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# Wearable Kinesthetic Systems and Emerging Technologies in Actuation for Upperlimb Neurorehabilitation

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*Abstract*—**Kinesthetic and haptic interfaces between humans and machines are currently under development in a truly wearable form, using innovative technologies based on electroactive polymers. The integration of electroactive polymeric materials into wearable garments is becoming a viable mean to confer the garment strain sensing and actuation properties. In this paper, the implementation and testing of fabric-based wearable interfaces for the upper limb endowed with spatially redundant strain sensing are reported. Electroactive polymer actuators, which we are currently investigating, are discussed with emphasis given to their unique capabilities in the phenomenological mimicking of skeletal muscle actuation and control. Finally, current work in preliminary evaluation of prototypes in the field of post-stroke rehabilitation is also briefly presented.**

## I. INTRODUCTION

WEARABLE kinesthetic and haptic interfaces between humans and machines are regarded today as systems capable of supporting a large number of activities in different health-focused disciplines, such as biomonitoring, rehabilitation, telemedicine, teleassistance, ergonomics and sport medicine. Such wearable interfaces are conceived as innovative fabric-based garments [1], integrating sensing and actuation devices. Due to their multifunctional interactivity, enabled by wearable devices that are flexible and conformable to the human body, these kinds of interfaces may be considered as promoters of a higher quality of life and progress in several fields of application. Garments with strain sensing capabilities would enable the tracking of posture and gesture of a subject and would permit analyses of kinematic variables of interest [2]. Likewise, the integration into skin-adherent clothes of actuators represents a potentially useful tool for disciplines like rehabilitation. Actuators may provide enduring mechanical support to lost motor functions (compensation of disabilities) or to their physiotherapeutic restoration. The active support offered by wearable actuators could also favor the improvement of sports training techniques or the prevention of risks related to abnormal stress distributions and overloading.

The long-term goal of our research is to develop a family of truly wearable and bidirectional (i.e. embedding sensing and actuating functions) interfaces. In order to achieve this distant goal, several methodologies and techniques still need to be developed, in terms of sensing (tactile and kinesthetic) and actuation.

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## II. WEARABLE KINESTHETIC SYSTEMS

### *A. Redundant, bioinspired strain sensors arrays*

Kinesthesia, taken literally, means a sense of movement, although current usage of the term often includes the sense of static limb position. Today the term kinesthesia is used in the broadest sense to include the awareness of the positions and movements of the limbs (and other body parts), whether self-generated or externally imposed [3].

In biological systems, the intrinsic noisy, sloppy and poorly selective characteristic of individual mechanoreceptors are masterfully compensated by redundant allocation, powerful peripheral processing and efficient and continuous calibration through supervised and unsupervised learning and training. A truly biomimetic sensing system should possess these features to some extent, not just as a mimicking exercise, but as a result of solid engineering reasoning. Guided by these arguments we are continuing investigating strain sensing fabrics to realize adherent wearable systems with excellent mechanical matching with body tissues. [4]

Smeared conductive elastomer (CE) sensors can be employed to realize wearable sensing systems able to record and detect human postures and gestures [5]. Our main aim has been to produce kinesthetic garments which can be worn for a long time with no discomfort. To obtain this result we have integrated sensor networks made by CE into elastic fabrics used to manufacture the kinesthetic garment collection. In particular, two different kinesthetic garments have been object of extensive development:

- a sensorized shirt for the analysis of the upper limb kinematics, (ULKG); [6]
- a sensorized glove for hand posture and gesture classification [7]

### *B. Kinesthetic Garment Prototypes*

**Upper limb kinesthetic garment**: The ULKG prototype shown in Fig. 2 has been realized by using the mask shown in Fig. 1, and by printing the CE sensors and the connecting tracks on a Lycra/cotton fabric. In order to employ the same CE material both for sensors and for tracks, a dedicated topology was used. In this way, no conventional and cumbersome cables are necessary to connect the sensors to the electronics. The bold black track in Fig. 1 represents the set of sensors connected in series ($S_i$) and it covers the most important joints of the upper limb.

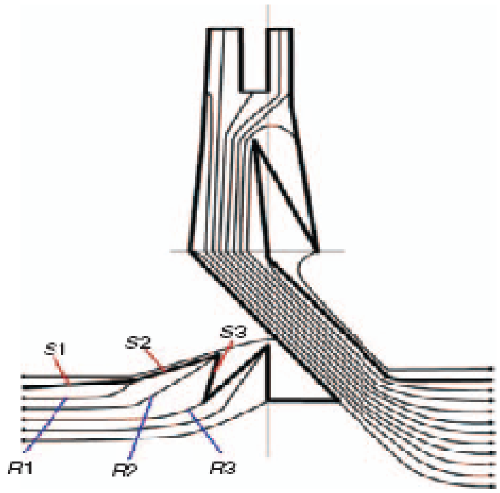


Figure 1. Mask used to realize the ULKG. Sensors are marked as S, connection track are marked by as R.

Figure 2. The ULGK

The thin tracks ($R_i$) represent the connection between sensors and the electronic acquisition system. Since thin tracks are realized by the same CE material, they undergo a significant change in electrical resistance when limbs move. For this reason the front-end electronic unit [5] is designed to compensate for this variation. Practically, a sensor $S_i$ consists of a segment of the bold track between two consecutive thin track intersections. All these remarks on the ULKG also hold for the sensorised glove

**Sensorized glove**: Sensors and wires have been printed on a Lycra/cotton glove (Fig. 3). The bold black track of Fig. 4 represents the set of sensors connected in series and covers the most important joints of the hand. In the same way, a sensor consists of a segment of the bold track between two consecutive thin tracks (Fig. 4).

Figure 3. Worn sensorized glove during a testing phase in comparison with commercial electrogoniometers

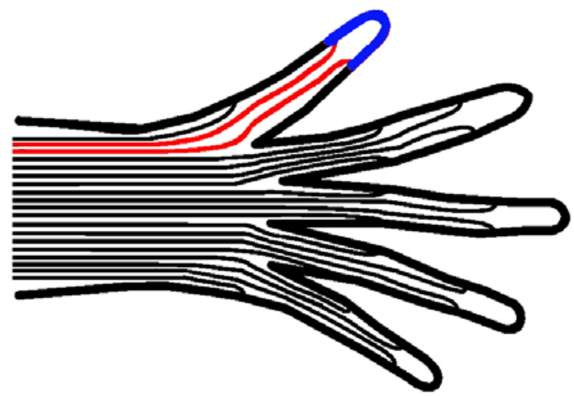

Figure 4. Mask used to realize the sensorised glove. Bold track segment representing sensor series. Thin tracks represent piezoresistive signal acquisition wires.

## III. WEARABLE ACTUATING SYSTEM

One of the main aims in biological motion control theory is to provide a collection of variables defined by the central nervous system which may consent position control [8].

Any control strategy cannot leave out of consideration the characteristics of the muscle model adopted. Certain models of muscle do not ensure the possibility of control of global variables characterizing the status of the chain. One of the possible choices in controlling a biological kinematic chain is to fix some variables related to the robustness of the trajectory that the chain has to follow. In the Equilibrium point theory [9], for example, A. Feldman postulated a direct link between the magnitude of the variation of trajectory and the muscular co-activation. It also established a connection between protagonist-antagonist co-activation and stiffness or compliance of the driven kinematic chain. Obviously, the muscle model which ensure this connection must satisfy some particular requirements, as investigated in this paper. It is proved [10] that muscle-like actuators consenting stiffness or compliance control of the driven system must have a non-linear length-force characteristic.

### *A. Linear Muscle-Like Actuators Made of Dielectric Elastomers*

In this section we analyze the controllability of actuators made of dielectric elastomers (DE), which are assumed to have the following properties

i) they are isotropic and homogeneous

ii) their Young's modulus E is constant under mechanical deformation.

iii) their volume does not vary under mechanical deformation (Poisson's ratio = 0.5).

This class of actuators is becoming important because of their large deformations, sizable forces and reliable operation under electrical drive [11]. Although actual performances of DE actuators are still inadequate to power upper limb prosthesis [12] current efforts and progresses in materials and construction technology make them worth investigating also in conjunction with appropriate control laws. Property (ii) is verified when DE actuators work in their linear range as it happens with pre-stretched, interpenetrated networks [13]. The stress-strain relation for a linearly elastic, isotropic and homogeneous body, subject to axial stress (according x) and when its length is greater than its rest length μ and section area A(x) is given by:

$$df = EA(x)\frac{dx}{x}$$

where $df$ is the infinitesimal force exerted by the DE corresponding to a stretching $dx$. If $V_0$ represents the volume of the fiber, due to the iso-volumetric hypothesis, by integrating on the interval [μ,x] , we have obtain the force exerted by a fiber having length equal to x:

$$f = EV_0\left(\frac{1}{\mu}-\frac{1}{x}\right)He(x-\mu)$$

where He(·) represents the Heaviside function. The previous formula states how the force generated by a DE specimen depends on the actual length of it, $x$, like in the case of a biological muscle included in a kinematic chain whose state depends on antagonist and external loads, and on a parameter μ, which can be used as a "central" control variable. If DE are employed as a dielectric in a capacitor whose plates are placed on surfaces orthogonal to $x$, by assumption (iii), dimension of the DE, included the rest length μ, are controllable by driving the applied electric filed and the DE cylinder can be used as muscle-like actuators (Fig. 5).

### *B. Mimicking Muscle-Like Actuators Via DE Fiber Bundles*

DE fibers are under development [14] and they can conceivably be assembled in a bundle. When many collinear fibers of DE, a set I, having actual length x, are grouped in a bundle, the resultant generated force $F_B$ is :

$$F_B = \sum_{i\in I^*} f_i = \sum_{i\in I^*} EV_0\left(\frac{1}{\mu_i}-\frac{1}{x}\right)u(x-\mu_i)$$

where I* is the set of active fibers, the ones for which

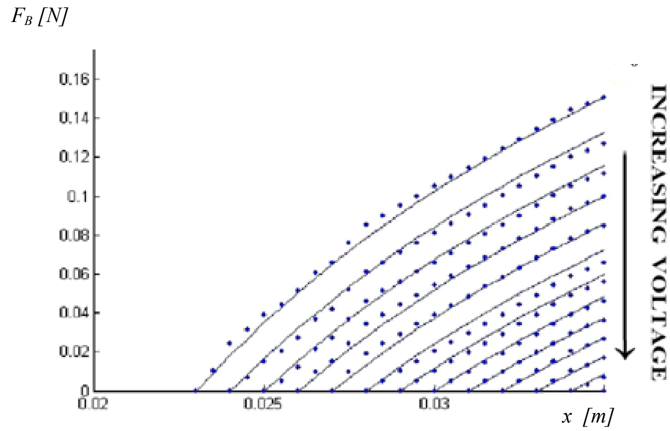


Figure 5. Force exerted by a DE fiber VS. length. Solid line represents the expected behavior for $f$ while e dotted line represents experimental data

$x > \mu_i$ holds. It is worth noting that the global behavior of the bundle can be modified by selecting a suitable activation order for the fibers I*. In the following we show as a DE bundle may operate to approximate Feldman model mechanical behavior [9, 10]. In other terms it must be possible, for a chosen tolerance ε, to define an activation order that satisfies the following relation ($x > \mu_i$, $i \in I^*$),

$$\left\|F-F_B\right\|_{C^1} = \sup_{x\in[\lambda,l]}\left|F-F_B\right| + \sup_{x\in[\lambda,l]}\left|\frac{\partial F}{\partial x}-\frac{\partial F_B}{\partial x}\right| < \varepsilon$$

where λ coincides with the smallest μ and derivatives are computed only on open intervals. The choice of the $C^1$ norm, accounts not only for the intensity of the force but also their derivative with respect to the actual length, i.e the muscle stiffness. This ensures the possibility of obtaining a position and stiffness control for a kinematic chain according to the required non linearity of the muscle model. In [10] further details have been reported on this compliance control strategy for equilibrium points performed by a protagonist-antagonist system. In Fig 6 the results obtained in mimicking a Feldman muscle are represented (solid lines), compared with experimental data obtained by driving a fiber bundle (dotted lines) at different voltage [15].

By exciting all the fibers simultaneously, the characteristic shifts when the global rest length changes (by decreasing $\mu_i$ with $i \in I^*$), as for biological muscles. When dynamics is considered the simple control arguments treated here become much more involved and rate dependent properties of muscles and DE fiber bundles should be properly accounted for. Work is in progress in our laboratory to find methodological and technical solution toward the development and testing of DE fiber bundles and their controllability in dynamic case [16].

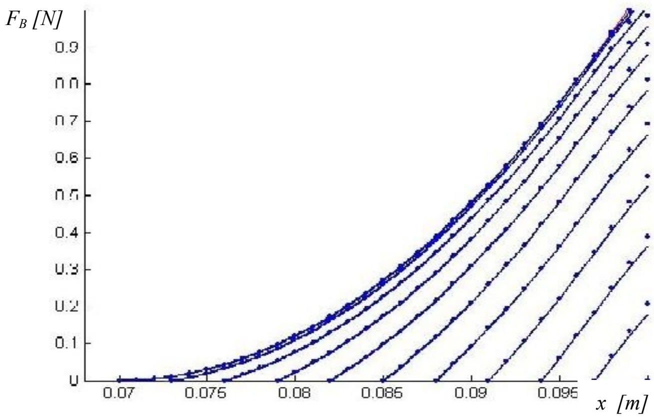


Figure 6. Force $F_B$ exerted by a DE bundle vs. length $x$ for a chosen fiber activation order. Solid line represents the expected behavior for $F_B$ while dotted lines represent experimental data.

## IV. APPLICATIONS IN THE FIELD OF NEUROREHABILITATION

Three applications we envisaged in the field of rehabilitation and they all are under early clinical testing.

These applications are:

- telerehabilitation for post-stroke patients in out – of-hospital environment; [17]
- development of sensing garments for immersive games in rehabilitation [18];
- active splints for hand rehabilitation [19].

While sensorised garments have reached a reasonable level of accuracy and reliability, much more work has to be done to develop muscle-like actuators. Although not yet at reach, wearable DE actuators providing support and biomotion control represent a major challenge which definitely deserve attention and work to enable a new class of revolutionary wearable devices.

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