MODELING OF ARTIFICIAL MUSCLE MADE OF A FIBER-REINFORCED CONDUCTING POLYMER FOR BIOMECHANICAL APPLICATIONS

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Introduction

Conducting polymer-based artificial muscle (CPAM) gained popularity in biomechanical applications, like an exoskeleton, drug delivery, cell biology and biomedicine due to their promising low voltage (< 5 V) driven input as well as due to high output force and work density compared to natural muscle [1,2]. The existing models of CPAM use linear stress-strain relations [3]. However, these polymers tend to show nonlinear behavior [4]. Additionally, fiber reinforcement can improve the electrical and mechanical properties of the CPAM [5].

This paper develops a coupled electro-chemo-mechanical model for a fiber-reinforced tubular conducting polymer-based artificial muscle (FTCPAM) using a continuum-mechanics approach. The model predicts the nonlinear response of the FTCPAM. Moreover, the change in fiber properties depicts a transition from swelling to deswelling of the actuator for the same applied voltage.

Methods

Consider the FTCPAM as a working electrode in a general electrochemical cell setup. FTCPAM shows an axial elongation or contraction due to ion diffusion through them with an application of voltage. A coupled electro-chemo-mechanical model for FTCPAM is developed. The model can be represented by three major sub-domains, i.e., electrochemical, coupling, and mechanical. The electrochemical sub-domain represents the total charge stored in the actuator for an applied voltage. An electrical circuit analogy illustrates the redox reaction, which results in ion diffusion from the electrolyte to the FTCPAM. [3]. In this work, solving the diffusion equation in cylindrical coordinates derives the relation between the output current and input voltage in terms of modified Bessel function, electrical, and geometrical parameters [6]. The steady-state charge stored in the actuator per unit reference volume for a step voltage V_0 is given by Eq. (1).

$$Q = \frac{CV_0\left(1+\left(\frac{R_2}{2\delta}\right)\left(1-\left(\frac{R_1}{R_2}\right)^2\right)\right)}{2\pi R_2 L} \qquad (1)$$

The coupling domain represents the volume change due to the steady-state charge stored in the actuator, which is given by $v = 1 + \alpha Q$, where α is the coupling coefficient determined through experiments. Finally, the nonlinear axial actuation force is modeled following the invariant-based hyperelasticity approach. The actuator output force is given by Eq. (2).

$$F = 2\pi\left[\left(\frac{4\mu\zeta\sin^2\Phi}{\lambda_3}\right)\left(\frac{v}{\lambda_3}\sin^2\Phi+\lambda_3^2\cos^2\Phi-1\right)\left(\frac{r_2^2}{2}\ln\frac{r_2}{r_1}-\frac{r_2^2-r_1^2}{4}\right)+\frac{\mu}{v}\left(\lambda_3^2-\frac{v}{\lambda_3}+4\zeta\lambda_3^2\cos^2\Phi\left(\frac{v}{\lambda_3}\sin^2\Phi+\lambda_3^2\cos^2\Phi-1\right)\right)\right] \qquad (2)$$

Results

Fig. 1 shows the effect of fiber properties on the free strain (F = 0) of FTCPAM. The dimensions used in the simulation are similar to [5]. Fig. 2 shows a comparison between the existing experimental data and our model.

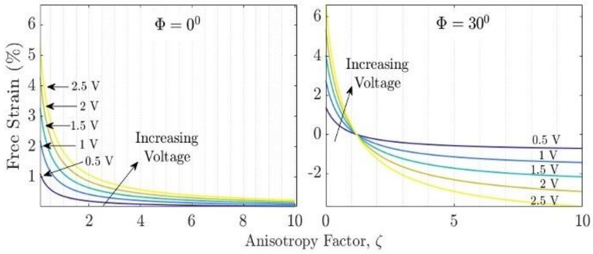

Figure 1: Effect of fiber angle and anisotropy factor on the free strain of FTCPAM.

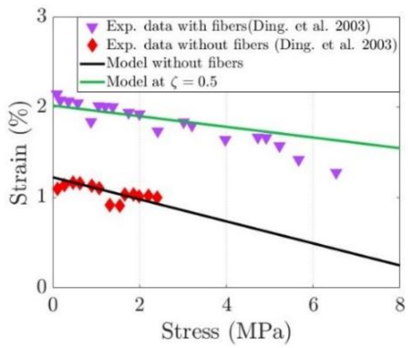

Figure 2: Comparison of the developed model with the experimental data at 2.5 V [5].

Discussion

Fig. 1 shows a transition in response of the actuator at a certain fiber angle and anisotropy factor. FTCPAM starts contracting instead of elongating for the same applied positive DC voltage. This axial deformation of the artificial muscle can be used to develop surgical tools for minimally invasive surgery, integrated with the exoskeleton by tuning the fiber properties.

References

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