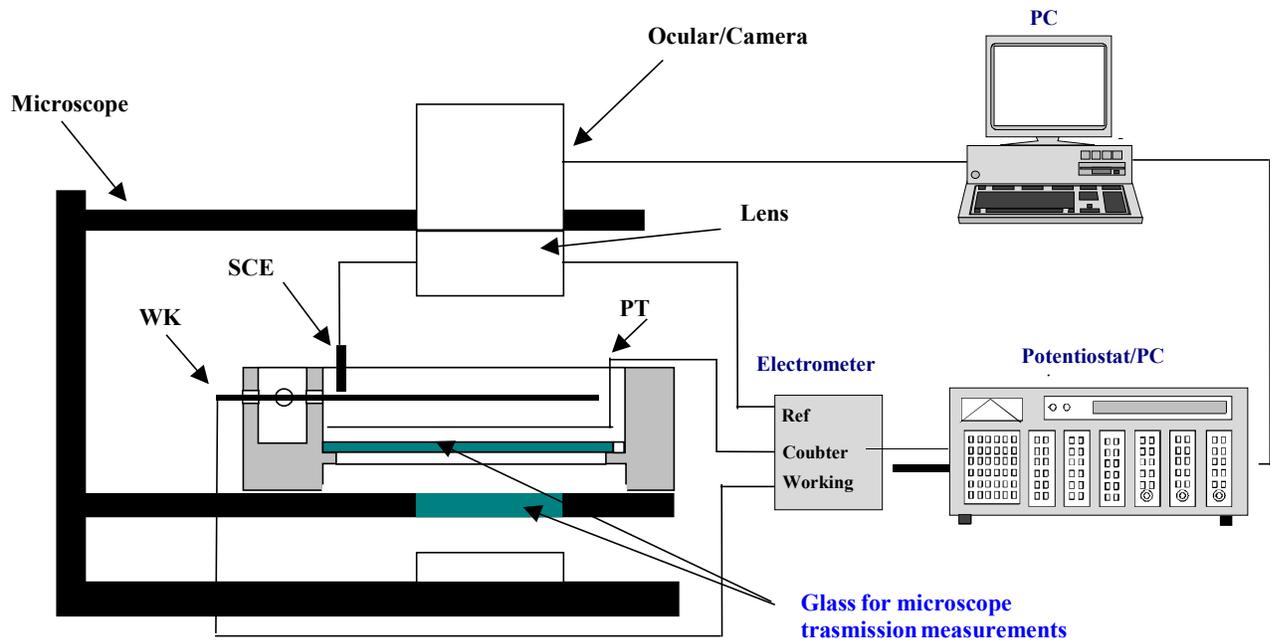


Fig. 2.2. Electrochemical cell for dimensional measurements under microscope.

As reported above, the radial strain produced by the fibers could be transformed in an axial strain. A useful device to transfer and amplify the strain is the Mc Kibben actuator [24].

In the classical version of this device a cylindrical rubber bladder is covered by a braid mesh, made of flexible, but not extensible, threads. Both ends of the bladder are connected to the mesh. By changing the force applied to the free end of the mesh and the pressure inside the bladder, the mesh shape changes: its diameter increases and its length decreases.

In our version of Mc Kibben actuator, the bladder is substituted with a bundle of conducting polymer hollow fibers. In the center of the hollow fiber a rigid metal wire works as electrode. A liquid electrolyte and a metal counter electrode complete the system.

By neglecting boundary effects, we have an active bundle made of two kinds of materials: the fibers and the interstices. These latter can be filled with an incompressible material. In this case, we expect that the load will be ideally transmitted through the fibers and that the bundle will have the same radial strain of any single fiber.

Let us consider the following assumptions:

- the length of each fiber is constant (fiber length is constrained to maximize the radial strain);
- the mesh is wrapped around the fiber bundle;
- the mesh has to be able to shorten, accordingly to the fiber diameter change;
- the mesh is clamped only to one end of the bundle and an external mechanical load ( $F$ ) is applied to the other end of the mesh (Fig. 2.3);
- the mesh is longer than the fibers: it always covers them and the pressure produced from the external load on the bundle can be assumed to be constant on each point;
- the mesh has to be free to move over the fiber length;
- the thread of the mesh is assumed not to be extensible and to produce no stress on the fibers, in absence of load. Nonetheless, it will maintain compact the fibers of the bundle also at  $F = 0$ . This means that the mesh diameter is exactly the fiber group diameter.

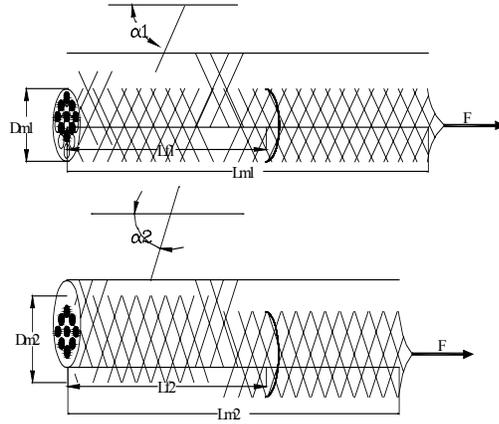


Fig. 2.3. Scheme of the modified Mc Kibben actuator.

The actuation mechanics of such a device has been recently described [21], by performing an electro-chemo-mechanical analysis, splitted in the following two parts. In the first one, the relation between the radial strain of each fiber and the electro-chemical stimulation was evaluated by means of an electro-chemo-elastic analysis; this was carried out by using the well-known theoretical tools proper of thermo-elastic descriptions, where the thermal factor was substituted with a term depending on the exchanged density charge [25-28]. The second part of the analysis was focused on a geometrical study of the device, aimed to the determination of the relation between the radial and the axial strain. As an assumption, the load was considered ideally transmitted from fiber to fiber and the radial strain of the entire bundle was assumed equal to that of each single fiber. The following expressions of the radius  $R_m$  and length  $L_m$  of the mesh, having helical threads, were considered:

$$R_m = \frac{b}{2n\pi} \sin(\alpha) \quad (2.1)$$

$$L_m = b \cos(\alpha) \quad (2.2)$$

where  $b$  is the length of one thread,  $n$  is the number of turns of a thread around the axis and  $\alpha$  is the angle between a thread of the mesh and the axial direction (Fig. 2.3). The entanglement of the fibers guarantees the ratio  $b/n$  to be constant during motion. It has been reported [22] that the radial strain

$$\rho = \frac{\Delta R_m}{R_m} \quad (2.3)$$

and the axial strain

$$\lambda = \frac{\Delta L_m}{L_m} \quad (2.4)$$

are related by means of the following equation:

$$\lambda = \frac{(\rho + 1) \cos(\alpha_2)}{\sqrt{(\rho + 1)^2 - \sin^2(\alpha_2)}} - 1 \quad (2.5)$$

being  $\alpha_2$  the final thread angle (Fig. 2.3).

Eq. 2.5 shows that, under the mentioned hypotheses, the axial strain of the actuator is correlated to the mesh thread angle and to the fiber radial strain, while it is independent of the number of turns of the mesh thread.

As mentioned above, the Mc Kibben mesh can be able to operate an amplification of the strain, as well as its radial to longitudinal conversion. The amplification gain ( $k$ ) can be defined as the ratio between the axial and the radial strain of the actuator:

$$k = \frac{\Delta L_m / L_m}{\Delta R_m / R_m} \tag{2.6}$$

This gain factor depends on the maximum inclination angle  $\alpha_2$  permitted by the manufactured mesh; the higher the angle, the higher is the gain. The theoretical upper limit is reached when this angle equals  $90^\circ$ , which is an evidently ideal value, corresponding to a completely collapsed mesh with horizontal threads.

Simulations results presented in Fig. 2.4 show, for different values of  $\alpha_2$ , the dependence of the axial strain and amplification factor on the radial strain.

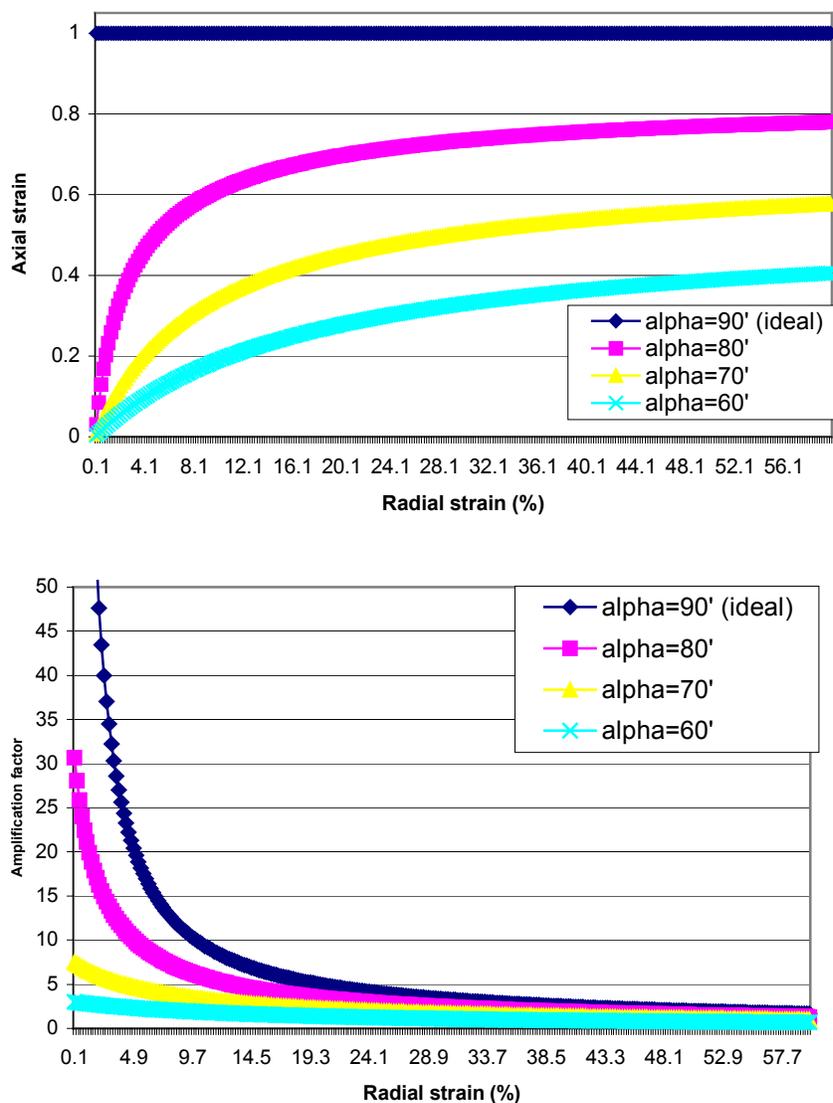


Fig. 2.4. Theoretical values of axial strain and amplification factor, achievable with the described Mc Kibben mesh.

According to the calculated results reported in Fig. 2.4, within the considered range of radial strain it is possible to reach axial strains of different magnitude, ranging from 25% up to 80%, depending on the inclination angle of the mesh. Table 1 summarizes significant results for a radial strain of 20%.

Table 1. Axial strain vs. inclination angle of the mesh, for a radial strain of 20%.

<b>Inclination angle <math>\alpha_2</math> of the mesh</b>	<b>Axial strain <math>\lambda</math> for a radial strain <math>\rho=20\%</math></b>
60°	25%
70°	40%
80°	65%
90° (ideal)	100%

These data show that the axial strain increases remarkably with the inclination angle of the mesh. Unfortunately practical reasons related to the complexity of the manufacturing process of the mesh limit the upper bound to the feasibility of a high inclination angle of the threads. Inclination angles between 60° and 70° are expected to be realizable, therefore, axial strains of 25-40% could be obtained, as a hugely exciting performance.

Nevertheless, such theoretical predictions have to face up with the inevitable losses due to internal friction, which, in general, increases with the mesh inclination angle. On this regard, an analysis of such a aspect has been recently proposed [22].

### 3. Dielectric elastomers

Dielectric polymers having a low elastic modulus and subjected to an high electric field exhibit large deformations, mainly due to a Coulombian effect, that is by means of the electrostatic interactions among the free charges on the electrodes. In particular, when a thin film of such materials is sandwiched between two compliant electrodes (e.g. made of carbon conductive grease) and a voltage difference is applied between them, the polymer undergoes an electric field-sustained deformation, consisting of a thickness squeezing and a related surface expansion (Fig. 3.1) [10,29-34].

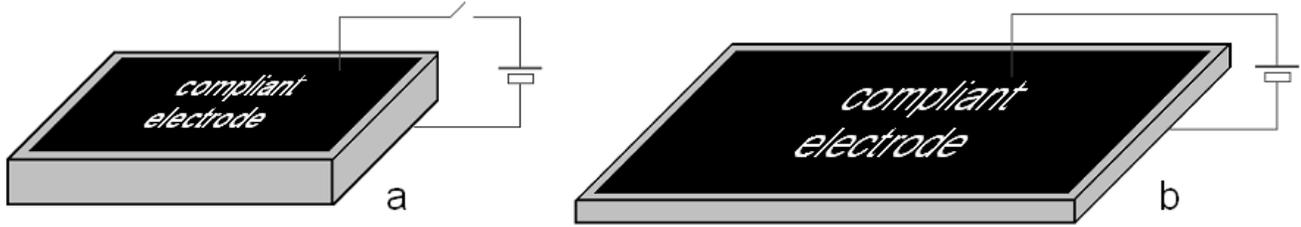


Fig. 3.1 Principle of operation of a dielectric elastomer planar actuator. The electrode compliance supports the polymer squeezing from the rest condition (a) to the electrically stressed one (b).

The Coulomb force acting between the electrode free charges represents the so-called Maxwell stress for this electromechanical phenomenon. This kind of stress acts to any kind of dielectric material subjected to an applied electric field [35]. However, it is evident that the corresponding deformations will be markedly emphasized by the eventual compliance of the electrodes, as well as by the polymer softness. These last two key-features macroscopically distinguish the actuating devices made of dielectric elastomers from those based on different electric-field-driven dielectrics, such as piezoelectric or electrostrictive materials.

Acrylic and silicone rubbers are the most representative materials of the dielectric elastomer class for actuation purposes. Such kinds of polymers are very compliant and show the highest actuating deformations among all EAP materials [10].

By assuming a dielectric elastomer as a linearly elastic body, having a Young's modulus  $Y$  and a relative dielectric constant  $\epsilon_r$ , linear strains  $S$  along the direction of an applied electric field  $E$  are described by the following equation [10,29]:

$$S = -\frac{1}{Y} \epsilon_0 \epsilon_r E^2 \quad (3.1)$$

where  $\epsilon_0$  is the free-space dielectric permittivity ( $\epsilon_0=8.85 \times 10^{-12}$  F/m). The quadratic dependence of the strain on the applied field, predicted by the previous equation, is shown by the examples of Figs. 3.2a and 3.2b.

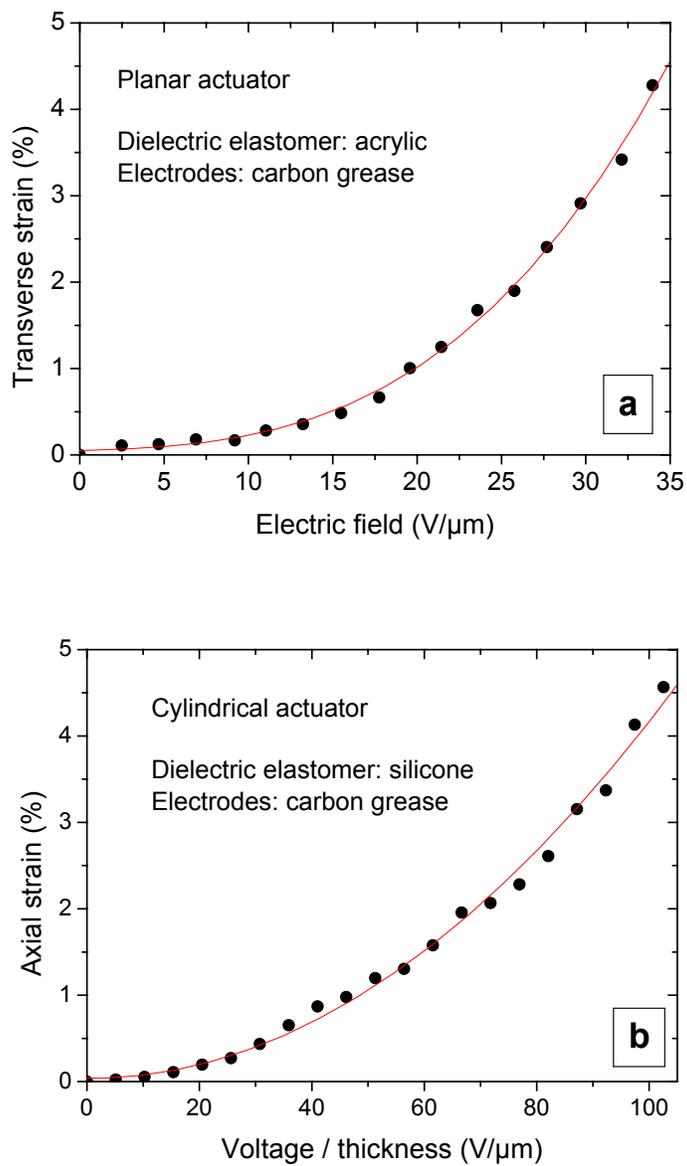


Fig. 3.2. Active strain plots and fitting curves for an acrylic planar actuator (a) and a silicone cylindrical actuator (b), realized in our laboratory.

Molecular structures of acrylic and silicone polymers are represented in Figs. 3.3a and 3.3b.

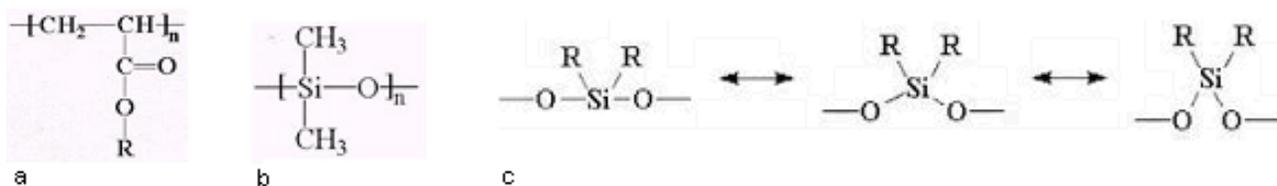


Fig. 3.3. Molecular structure of acrylic (a) and silicone (b) polymers and molecular origin of silicone elasticity (c).

Acrylic polymers (or acrylates) can exhibit high-level actuating capabilities: thickness strains up to 60-70 % at 400 V/ $\mu\text{m}$ , area strains up to 200 % at 200 V/ $\mu\text{m}$  and corresponding stresses of some MPa [10]. Such performances are enabled by low elastic moduli and high dielectric strengths (dielectric breakdown can occur at electric fields of about 500 V/ $\mu\text{m}$ ). Material prestretching, before the application of the electric field, was demonstrated to increase active performances, due to the increase of the polymer elasticity and dielectric strength [10].

Silicones are easily processable (by spin coating, casting, etc.) and permit the realization of good rubber-like dielectrics, owing to the high mobility of the Si-O bonds, which confer high flexibility to main chains, as shown by Fig. 3.3c. Dielectric breakdown of silicones occurs at electric fields ranging from 100 to 350 V/ $\mu\text{m}$ , enabling thickness strains up to 40-50 % and area strains up to 100 % with related stresses of 0.3-0.4 MPa [10].

The excellent figures of merit possessed by dielectric elastomers in several respects (high actuation strains and stresses, fast response times, high efficiency, stability, reliability and durability) make them the most performing materials currently available for polymer actuation. The price for achieving these high-level capabilities is represented by the high driving electric fields needed (order of 100 V/ $\mu\text{m}$ ). For a definite polymer thickness, such field levels can be reached by applying high voltages, which may be disadvantageous in many applications. For this reason, in order to reduce the driving electric fields, polymers with high dielectric constant are necessary (Eq. (3.1)).

With respect to this, new highly dielectric elastomers could be realized by means of a composite approach. By filling an ordinary elastomer (e.g. silicone) with a component having a greater dielectric permittivity, it is possible to obtain a resulting material showing the fruitful combination of the advantageous matrix elasticity and filler permittivity. On this regard, Fig. 3.4 shows preliminary results recently obtained in our laboratory, concerning the active deformations of a silicone elastomer (relative permittivity of about 5) mixed with titanium dioxide (relative permittivity of about 80).

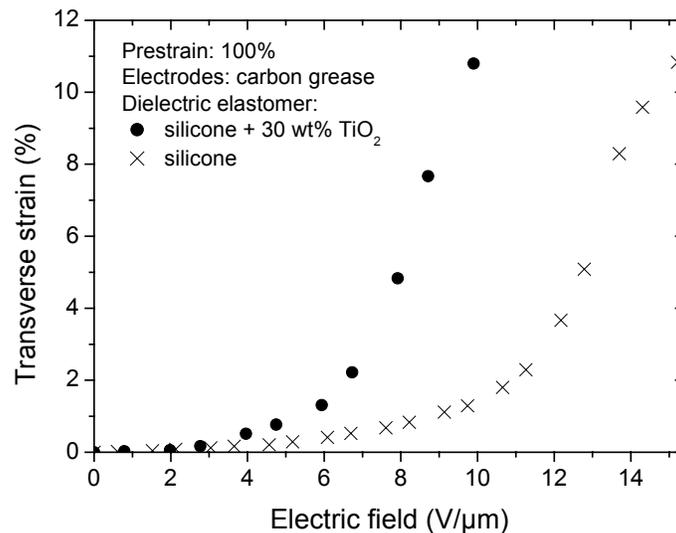


Fig. 3.4. Strain-electric field curves of planar actuators made of a pure (dots) and TiO<sub>2</sub>-loaded (crosses) silicone elastomer, realized in our laboratory.

So far, many configurations for dielectric elastomer actuators have been proposed and demonstrated, including planar, tube, rolled, extender, diaphragm, bimorph and unimorph benders [10, 29-34]. Tube and, mainly, rolled actuators (Fig. 3.5) would be potential candidate for the realization of “*artificial muscles*” with linear (along a line) actuation capabilities. However, they present a mechanism of actuation opposite to that of natural muscles: under electrical stimulation they elongate, instead of contract.

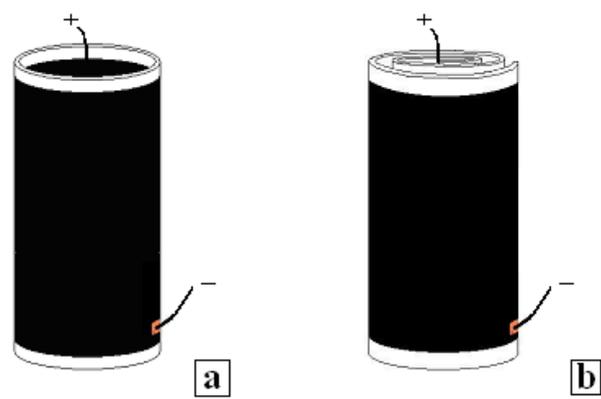


Fig. 3.5. Tube (a) and rolled (b) actuators made of dielectric elastomers.

In order to overcome this limitation, a new type of actuator, able to undergo electrically-activated contractions, is currently under development in our laboratory, based on a configuration recently patented [36] and reported in Fig. 3.6. It consists of an hollow cylinder of dielectric elastomer (e.g. silicone), having two helical compliant electrodes integrated within its wall. By applying a voltage difference between them, the attraction among opposite charges causes an axial contraction of the actuator.

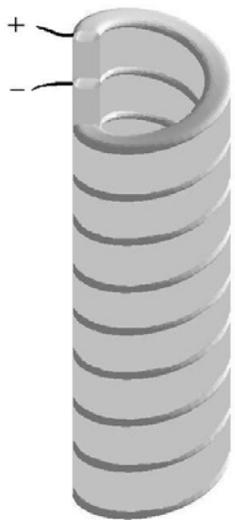


Fig. 3.6. Novel actuating configuration.

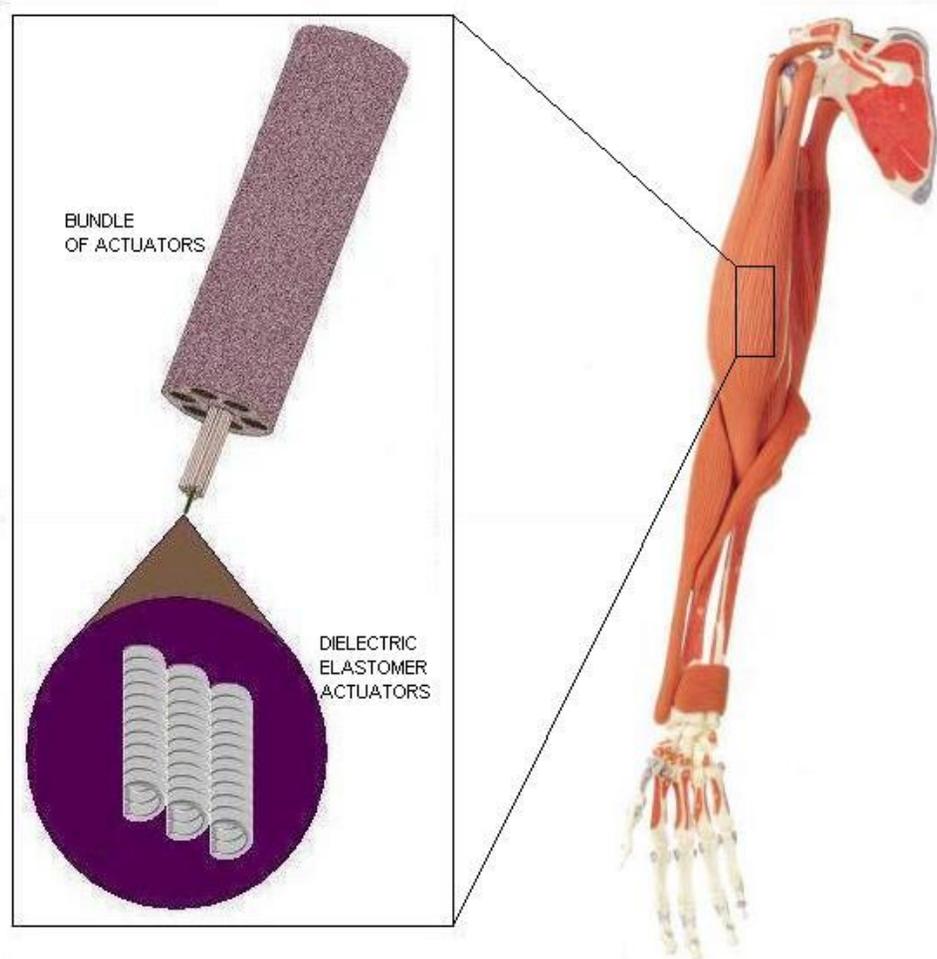


Fig. 3.7. Bundle of axially-contracting actuating units as “artificial muscle”.

By assuming the dielectric elastomer as a linearly elastic body, it can be demonstrated that the axial strain generable by this configuration is described by the following relation:

$$Axial\ Strain = \frac{\epsilon_0 \epsilon_r E^2}{Y} (\cos \alpha - \sin \alpha) \quad (3.2)$$

where  $\alpha (>\pi/4)$  is the angle of inclination of each electrode respect to the device axis. According to this equation, higher values of  $\alpha$  lead to higher generable axial strains (and stresses), owing to the improved alignment of the applied electric field along the working direction. This last feature makes the actuating configuration potentially able to generate strains and stresses higher than those permitted by several previously existing polymer devices.

This new actuator is believed to be useful in each application requiring spring-like contractions activated and modulated by an electrical signal. For this reason, it represents a suitable candidate to be used in a bundle of elementary actuating units, for the realisation of “artificial muscles” able to replicate the functional properties of natural muscles (Fig. 3.7). “Artificial muscles” are intended here

not as implantable actuators, but as devices useful to move bioinspired robotic arms or exoskeletons.

Such a functional mimicking is intended to be addressed with respect to the capability of a bundle of actuators to show both bio-inspired electrically activated contractions and bio-inspired force-length characteristics. Concerning the second requirement, two alternative approaches, described below, are under development and evaluation in our laboratory, in order to replicate the substantially quadratic characteristic of muscles, as described by the Feldman's muscle model [37,38]. In fact, according to this model, the contracting force ( $F_m$ ) exerted by a muscle depends on its intrinsic pseudo-stiffness ( $k$ ), its effective length ( $x$ ) and its rest length ( $\lambda$ ) as follows:

$$F_m = k(x - \lambda)^2 u(x - \lambda) \quad (3.3)$$

where  $u(x - \lambda)$  is the Heaviside function.

The first approach considered for the mimicking of such a curve is quite rough and it simply exploits basically the passive mechanical properties of the elastomeric material with which the actuating devices of the bundle are made. In fact, the quadratic trend of the muscle force-length curve (Eq. 3.3) can be regarded as analogous to the trend of the passive characteristic shown by any rubber-like material in the region of high strains [39]. In this respect, Fig. 3.8 shows a comparison between typical muscle curves and passive engineering stress-strain characteristics of two dielectric elastomers. The replication of the muscle active behaviour may be obtained by a superimposition of both the passive and active contributes of the force developed by each actuator of the bundle.

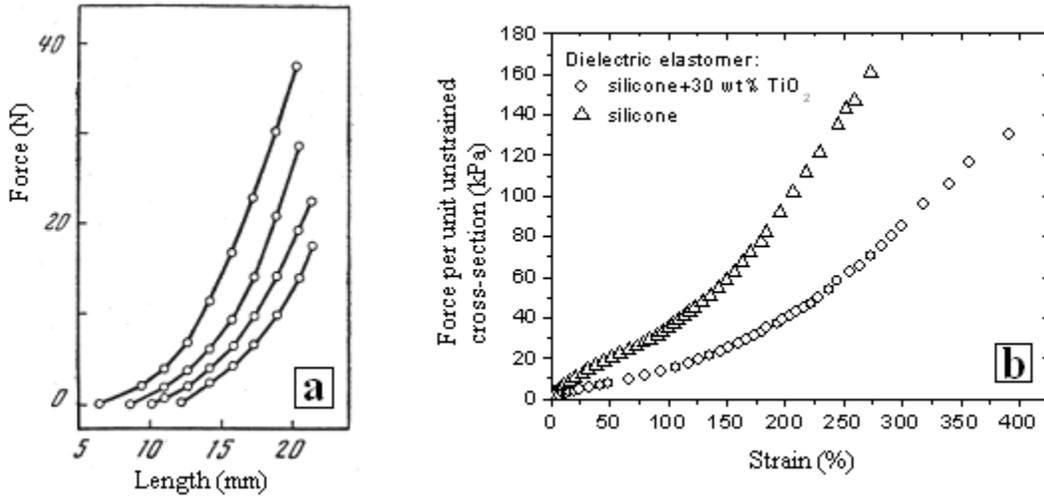


Fig. 3.8. Typical force-length curves of natural muscles for different stimulation frequencies (a) and engineering stress-strain passive characteristics for dielectric elastomers, presenting approximately a quadratic trend for high strains (b).

A more refined, but also complicated, alternative has been formulated as second approach for the mimicking of the muscle curve. This method goes more deeply in the functioning of a natural muscle, as modelled by Feldman. Since the different muscle characteristics for different stimulation frequencies analytically depend only on the value of the muscle rest length  $\lambda$  (Eq. 3.3 and Fig. 3.8a), Feldman assumed that the muscular control exerted by the central nervous system (CNS) could be regarded in terms of a  $\lambda$  control. This led to a new conception of muscular contraction: the CNS is able to set and control values of  $\lambda$ , to make a muscle to work on a definite characteristic. To improve the force exerted

by a muscle, for a given length, the CNS sets a lower value of  $\lambda$ , to shift the working point onto an higher characteristic. For this reason, an actuating system showing a quadratic force-length characteristic and having the capability to hold sizable values of a parameter analogous to the muscle rest length  $\lambda$  would perform a smart replication of muscle mechanics. Moreover, the mimicking of a further muscular feature is required to a candidate “*artificial muscle*”: the capability to permit a control of its geometrical status, as well as of its stiffness. In fact, the variable slope of the force-length characteristic of a muscle (Fig. 3.8a) allows humans to modulate their interaction with their surrounding environment, more or less “softly”, in relation to the variable compliance held by their muscular system. With respect to this, it is to underline that biological actuation is mainly exerted by couples of muscles (agonist and antagonist for a given movement), which contract (in terms of  $\lambda$ ) in different ways, modulated by the central nervous system. For an agonist/antagonist configuration it is possible to demonstrate [40,41] that an actuating system showing a force-length relation similar to that of Eq. 3.3 would enable the requested control of both position and stiffness (or, dually, compliance). Such an actuating system having a quadratic force-length characteristic with sizable  $\lambda$ , mimicking muscle mechanics and enabling position and stiffness control, can be designed with a functional structure and a driving strategy inspired to those adopted by natural muscles. The muscle is constituted by a set of active elements (motor units) with different rest lengths. These units are activated by a progressive recruitment, whereby the quadratic characteristic of the entire muscle is due to the superimposition of those of the elementary units. Following this concept and taking in consideration a bundle of active units made of dielectric elastomer actuators, a suitable algorithm was developed in our laboratory [40,41], to design the actuating units and define their driving strategy, in order to reproduce the characteristic of Eq. 3.3. The idea consists of an approximation of the quadratic characteristic, by using appropriate recruitment and driving strategies of ad-hoc designed units. The algorithm is valid for both elongating or contracting units in response to an electrical stimulus, regardless of their physical constitution, and it could be fruitfully applied to a bundle of the new contracting actuators under development, described above.

#### 4. Conclusions

Some of the latest findings and research avenues in the realization of “*artificial muscles*” have been shown in this chapter.

Presented results related to conducting polymer actuators show that dimensional change in the radial direction can be more than one order of magnitude higher than its axial counterpart. The latest methodologies here discussed assess the worth of exploiting the interesting radial strain values of conducting polymer fibers. A possible device able to convert radial to longitudinal strain via braided mesh acting as a merely mechanical transducer has also been described. These technical improvements and observations, together with a voltage drop range acceptable for biomedical applications, give conducting polymers a new appeal for this kind of utilization and promise new interesting applications.

Concerning dielectric elastomer actuators, recent progresses on material development, toward a decrease of the driving electric fields, were here reported, as well as the presentation of a novel actuating configuration, designed to generate biomimetic axial contractions. Furthermore, two research avenues aimed to complete the functional biomimesis of muscles, including a faithful reproduction of their force-length characteristic, were separately described.

As can be easily deduced from presented results, a great deal of work has still to be done and the solution of a large number of problems has to be given, in order to reach the ambitious objective consisting in the realization of “artificial muscles”.

In particular, as examples of basic technical challenges on which we are currently working, one should consider the following main challenges: concerning conducting polymer actuators, by using liquid electrolytes and IR compensation lower response times and higher lifetime and reliability may be reached; with respect to dielectric elastomer devices, materials capable to be driven by lower fields are needed. Suitable new actuating configurations are currently under development and require many technological skills. Moreover, the definition and implementation of sound methodologies to design and drive a bundle of actuators showing muscle-like mechanics requires the concentration of further efforts.

Fulfilment of so many and different needs would significantly contribute to enable the mimicking of the great performances of natural muscles, which represent an ideal model for our systems of actuation.

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