

Artificial Muscles for Humanoid Robots

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1. The question of actuator choice for humanoid robots

It is important to recall that humanoid robot technology derives from the technology of industrial robots. It is obvious that the developments of bipedal robots such as the integration of robot-upper limbs to complex anthropomorphic structures have benefited from progress in mechanical structures, sensors and actuators used in industrial robot-arms. A direct link is sometimes made between the technology of redundant robot-arms and humanoid robots as underlined in some technical documents of the Japanese AIST where it clearly appears that the HRP2 humanoid robot upper limb is directly derived from the Mitsubishi PA10 7R industrial robot-arm.

Due to its high number of degrees of freedom in comparison to industrial robots, a humanoid robot requires great compactness of all actuator and sensor components. This is why we believe that the harmonic drive technology associated with direct current electric motor technology has played a non-negligible part in humanoid robot development. The DC actuator offers the great advantage of being a straightforward technology, associated with simple and well-known physical models, its integration into mobile robots benefits from new developments in embedded batteries. However, its low maximum-torque-on-mass and maximum-torque-on-volume ratios are a serious drawback for its use in direct drive apparatuses. On the other hand, the ability of electric motors to generate very high velocities in comparison with moderate jointed velocities needed by industrial robot-arms and more by jointed anthropomorphic limbs, gives the possibility of using high ratio speed reducers to amplify motor-torque. Moreover, the choice of a high ratio speed reducer offers the advantage of masking inertial perturbations such as external torque perturbations. The technical achievement of such ratios induces specific mechanical difficulties due to the bulkiness of successive gears; harmonic drive technology - represented for example by Harmonic Drive AG - resolves this problem in a very elegant manner: the harmonic drive and the actuator fit together without excessive increase in mass and volume in comparison with the actuator alone. It can be considered that most of today's humanoid robots are actuated by DC motors with harmonic drives (this actuation mode is mentioned, for example, by Honda from its first paper about the P2 robot onwards (Hirai *et al.*, 1998) and then in the official ASIMO web site, as well as in papers concerning other Japanese and European humanoid robots). But if this technology simplifies actuator mechanical integration and leads to the use of simple joint linear control, despite the highly non-linear character of robot dynamics, it is well-known that the use of a speed reducer multiplies the

joint stiffness by the its ratio squared. A high joint stiffness contributes to joint accuracy and repeatability, but also induces a high danger level for users, which can be acceptable in the case of industrial robot-arms separated from factory staff by special safety devices, but becomes very problematical in the case of humanoid robots intended for working in public environments. The need to find an actuation mode which associates power, accuracy and a 'softness' adapted to human presence, that is the question of actuator choice in humanoid robotics. To address this problem we will first try to define the notion of artificial muscle in paragraph 2, then deal with the question of artificial muscle actuators for humanoid robots in paragraph 3, before analysing their integration within anthropomorphic limbs (paragraph 4) to finish with their control (paragraph 5).

2. Notion of artificial muscle

2.1 Performance criteria in the research of new actuators

A general theory of actuators does not exist; each actuator is defined according to the physical theory on which its legitimacy is founded. A comparison of actuators can as a consequence be delicate. This is why actuator designers have introduced a certain number of performance criteria aimed at making such comparisons easier. In general, actuation can be defined as a process of converting energy to mechanical forms, and an actuator as a device that accomplishes this conversion. **Power output per actuator mass**, and **per volume**, as actuator **efficiency** - defined as 'the ratio of mechanical work output to energy input during a complete cycle in cyclic operation' (Huber *et al.*, 1997) - are three fundamental properties for characterizing actuators. However, artificial muscle technology considers more specific performance criteria so as to accurately specify new actuator technology in comparison with 'natural muscular motor' properties. The following terminology, justified by the linear actuator character of the artificial muscle, generally completes the power criteria - the definitions given in inverted commas are from (Madden *et al.*, 2004) :

- **Stress** : 'typical force per cross-sectional area under which the actuator materials are tested'; **maximum stress** corresponds to the maximum stress that can be generated in specified functioning conditions; as will be seen later, it is important for a given technology to specify the 'actuator materials' relating to stress: typical stresses of strips or fibres of a given technology is not obligatorily similar to that of the artificial muscle composed of a set of these strips or fibres;
- **Strain** : 'displacement normalized by the original material length in the direction of actuation'; **maximum strain** and maximum stress are according to Huber & others 'basic characteristics of an actuator [since] for a given size of actuator they limit the force and displacement' (Huber *et al.*, 1997, p. 2186); the terms **contraction ratio** and **maximum contraction ratio** will also be used;
- **Strain rate** : 'average change in strain per unit time during an actuator stroke'; the term '**response time**' - in the sense given by control theory, will also be used to characterize the speed of artificial muscle dynamic contraction;
- **Cycle life** : 'number of useful strokes that the material is known to be able to undergo'; this notion specifies the heavy-duty character of artificial muscle in given working conditions; in practice this is an important notion since artificial muscles are essentially made of 'soft' materials which can be weakened by shape changes imposed by the actuation mode;

- **Elastic modulus:** ‘material stiffness multiplied by sample length and divided by cross-sectional area’; this is a typical material science notion; when the artificial muscle is considered as an actuator to be controlled, **stiffness** – and its inverse – **compliance** notions can appear more appropriate.

It is important to note that this criteria list is not exhaustive; depending on author, other performance criteria can be found: as an example, Huber *et al* consider two criteria not listed above: **actuator density** (the ratio of mass to initial volume of an actuator) and **strain resolution** (the smallest step increment of strain) (Huber *et al.*, 1997). The first directly concerns humanoid robotics: like skeletal muscles, artificial muscles are assumed to be located in moving robot links; the second criterion can be interesting in a control perspective to specify artificial muscle sensitivity.

Furthermore, we believe that the theoretical framework proposed by Hannaford and Winters (Hannaford & Winters, 1990) for the analysis of actuator properties based on Paynter’s terminology of generalized dynamic systems (Paynter, 1961) can be particularly useful in our study: these authors propose characterizing any actuator by its two effort-flow and effort-displacement characteristics, where ‘effort’ represents the output force or torque, and ‘flow’ the linear velocity or angular velocity. For example, the DC motor is ideally characterized by a linear effort-flow curve and a constant effort-displacement curve, as illustrated in Figures 1.a and 1.b. Figures 1.c and 1.d give the typical characteristics of the skeletal muscle by the association of ‘effort-flow’ characteristics corresponding to the so-called tension-velocity curve with the ‘effort-displacement’ characteristics corresponding to the so-called tension-length curve. We will return to this in the modelling of skeletal muscle, but it is important to note the fundamental originality of skeletal muscle characteristics: while most of the actuators have constant or pseudo-constant ‘effort-displacement’ characteristics, this is not so for skeletal muscle. As a dynamic system in Paynter’s sense, the ‘effort-displacement’ relationship defines passive element C (for compliance or capacitance). Classical actuators generally have infinite compliance; a dependence of force/torque on position can even appear as a drawback: it is up to the actuator control and not the actuator itself to impose this dependence. Conversely, living actuators aimed at a ‘relationship life’ have to combine the generation of the power necessary for body mobility with a non-infinite compliance making for easy contact with the environment – we will use later the term ‘natural’ compliance to characterize this compliance peculiar to the skeletal actuator. Research on artificial muscles can be understood as a new attempt to mimic the living so as to integrate it into a machine – the humanoid robot – an original power-softness combination, yet glaringly absent in machine technology.

2.2 The historical Kühn and Katchalsky notion of artificial muscle as a gel swelling and deswelling under the effect of a chemical agent

The origin of the artificial muscle notion must be found in the first works of chemists on certain materials whose swelling can be controlled in a reversible manner. At the end of the 1940s, Kühn & Katchalsky did indeed prove that an aqueous polymeric gel essentially composed of polyacrylic acid ‘... is found to swell enormously on addition of alkali and to contract rapidly when acid is added to the surrounding solution. Linear dilatations and contractions of the order of 300 per cent were observed. This cycle of swelling and deswelling is reversible and can be repeated at will’ (Kühn *et al.*, 1950, p.515). At that time the

authors had designed a device transforming chemical energy into mechanical working in the form of a 0.1 mm thick filament able to lift up a 360 mg load in some minutes when swollen. Thus the original Kühn & Katchalsky experiment historically created the present artificial muscle notion as a reversible contractile device. Katchalsky emphasized the natural tendency of artificial muscles to generate an 'equilibrium swelling' brought according to him about two opposing tendencies : first, '...the solution tendency of the polymeric molecules and the osmotic pressure of the cations of the alkali bound by the gel', secondly '...the caoutchouc-type contraction tendency of the stretched polymer molecules' (Katchalsky, 1949, p.V/8). More generally, we will go further and apply this natural tendency to an equilibrium state of any artificial muscle type as an open-loop stability in position. It is important to note, however, that this ability to pass from an equilibrium state to another state would be nothing if it were not associated to a reversibility whose life cycle is finally its measure. Note also that the life cycle of natural muscle is greater than 10^9 (Hunter & Lafontaine, 1992); no present-day artificial muscle is able to approach this value, which is linked to the ability of living tissues to self-repair. Kühn & Katchalsky's historical studies were reconsidered in the 1980s within the framework of a renewed interest for artificial muscles due to technological developments in robotics, and a demand for implantable artificial biological organs. However, from a practical point of view, the Kühn & Katchalsky actuator displays a major disadvantage: its excessively slow response time (in the order of minutes). More generally, it will be seen throughout this chapter that the major difficulty in the design of artificial muscles for robots consists of obtaining both quick response time and high-power-stress to mass-and-power to volume, adapted to the integration of artificial muscles to human-dimensions and mass anthropomorphic limbs. There now follows a brief overview of present-day artificial muscle technologies.

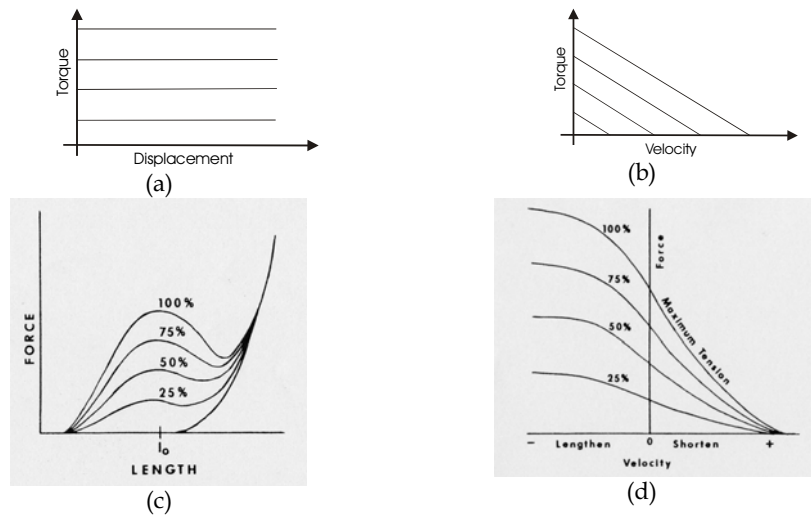


Figure 1. Characterization of artificial and natural actuators based on 'effort-flow' and on 'effort-displacement' relations (inspired from (Hannaford & Winters, 1990), (a) and (b) Case of the DC motor, (c) and (d) Case of the skeletal muscle (from (Winter, 1969))

2.3 The artificial muscle as any material structure exhibiting a reversible shape change by a chemical or a physical agent

a. Contractile polymer gels

Aqueous solutions of polymeric acids considered by Kühn & Katchalsky are a particular case of solvent-swollen crosslinked polymer gels that respond to changes in temperature, pH, solvent, illumination, or other physical or chemical stimuli. The choice of pH or a solvent as a stimulus applied to typically polyacrylamide (PAM), polyvinyl alcohol-polyacrylic acid (PVA-PAA) or polyacrylonitrile (PAN) fibers or strips (all commercially available) is particularly interesting due to the facility of producing the pH variation by adding of acid or alkali solutions (typically through the control of the hydrogen ion by means of chemical reactions with HCl and NaOH) or the facility of using cheap and readily available solvent like acetone. Parallel to a better understanding of gels physics it has been possible to develop micro sized gel fibers contracting in seconds and even tenths of seconds and to increase both the force produced by gel fibers as resistance fibers (some of them can support loads of about 100 N/cm² approximately equal to that of human muscle. M. Suzuki, for example, has developed in the nineties a 12 cm length artificial muscle model by PVA-PAA-PA1Am gel film of 50µm thickness able to raise a ball of 2 g from a lower equilibrium position to an upper equilibrium position within 10 s (Suzuki, 1991). Although in these conditions, the maximum power density obtained with gel fibres can indeed be estimated as being close to that of skeletal muscle, it still appears difficult to apply this technology to the actuation of human-sized robot limbs. However, later we will analyse the interesting attempt made at the MIT in the 1990s to promote 'pH muscle'

The relative slowness of the complete contraction-elongation cycle of ion sensitive gel fibre is mainly due to the relative slowness of the ion sensitive mechanism itself. For this reason, other stimulus modes were considered, in particular heat transfer since heat diffusion is better than ion diffusion. Thermo-responsive polymer hydrogels are already used in drug release, and the study of tensile properties of thermo-responsive gels have been clearly established (Kim *et al.*, 2005). In the field of artificial muscles, this stimulus mode seems, however, to have been more studied in the case of fairly recent polymer types - in particular liquid crystal polymers - assumed to respond quicker than gel fibres. In any case, for gels as for other materials, thermal response always appears quicker in the heating than in the cooling phase (we will not deal in this chapter with the promising technology of liquid crystal polymers able to respond to a luminous flash in less than a second, but as noted by De Gennes (De Gennes *et al.*, 1997) which predicted their contractile possibilities and talked of them as 'semi-quick artificial muscles', the return to low temperatures is slow).

b. Shape memory alloys

Another form of reversible thermal response can be generated by means of thermally activated shape memory alloys based on the 'shape memory effect' discovered at the beginning of the 1950s. Among available materials nickel-titanium alloys (NiTi or nitinol) are particularly interesting for actuation use because of their low cost and electrical resistivity which heats the material by passing an electrical current. Due to this Joule heating, NiTi contraction and expansion cycles can occur based on transformations from martensitic to austenitic phases. Typically a shape is 'memorized' at a high temperature (600°) placing the nitinol in austenite phase. On decreasing the temperature, the material reverts to the martensite phase. When an external stress is applied, and as the material is reheated at a lower temperature the material, because of the instability of the martensite

phase at these temperatures, returns to its well-defined high-temperature shape. In the shape memory effect, the material exhibits residual strains that can be used to generate a linear displacement. The possibility of developing NiTi fibres exhibiting both very high strain rates (300%/s) and very high stress (200 MPa) (Hunter & Lafontaine, 1992) naturally interested the designers of new robot actuators. Safak and Adam (Safak & Adams, 2002), for example, developed a lobster robot actuated by antagonistic nitinol artificial muscle pairs. Figure 2 shows an interesting application of nitinol to miniature humanoid robotics: the five fingers of the miniature robot hand can react at 0.2 s constant times to grasp light objects (Maeno & Hino, 2006). Besides the current need for small sizes to obtain quick responses, the main drawback of nitinol-type artificial muscles is its limited maximum strain: about 5%, which is very far from the natural skeletal muscle's 40 %. Moreover, the life cycle becomes more limited as the strain increases.



Figure 2. Miniature robot-hand actuated by Nitinol shape memory alloy artificial muscles controlled by heat transfer (from (Maeno & Hino, 2006))

The shape memory effect can be controlled not only by heat transfer, but also by means of an electric field which offers the advantage of being an easier control parameter. Ferromagnetic shape memory actuators have been largely studied and commercial versions exist able to develop maximum strains of 10%, typical strain rates of 10 000%/s and typical stresses of 1 MPa to 9 MPa (Madden *et al.*, 2004). However, the requirement of high intensity magnets is a major drawback for human-sized robot-limbs. Finally, as for thermally activated shape memory alloys, the difficulty of accurately controlling muscle contraction is not a good point for applications to robotics. The same difficulty occurs with recent conducting shape memory composites.

c. Electroactive polymers : From ionic polymer gels to ionic polymer-metal composites

It was demonstrated in the 1960s that the swelling/shrinking of ionic polymeric gels generated by pH variation can also be obtained electrically. When an electric field is applied to a strip of PAM, PAMPS or PAA-PVA, for example suspended in a surfactant solution, the gel shows significant and quick bending towards the anode. According to Segalman and others (Segalman *et al.*, 1992) this is caused by the migration of unbound counter ions in the gel, and the impingement of solvent ions at its surface which produce the strip bending. The reversible phenomenon has been applied to the design of chemomechanical mobile systems such as Shahinpoor's swimming robot (Shahinpoor, 1992) or to polymer gel fingers (Hirose *et al.*, 1992). However, the application to linear actuators appears disappointing, as recently reported by Choe and Kin who studied polyacrylonitrile linear actuators (Choe & Kim, 2006): the tested fibre is a 10 mm long 'single strand' consisting of multiple filaments

(about 2000) whose diameter in the contracted state is about 0.5 mm, and 1.2 mm in the elongated state; experimentally the fibre can produce a 0.1 N maximum force in both pH activation and electric activation cases, but while this static tension is generated in fewer than 10 s with a 1M HCl acid solution, it takes approximately 10 min for the same result with a 5V electric field..

If the electrostatic approach of ionic polymer-gel based linear actuators is unsatisfactory, it can be asked if more interesting results could be obtained using conductive polymers. Efforts to develop conductive polymer artificial muscles can be viewed as the search for associating polymer mechanical properties with the ease of electric field control. In general, conductive polymers are electronically conducting organic materials featuring conjugated structures, able to undergo dimensional changes in response to changes in the oxidation state. Polyaniline, trans-polyacetylene and polypyrrole are three examples of such employed structures. When, for example, a conducting polypyrrole film is electrochemically oxidized, positive charges are generated along the chains and hydrated counter ions from the solution are forced to penetrate into the polymer in order to compensate for these positive charges. This induces an opening of the polymer structure and an increase in its volume. A reduction in polymer induces the reverse effect: positive charges are eliminated due the injected electrons: the polymer recovers its neutral state and the film volume decreases. This swelling-shrinking property can be applied to the design of artificial muscles such as the bilayer structure, consisting of a conducting and flexible polypyrrole layer adhering to an elastic and non-conducting film, proposed by Otero and Sansinena (Otero & Sansinena, 1995), or the monolayer structure by assembling two different polypyrroles (Ochoteco *et al.*, 2006). The resulting actuator is a bending type reacting in an aqueous solution under low intensity current variation. According to Madden & others a maximum 12% strain and a maximum 12%/s strain rate can be expected (Madden *et al.*, 2004), consequently quicker than electrically-activated ionic polymer gels; however, conductive polymers suffer from the same major drawback as ionic polymer actuators: they need to be contained in a solvent bath.

The class of Ionic Polymer-Metal Composites (IPMC) can thus appear as a successful attempt to maintain the ionic polymer contractile principle without the need for a solvent bath. An IPMC is an ionic polymer in a composite form with a conductive metallic medium such as platinum. The nafion developed by DuPont de Nemours & Co. is generally used as a cation exchange thin membrane with metal electrode plated on both faces. Recently, liquid nafion has been used to manufacture IPMC strips in various thicknesses (Kim & Shahinpoor, 2002). Although they operate best in a humid environment, they can be designed as self-contained encapsulated actuators to operate in various environments. As theorized by Shahinpoor (Shahinpoor, 2002) an electric field induces a change in ion concentration which attracts water molecules to one side of the polymer. Non-uniform distribution of water produces a swelling on one side of the actuator, while the other side contracts. The bending movement of the strips toward the anode is obtained at a low voltage (typically 2V), and increases for higher voltages (typically up to 10 V) with a reaction speed between μ s and s. Because the IPMC does not produce linear actuation, except in the case of fish-type mobile robots, its application to robotics is limited to gripping mechanisms able to lift a few grams (Shahinpoor *et al.*, 1998). The low stress generated by IPMC - 10 to 30 MPa (Shahinpoor *et al.* 1998) - is another drawback of this type of artificial muscle.

d. Electroactive polymers : Dielectric elastomers

Dielectric elastomer actuator technology appeared in the middle of the 1990s. As will be analysed in paragraph 3, this technology is considered as one of the most promising for developing artificial muscles adapted to robot-limbs (Bar-Cohen, 2004). As in IPMC technology, polymer shape change can be induced in dry environments, but at the expense of higher stresses. A dielectric elastomer actuator can be considered as a compliant capacitor inducing a stress when the capacitor is charged. According to Maxwell's theory applied to a constant volume elastomer, stress p and strains S_X , S_Y , S_Z of the dielectric elastomer (assuming small, e.g. < 10%), respectively, in X, Y and Z directions, as illustrated in Figures 3.a and 3.b, can be written as follows (Pelrine *et al.*, 1998), (Pelrine *et al.*, 2002) :

$$\begin{cases} p = ee_0 (V/t^2) \\ S_Z = -ee_0 V^2/(Yt^2) \\ (1+S_X)(1+S_Y)(1+S_Z)=1 \end{cases} \quad (1)$$

where e_0 is the permittivity of free space, e the polymer dielectric constant, V applied voltage, t polymer thickness and Y elasticity modulus of the polymer-electrode composite film. These three equations constitute a simplified model (limited by the assumption of small strains) but which highlights the main parameters participating in the dimensioning of the actuator: polymer dielectric constant e - e is equal to 1 for air - varies between 3 to 10 according to elastomer choice (Mathew *et al.*, 2006) and electric field $E = V/t$ whose value, depending as it does on the elastomer, is an important factor in increasing actuation stress by a factor of 10 to 100 compared to conventional electrostatic actuators. The simplicity and relative efficiency of this model contrasts with the complexity of Flory's model which includes polymer gels swelling, in association with the difficulty of parametrizing the conductive polymers. Performances of dielectric elastomers are particularly interesting in comparison with other artificial muscle technologies as well as with natural muscle (Pelrine *et al.*, 2000) because they can associate high maximum strain, high maximum stress and low response times: in the case of a silicone elastomer, a maximum strain of 32 % can be generated with a maximum stress of 1.36 Mpa, and a time response of some msec. Furthermore, their ability to be configured in many ways is particularly important for robotics. The next paragraph analyses how linear actuators can be derived from this technology. Two main disadvantages, however, have to be highlighted: firstly, dielectric elastomers operate at relatively high voltages (100-4000 V), and secondly, due to their capacity, electrical energy remains after actuation, which in practice requires energy recovery circuits.

e. Elastomer inflatable structures

Physical elastomer properties are also at the origin of another large class of artificial muscles that recent advances in chemo-mechanical actuators have pushed into the background. They are not mentioned in numerous synthesis articles. We propose to call them 'elastomer inflatable structures' because they are based on the unique property of elastomers to support very large strains (up to 600%) without damage. Whereas the dielectric elastomer operation principle is based on the generation of a compression stress, the operation principle of elastomer inflatable structures is based on tensile stress. Thanks to a specific guiding structure, which will be specified in paragraph 3, the stress generated inside the

elastomer inflatable structure is transformed into a linear force able to contract the artificial muscle, as illustrated in Figures 3.c and 3.d.

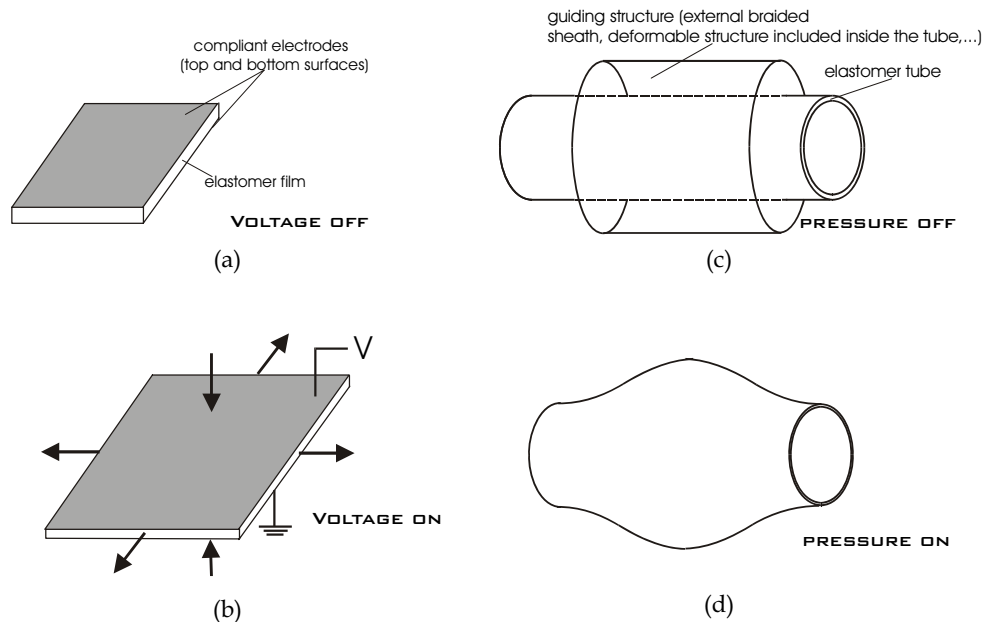


Figure 3. Use of elastomer elasticity in artificial muscle designing, (a) and (b) operation principles of dielectric elastomers (inspired from (Peline *et al.*, 2000)): the elastomer film placed between two compliant electrodes is compressed and expanded when an electric field is applied, (c) and (d) operation principles of contractile inflatable structures: a guiding structure is used to transform lateral stress generated by the pressure into a linear contraction stress

2.4 The artificial muscle as a linear actuator mimicking natural skeletal muscle behaviour: notion of skeletal artificial muscle

Until now the artificial muscle notion has been defined as a reversible contractile device, independently in some ways from any model of the skeletal muscle. In the framework of humanoid robots, based on a general anthropomorphism principle, the notion of artificial muscle can be considered from a more physiological point of view, as a linear actuator mimicking the natural skeletal muscle's static and dynamic behaviour. According to this approach, we suggest considering a fundamental abstract notion of artificial muscle inspired from the macroscopic modelling of skeletal muscle: the derived 'skeletal artificial muscle' notion will be used as a reference biomimetic model independent of the technology considered to achieve the abstract notion. According to the consideration developed by Hannaford & Winter on the actuator notion, referred to in paragraph 1, the skeletal artificial muscle notion can be specified by combining both tension-length and tension-velocity

relationships, *i.e.* static and dynamic models of natural skeletal muscles. The resulting artificial muscle model will be, as a consequence, a phenomenological model which puts aside the microscopic phenomenon at the origin. It is well known that skeletal muscle contraction is induced both by the recruitment of motor units and their firing rate (Ghez, 1991). In a phenomenological muscle model, these factors are set together in the form of a global variable which is traditionally called 'neural activation' and can be defined as a normalized variable u between 0 and 1. In isometric conditions, *i.e.* at a constant length, the typical skeletal muscle tension-length has already been illustrated in Figure 1.c. The static behaviour of a skeletal muscle of initial length l_0 is characterized both by an active tension corresponding to the contraction force produced when $l \leq l_0$, and by a passive tension due to the elasticity of muscle fibres beyond l_0 , as is also well known. If we focus on active tension, skeletal muscle behaves like a linear spring whose stiffness can be controlled by neural activation. The following linear model, illustrated in Figure 4.a, results from these considerations:

$$F_{stat} = uF_{max}\left(1 - \frac{\varepsilon}{\varepsilon_{max}}\right), \quad 0 \leq u \leq 1 \quad \text{and} \quad 0 \leq \varepsilon \leq \varepsilon_{max} \quad (2)$$

where F_{stat} corresponds to the static tension, ε to the muscle contraction ratio - defined as $\varepsilon = (l_0 - l)/l_0$ where l is the current length of the muscle - F_{max} the maximum tension and ε_{max} the maximum contraction ratio. This linear model captures the main static property of skeletal muscle: its 'natural compliance' C proportional to u which physically expresses the existence of a restoring force F_r which brings back the muscle to its equilibrium position when it is deviates of $\delta\varepsilon$; we get:

$$\begin{aligned} F_r &= - \frac{uF_{max}}{\varepsilon_{max}} \delta\varepsilon \\ \Rightarrow C &= + \frac{\varepsilon_{max}}{uF_{max}} \end{aligned} \quad (3)$$

In our opinion the 'natural compliance' factor - or its inverse, the stiffness - has a greater importance than the 'young modulus' generally considered in artificial muscle technology, as defined in paragraph 1: young modulus reflects material characteristics, while compliance actuator characteristics.

Equation (2) is a purely static model; when applied to a dynamic contraction of the muscle, this muscle tension model results in purely oscillating behaviour. In order to take into account the natural damping of the natural muscle, it is consequently necessary to add a supplementary term to the static tension model, we note $F_{damp}(u, \varepsilon, \dot{\varepsilon})$ and which can depend on $\dot{\varepsilon}$ only if a pure viscous damping is assumed or on both u , ε and $\dot{\varepsilon}$ if the damping is associated to more complex friction phenomena:

$$F = uF_{max}\left(1 - \frac{\varepsilon}{\varepsilon_{max}}\right) - F_{damp}(u, \varepsilon, \dot{\varepsilon}), \quad 0 \leq u \leq 1 \quad \text{and} \quad 0 \leq \varepsilon \leq \varepsilon_{max} \quad (4)$$

How to define this damping function? No simple direct model can be derived from skeletal muscular physiology. However, Hill's model provides a particularly powerful and elegant indirect model. The Hill model is generally presented as the relationship between muscle shortening velocity V and its corresponding tension, denoted for reasons to be explained later, F_{Hill} , as follows :

$$(F_{Hill} + a)V = (F_0 - F_{Hill})b \quad (5)$$

where F_0 is the isometric contraction force at zero contraction ratio in given stimulation conditions - according to our static : $F_0 = uF_{max}$ - a is a constant having the dimensions of a force and b a constant having the dimensions of a velocity. Ratio (a/F_0) is typically between 0.2 and 0.3 which gives Hill's equation curve a hyperbola shape, as illustrated in Figure 4.b. Let us recall that the Hill equation was established in a specific isotonic quick-release test invented by Hill in his famous 1938 paper (Hill, 1938): during the almost constant speed phase of the quick-release in 'after-load' mode, F_{Hill} corresponds to the load driven by the muscle and captures the dynamic force produced by the muscle, including its damping component. As a consequence, the quick-release test applied to any artificial muscle allows an appreciation of its dynamic damping in comparison with that of skeletal muscle. Beyond the purely biomimetic argument, we have recently tried to show the interest for an artificial muscle to be in accordance with the Hill model in order to benefit from the natural load variation adaptivity which seems to be an integral part of this dynamic model (Tondou & Diaz, 2006).

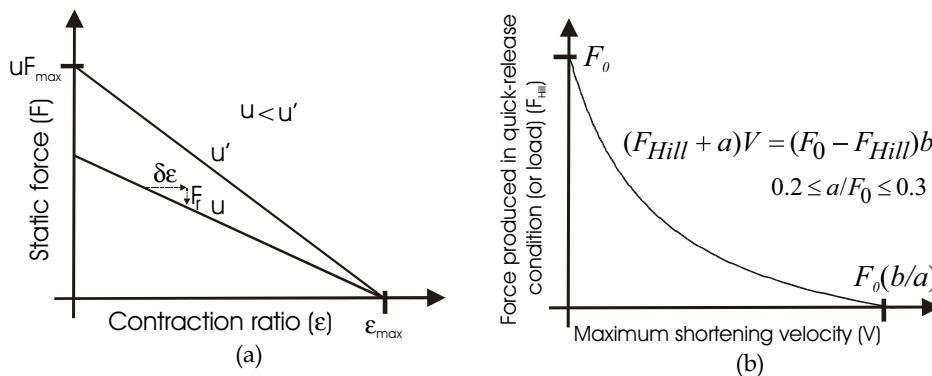


Figure 4. Physical model of skeletal artificial muscle based on both Force-Displacement and Force-Velocity curves of the natural muscle: a linear static force-displacement whose slope is proportional to the control variable is considered (a) with an additive damping term in such a way that the artificial muscle can be dynamically in accordance with (b) the fundamental Hill curve

Among all the technologies reported at the beginning of the paragraph, only the ones able to be adapted to this restrictive notion of skeletal artificial muscle would really be suitable for humanoid robotics. For example, the technology of certain conductive polymers essentially

leads to bending artificial muscle difficultly applicable to linear actuators. Let us now analyse some attempts at artificial skeletal muscle.

3. Which artificial muscle technology for anthropomorphic limbs?

The three technologies that we present have all led to artificial muscles of skeletal muscle size; they illustrate the present diversity of the stimulus mode: chemical for pH muscle, electrical for roll actuator and fluidic for pneumatic muscles. It could be said that because nobody is currently able to mimic the complex sequence of biomechanical microscopic phenomena of natural skeletal muscle, the energy choice question of artificial muscle remains open.

3.1 pH-sensitive muscles and Solvent-sensitive muscles

The possibility of preparing relatively easily gel contractile fibres from commercially available polyacrylonitrile (PAN) or polyvinylalcohol (PVA) filaments led several researchers in the 1980s and 1990s to develop skeletal artificial muscles on roughly the same principle: a set of gel fibres is placed inside a rigid cell in which the circulation of two solutions alternatively generating the swelling and shrinking of the gel fibres is made easy. The resulting volume change can be used to pull and push a 'tendon'. Figure 5 illustrates an interesting prototype developed at MIT in the frame work of the *Artificial Muscle Project* by Brock and others (Brock *et al.*, 1994): acid and base are alternatively delivered to PAN gel fibre bundles through an array of perforated Teflon tubes

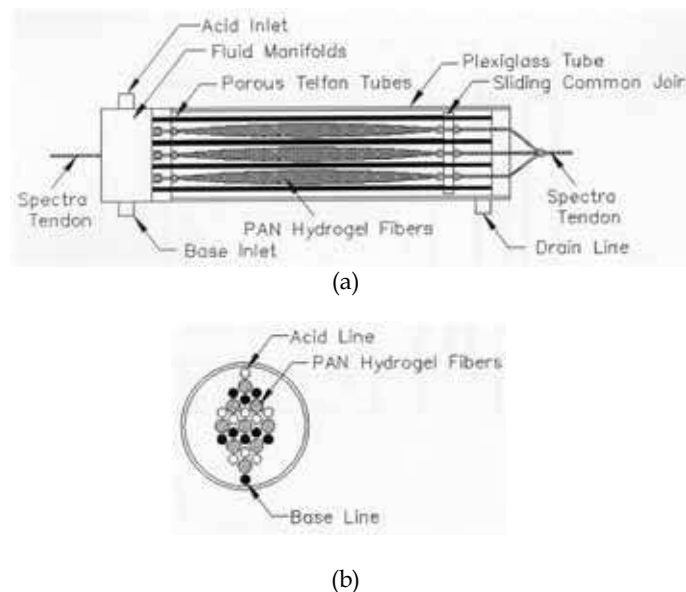


Figure 5. Cylindrical polymer gel 'pH muscle' using PAN gel fibers, (a) Gel bundles cell irrigated by perforated Teflon tubes connected to a common fluid manifold allowing alternatively acid (HCl) and base (NaOH) circulation, (b) Arrangement of gel fiber bundles, acid and base conduits in cross section (from (Brock *et al.*, 1994))

The 'pH muscle' is used in antagonism with a spring and a modular pulley system: using a simple bang-bang control a rotation from 0 to 70° is generated in approximately 45 seconds with a maximum tension of 0.35 N. The confrontation of this experimental device resulting from Caldwell's experiments leads to a better understanding of the difficulty of adapting this technology to powerful skeletal artificial muscles. The 'solvent-sensitive muscle' cell imagined by Caldwell (Caldwell & Taylor, 1990), (Caldwell *et al.*, 2000) is made of parallel PAA/PVA strips inside a container where an acetone and 4 molar NaCl solution can alternatively circulate. Originally (Caldwell & Taylor, 1990) 100 µm thick strips were used, but more recently it was shown (Caldwell *et al.*, 2000) that it was possible to optimize the thickness choice so as to highly increase contraction velocity without losing in stress: a 50 µm thickness gives a 43 %/s strain per s - instead of 11%/s for 100 µm strips and close to slow movements of the natural muscle - for a 30N/cm². With a 43% maximum contraction strain, the performances approach slow movements of the natural muscle. Ten strips are used in Caldwell's linear actuator, but the author highlights a major limit of this technology: the difficulty of use on a large scale. Because strain rate and stress evolve inversely with fibre thickness, it is necessary to constitute the artificial muscle of a great number of parallel fibres to generate useful forces for robot arms, thus inducing a discouraging complexity. To our knowledge, this interesting technology has not been considered for robot limb actuation. These attempts, however, highlight a central point for the development of artificial skeletal muscle: an mm² size scale technology is not necessarily efficient for a cm² size scale. The central notion of artificial muscle stress as defined in paragraph 2 can finally appear deceptive: in association with maximum strain and maximum strain rate it leads to making believe that a certain number of technologies are perfectly adapted to mimicking the human skeletal muscle, whereas their adaptation to a human size and weight robot has not been proved.

3.2 Electroelastomer rolls

SRI International (MenloPark, California) which claims the discovery of electroelastomers for new actuation has designed original linear actuators aimed at mimicking the shape and behaviour of natural skeletal muscles. These cylindrical actuators called *elastomer rolls* or simply *rolls* are composed of a central spring around which several layers of dielectric elastomers are rolled, as defined earlier. The actuator is closed by two caps, used as the electric poles between which the functioning high tension is placed. At rest, the central spring is maintained at compression by its surroundings and when tension is applied the compliant dielectric elastomers extend inducing actuator extension. Figure 11.a illustrates this actuator technology: the presented skeletal artificial muscle is 65 mm long with 45 mm active length, its highest strain is about 26% of active length for a 5 Hz response time, its diameter is 1.2 cm; composed of 20 layers of acrylic films generating a maximum force of 15 N. For this prototype there results a maximum stress of about 0.4 MPa significantly lower to the postulated value for dielectric elastomer artificial muscle technology (Pelrine *et al.*, 2000) - 1.36 Mpa as mentioned earlier - and also significantly lower to that of natural skeletal muscle. Although the roll appears to be really more powerful than 'pH muscle' it suffers from the same drawback: the need to add layers - or fibres - in order to amplify its force, consequently bringing about an increase of its section like a greater design complexity : 35 layers are necessary, for example, to increase the muscle maximum force from 15 N to 21 N (Pei *et al.*, 2003) . For this reason, roll technology remains difficult to apply to a direct

actuation of a robot-arm as the beautiful picture of the ‘wrestling match between EAP actuated robotic arms and human’ tends to prove (Figure 11.b) The roll actuators are too big to be placed on the skeleton arm and have been removed to an appendant box. Furthermore, unlike pH muscle (Caldwell *et al.*, 1990) it has not been proved that roll actuators accord with Hill’s model : as mentioned by Pei, ‘...the stiffness of the roll is the sum of the central spring and the rolled polymer films’ and it can be asked if the central restoring spring does not have a negative effect on the Hill-like damping behaviour which polymers seem to favour.

3.3 Pneumatic muscles

The roll actuator uses rubber stress as a force generator; in a complementary way, it can be said that rubber inflation-based actuators generate their contractile tension by means of the exceptional possibilities of rubber strain. McKibben artificial muscle is the most impressive example of this artificial muscle technological approach. After firstly analysing it, we will briefly discuss recent competitive approaches.

a. The McKibben artificial muscle

Without trying to accurately specify its origin which seems to lose itself in the golden age of rubber-industry derived technologies, McKibben muscle is a typical braided artificial muscle technology whose name comes from the American nuclear physicist Joseph L. McKibben who developed its prototype for bracing handicapped hands. As we have tried to justify (Tondu & Lopez, 2000), pneumatic McKibben artificial muscle is like a tyre whose braided sheath is free to open itself in order to allow the inflation of a pressurized inner tube. McKibben muscle is indeed composed of an inner tube surrounded by a double helix braid, characterized by its initial braid angle (α_0) as illustrated in Figure 6.a. If we assume that the inner tube integrally transmits its circumferential stress to the braided sheath, a simplified model of the resulting contractile linear tension can be deduced from the application of a virtual works theorem to an ideal cylinder the radius and length of which evolve according to the braid opening (Tondu & Lopez, 1995). Let us note l_0 as the cylinder initial length - *i.e.* initial active muscle length - and r_0 the cylinder initial radius - *i.e.* initial active muscle radius. The following static tension/strain model results :

$$F_{stat} = (\pi r_0^2) P [a(1-\varepsilon)^2 - b], \quad 0 \leq \varepsilon \leq \varepsilon_{max} \quad \text{with } a = 3/\tan^2 \alpha_0 \text{ and } b = 1/\sin^2 \alpha_0 \quad (6)$$

where F_{stat} is the generated static tension for a control pressure P in function of the contraction ratio ε . This purely geometrical model does not take into account bound effects as force loss due to material effects whose major one is friction braid fibre on braid fibre - more accurate physical models can be found in (Chou & Hannaford, 1996), (Tondu & Lopez, 2000), (Colbrunn *et al.*, 2001), (Davis *et al.*, 2003)). It however captures three main characters which found the static analogy between McKibben muscle and natural skeletal muscle :

1. Static tension is proportional to P which can so play the part of the neural activation u (let us recall that static roll tension is proportional to the electric tension squared due to the equation (1) model);
2. Static tension is proportional to initial muscle section (πr_0^2) which is an important property for scaling robot-limb actuators as already discussed in the case of the two previous technologies;

3. Static tension continuously decreases when contraction ratio ϵ increases making the actuator a true contractile actuator as opposed to roll actuator.

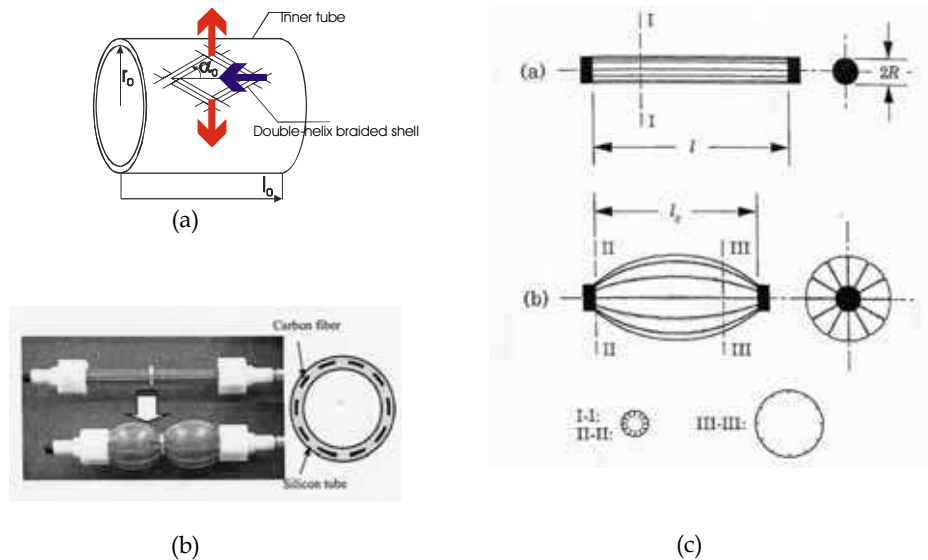


Figure 6. Pneumatic artificial muscle based on the use of a structure guiding the inflation of an elastomer chamber, (a) external structure case: operation principle of the McKibben muscle (see text), (b) internal structure case: longitudinal fibres included in a silicone tube (from (Saga *et al.*, 2005) , see also (Nakamura *et al.*, 2003)) - this type of muscle is as 'old' as McKibben muscle (Matsushita, 1968) - (c) Typical membrane case : pleated muscle (from (Daerden & Lefeber, 2001)) (see text)

The static tension-contraction ratio curve parametrized in P offers a large linear range in a similar way as natural muscle (see Figures 1.c and 7.a). If the maximum contraction ratio, typically between 25% and 35% depending on the pressure, is relatively high in comparison with other artificial muscle technologies, it is still below the typical 40% maximum ratio of skeletal muscle (which even reaches 50% due to passive tension of skeletal muscle *in vivo*). Furthermore, if the fibre-on fibre-friction acting during muscle functioning necessarily induces a problematical dry and kinetic friction (which we discuss later) we also consider that kinetic friction caused by specific textile braided sheaths - like in rayon - can be responsible for a very good accordance of the dynamic McKibben muscle with Hill's model (Tondu & Diaz, 2006). Figure 7.b illustrates the tension-velocity curve obtained in quick-release conditions, at a constant 2 bar pressure, for a McKibben muscle characterized by initial length $l_0=335$ mm, initial radius $r_0=10$ mm, and of initial braid angle $\alpha_0=23^\circ$. Finally, McKibben artificial muscle appears to have a close analogy with natural skeletal macroscopic static and dynamic behaviour. However, its purely mechanical principle implies an increase in artificial muscle volume while skeletal muscle works at constant volume: McKibben muscle volume typically doubles between its zero-contraction and full-contraction states. The maximum contraction ratio is less than the skeletal muscle 40% typical value, and which cannot be increased by its own passive tension, is a second

limitation for a perfect use of McKibben muscle in the actuation of anthropomorphic robot-limbs. Non-constant volume behaviour – which in fact is a characteristic of all fluidic muscles – induces another problem: fluid consumption of the artificial muscle, which implies a specific energy autonomy difficulty for its application to mobile humanoid robots. It is, however, possible in the case of an anthropomorphic musculature based on an antagonistic principle (see next paragraph) to imagine a recovering of consumed air from one elongated muscle to one contracted muscle but this is technologically delicate to put into work. This is the true paradox of McKibben artificial muscle which we will discuss at paragraph end: a unique phenomenological behaviour in analogy with skeletal muscle associated to a traditional and costly energy power source.

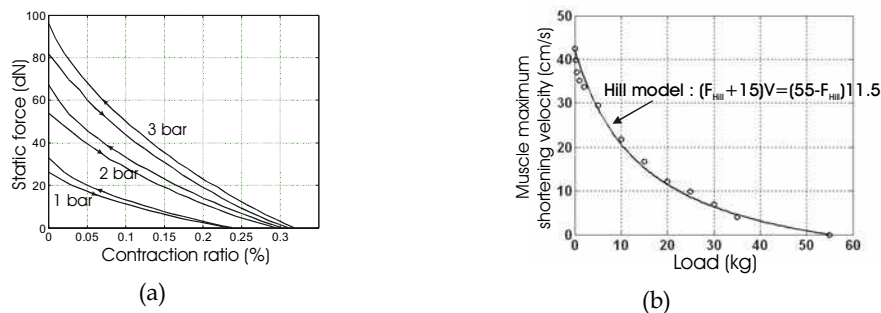


Figure 7. Characteristics of a typical pneumatic McKibben muscle (of initial length $l_0=335$ mm, initial radius $r_0=10$ mm and initial braid angle $\alpha_0=23^\circ$), (a) Tension-Displacement curves at constant pressure 1, 2 and 3 bar, (b) Tension-Velocity curve at constant 2 bar pressure

b. Alternatives to McKibben muscle

McKibben muscle is one of the most known and used pneumatic artificial muscles in Robotics due to its ease of construction and its oblong shape which mimics the natural spindle shape of skeletal muscle, but a large number of other pneumatic muscles have also been developed. It is not easy to classify them; we propose to distinguish three main kinds of pneumatic artificial muscles according to the type of guiding structure of the elastomer inflatable chamber, as illustrated in Figure 6: firstly, the guiding structure is external to the inner tube, as in the case of McKibben muscle; secondly, the guiding structure is included inside the inner tube as considered when longitudinal fibres are embedded in the elastomer (Figure 6.b) and, thirdly, the elastomer chamber is specifically designed to inflate and contract (Figure 6.c). The first type is the easiest to elaborate while the third is the most sophisticated. 'Pleated muscles' developed at Brussels Vrije's University appear at the moment to be one of the most interesting examples of the third type applied to humanoid robotics. It was developed in opposition to McKibben muscle technology. If indeed we ignore the energy problem of pneumatic muscles to focus on the control problem it can be asked if the non-linear character of the actuator is not a too-difficult obstacle to surmount for stable and accurate control. In particular, in the case of an external guiding structure, a friction force generally appears inside the guiding 'mechanical structure' during muscle contraction as between the fibres of a McKibben braided sheath. Due to dry friction, a hysteresis phenomenon results, as illustrated in Figure 7.a for the muscle alone, and in Figure 15.b for the actuator made of two antagonistic muscles. Hysteresis is an undesirable

non-linear effect; as will be analysed in the next paragraph; it is not, in our opinion, the major drawback to a robust control of anthropomorphic limbs actuated by artificial muscles. However, an artificial muscle without hysteresis is obviously a progress in the field; this is the case for the 'pleated muscle' which in the absence of any external braid rubbing against it, avoids the phenomenon. The 'pleated muscle' is a 'cylindrical membrane of high stiffness and high flexibility [folded] along its cylindrical axis like accordion bellows' (Daerden & Lefeber, 2001). When pressurized, inflation happens by unfolding and the artificial muscle shortens and bulges. As illustrated in Figure 12.a, static force characteristics – except for hysteresis – is globally similar to that of McKibben muscle, with a maximum contraction less dependent on pressure. The counterpart of braided sheath elimination is, however, a more complex geometrical shape of the artificial muscle (whereas the McKibben muscle globally keeps a cylindrical shape), pleated muscle offers a contracted complex 'flower'-like shape. What results, firstly, is an increase in the maximum muscle section which, at the same generated stress, can really be higher than that of McKibben muscle and which induces mechanical integration difficulties (see next paragraph). Secondly, muscle shape geometrical complexity implies a more complex mathematical model of muscle inflation. The three fundamental properties exhibited by McKibben muscle also appear present in pleated muscle technology, but in comparison to McKibben muscle no simplified corresponding model can be deduced. The following static model is indeed proposed (Daerden & Lefeber, 2001) :

$$F_{stat} = Pl_0^2 f(\varepsilon, l_0/r_0, a) \quad (7)$$

where l_0 is the initial active length, ε the contraction ratio, r_0 the initial muscle radius, a is a parameter depending on the Young modulus of the membrane elastomer and f is a dimensionless force function for which no closed form seems easy to exhibit (Daerden & Lefeber, 2001). Furthermore, the concordance of the force-velocity curve with Hill's model has not yet been established. Lastly, it is important to note that 'pleated' artificial muscle can have a more limited lifespan than external guiding structure muscles due to the stress imposed to the membrane. As mentioned by Verrelst, the first generation pleated pneumatic artificial muscle has a limited lifespan due to the overlap used to make a cylindrical pleated membrane. The proposed second generation has a 400 000 cycle life but induces a complex mathematical model (Verrelst *et al.*, 2006).

In conclusion, the current field of available artificial muscles for humanoid robots is paradoxical: on the one hand, electroactive polymer-based roll actuators, although very promising, do not yet have sufficient power-to-volume to be embedded in robot-limbs and to actuate them; on the other hand, purely mechanical pneumatic artificial muscles appear to be at present the most biomimetic of muscles, although their energy mode is questionable. This paradoxical situation can, however, be settled in the following manner: even if given up in the future in favour of EAP-based muscles or other bio-inspired artificial muscles, fluidic artificial muscles, in our opinion, tackle the difficult question of mechanical integration and control of anthropomorphic limbs actuated by biomimetic artificial muscles, as considered in the two next paragraphs.

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