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Enabling variable-stiffness hand rehabilitation orthoses with dielectric elastomer transducers

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ABSTRACT

Patients affected by motor disorders of the hand and having residual voluntary movements of fingers or wrist can benefit from self-rehabilitation exercises performed with so-called dynamic hand splints. These systems consist of orthoses equipped with elastic cords or springs, which either provide a sustained stretch or resist voluntary movements of fingers or wrist. These simple systems are limited by the impossibility of modulating the mechanical stiffness. This limitation does not allow for customizations and real-time control of the training exercise, which would improve the rehabilitation efficacy. To overcome this limitation, 'active' orthoses equipped with devices that allow for electrical control of the mechanical stiffness are needed. Here, we report on a solution that relies on compact and light-weight electroactive elastic transducers that replace the passive elastic components. We developed a variable-stiffness transducer made of dielectric elastomers, as the most performing types of electromechanically active polymers. The transducer was manufactured with a silicone film and tested with a purposely-developed stiffness control strategy that allowed for electrical modulations of the force–elongation response. Results showed that the proposed new technology is a promising and viable solution to develop electrically controllable dynamic hand orthoses for hand rehabilitation.

1. Introduction

Patients affected by post-trauma, post-stroke or post-surgery motor disorders of the hand and having residual voluntary movements of fingers or wrist can benefit from self-rehabilitation exercises performed with so-called dynamic hand splints [1–3]. They consist of orthoses equipped with mechanically compliant components, namely elastic cords or springs (Fig. 1), which either provide a sustained stretch for a set duration or exert a passive resistance to voluntary movements of one or more fingers (or the wrist). Users can perform rehabilitation exercises 'at any time' and 'anywhere', by simply wearing the orthosis, with no need for any continuous assistance from a therapist.

While such systems are advantageously simple, they are intrinsically limited by the impossibility of modulating the force. In fact, their mechanical stiffness is pre-defined by the properties of the elastic components (cords or springs). This does not allow for customized training and real time control of the rehabilitation exercise, which might be useful features to improve the rehabilitation efficacy.

Aimed at overcoming this limitation, 'active' versions of these orthoses, as opposed to the 'passive' systems of the state of the art, are needed. An 'active orthosis' is meant here as a system that allows for electrical control of its mechanical stiffness.

This work was aimed at developing a new technology enabling such active orthoses. The final target is to endow splints with compact and light-weight variable-stiffness devices, able to enhance the rehabilitation versatility and efficacy.

2. Concept

To conceive the variable-stiffness device, we considered a particular type of electromechanically active (EAP) polymers, as materials able to respond to applied electrical stimuli with changes of size and/or shape [4–6]. EAPs, which are referred to as artificial muscle materials, are highly promising to develop light-weight, deformable, efficient, silent and cost-effective electromechanical transducers for a variety of possible uses [4–6]. Today the EAP

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Fig. 1. Examples of state-of-the-art dynamic hand splints. They consist of orthoses equipped with elastic cords or springs that resist voluntary movements of fingers or wrist (the left and right images are adapted from pictures freely available respectively at www.splinting.com and www.homecraft-rolyan.com).

field is collecting networking efforts for research [7], while the first applications are undergoing transition from academia into commercialization [8,9].

Specifically, we considered the most performing type of EAPs, known as dielectric elastomers (DEs) [10–13]. In its basic elementary form, a DE electromechanical transducer consists of a layer of an insulating elastomer coated with compliant electrode material, so as to have a deformable capacitor. As it is electrically charged, the following electrostatic pressure (Maxwell stress), is exerted by the electrodes onto the intermediate dielectric [13]:

$$p = \varepsilon_r \varepsilon_0 \left(\frac{V}{d}\right)^2 \tag{1}$$

where ε_r is the relative dielectric constant of the elastomer, ε_0 is the dielectric permittivity of vacuum ($\varepsilon_0 = 8.85 \times 10^{-12}$ F/m), V is the potential difference between the electrodes, and d is the thickness of the elastomer film. This electrically induced stress can be used to change the tensional state of the elastomeric structure, so as to electrically control its stiffness. Indeed, DE transducers show an interesting potential for variable-stiffness devices, as described by Pelrine [14]. To elucidate this effect, let us refer to an elementary planar DE transducer, consisting of a cantilever-like elastomeric beam coated with compliant electrodes and subject to a potential difference V and a tensile force F, as sketched in Fig. 2. Pelrine showed that the structure's stiffness K, defined as the derivative of F with respect to the beam length L, is given by the following expression [14]:

$$K = \frac{dF}{dL} = k - bV^2 \tag{2}$$

where

$$b = \frac{\varepsilon_r \varepsilon_0 w^2}{\mathrm{Vol}} \tag{3}$$

Vol and w are the volume and width of the beam, and k is its elastic constant. By assuming a linearly elastic body, k is given by

$$k = \frac{F}{L - L_0} = \frac{\frac{YA(L - L_0)}{L_0}}{L - L_0} = \frac{YA}{L_0} \tag{4}$$

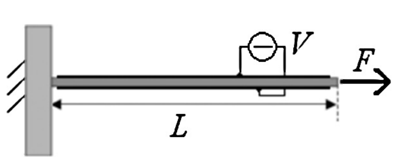

Fig. 2. Reference elementary configuration for a dielectric elastomer transducer shaped as a cantilever-like elastomeric beam coated with compliant electrodes, and subject to both a voltage V and a tensile force F.

where Y is the Young's modulus of the material, A is the cross-section of the elastomer film, and L_0 is its rest length. Eq. (2) clearly shows the possibility of changing the stiffness by controlling the applied voltage. This effect was used in this work to develop an enabling technology for active hand orthoses, as described below.

The concept relies on replacing the passive elastic components of standard splints with electroactive elastic transducers. In particular, we conceived an active dynamic hand orthosis as a system made of the following parts: a structure which supports a DE transducer able to generate electrically controllable forces; a load cell that continuously monitors the force applied by the patient; an external processing unit that controls the device according to the load cell feedback. The elastic transducer is connected to a tendon wire, to be pulled and released by the user for exercise. The processing unit controls the transducer's stiffness, so as to train the patient according to desired rehabilitation plans.

This general concept was already proposed by some of us a few years ago, when a prototype splint-like demonstrator equipped with contractile folded DE actuators was described [15]. At that time, the functionality of the active orthosis system was demonstrated. Nevertheless, the modest electromechanical properties and the large encumbrance of the developed actuators contributed to a significant limitation of both performance and viability for practical usage [15]. That work showed the need for new types of DE transducers based on materials with higher electromechanical performance and configurations with lower encumbrance. To address those needs, in this work we adopted a radically different approach, using both a different material and a different design for the transducer, as described below.

3. System specifications

For typical passive dynamic hand splints, forces and displacements are respectively on the order of magnitude of 1 N (for a single finger) and 1–10 cm [1–3]. The force requirement maps into a specification for the cross-section of the device: greater sections deliver higher forces. On the other hand, greater sections mean bulkier and heavier devices, so that a trade-off is always required. In this work, a target active force (i.e. an electrically achievable force) of at least 0.5 N was fixed as an acceptable trade off. Regarding the displacement, we considered an upper limit of 2 cm. The maximum encumbrance of the transducer was limited to a total length and width of respectively 25 and 5 cm (including the active and passive parts), as a choice consistent with state-of-the-art systems (Fig. 1).

4. Materials and methods

4.1. Dielectric elastomer film and compliant electrodes

As a dielectric elastomer layer we used a commercial silicone film (Danfoss PolyPower, Denmark) that comes with surface corrugation (5 μm height and 10 μm period) and metallization (100 nm-thick silver layer). The layer consisted of two half-layers coupled together, as shown in Fig. 3.

Each half-layer had a thickness of about 33 μm and was corrugated on one side only. The choice for this film was motivated by the intention of maximizing the uniaxial performance of the DE transducer, as it is actually allowed by the combination of corrugation with metal electrodes [16]. Indeed, the film is compliant along one direction, while it is stiff along the other (Fig. 3). This is advantageous for the intended application, as in a splint the elastic components have to work along one direction only.

4.2. Transducer design

The transducer was conceived as a stack of multiple rectangular strips of the film described above. Fig. 4 presents a schematic