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A Feasibility Study of a Novel Piezo MEMS Tweezer for Soft Materials Characterization

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Received: 1 May 2019; Accepted: 28 May 2019; Published: 2 June 2019



Abstract: The opportunity to know the status of a soft tissue (ST) in situ can be very useful for microsurgery or early diagnosis. Since normal and diseased tissues have different mechanical characteristics, many systems have been developed to carry out such measurements locally. Among them, MEMS tweezers are very relevant for their efficiency and relative simplicity compared to the other systems. In this paper a novel piezoelectric MEMS tweezer for soft materials analysis and characterization is presented. A theoretical approach has developed in order to carry out the values of the stiffness, the equivalent Young's modulus, and the viscous damping coefficients of the analyzed samples. The method has been validated by using both Finite Element Analysis and data from the literature.

Keywords: MEMS; tweezer; piezoelectric; soft tissue; microsurgery

1. Introduction

Early diagnosis is crucial to prevent disease progressions [1]. The correlation between the mechanical characteristics of a soft tissue and its status has been highlighted by several studies in many cell types like cancer cells, epithelial cells and laminopathies associated with diseases of the nuclear membrane [2]. In cancer cells, for example, the malignant transformation influences the mechanical properties by disruption and/or reorganization of cytoskeleton [3]. The stiffness, the elastic modulus, and the dynamic viscosity are among the most commonly used mechanical quantities to evaluate the state of a soft tissue. However the possibility to identify a diseased soft tissue in situ, at the microscale, can increase significantly the possibility of the cure.

Different techniques are used to perform such measurements as micropipette aspiration [4], magnetic bead twisting [5], atomic force microscopy [6], optical tweezers [7] or mechanical tweezers [8]. In recent years, researchers focused on the developing of micro-electro-mechanical systems (MEMS) tweezers because of their efficiency and relative simplicity with respect to other systems [9]. New gripping tweezers have been built taking cue from the kinematics of articulated mechanisms [10] as the property of parallelograms [11–15]. More complex motion, with an increasing of the number of degrees of freedom, can be obtained adopting the concept of lumped compliance. In fact, concentrated flexibility has been used in flexure hinges in several investigations [16–20]. In order to improve the displacement accuracy and to lower the stress levels in the flexible elements a new conjugate surfaces flexure hinge (CSFH) flexure has been proposed [21–25] and optimized [26,27]. The system has shown promising results in terms of of versatility and applicability to different types of tissue [28]. Thanks to their high level of miniaturization, MEMS-Technology based tweezers can be employed also in minimal invasive [29–31] or gastrointestinal surgery [32], and more generally speaking, in endoluminal surgery, for example, in TEM [33–35].

The actuation of the tweezer can be obtained by different systems, such as linear electrostatic actuators [36,37], rotatory electrostatic actuators [38,39] or electrothermal actuation [40–46]. However the advent of smart materials in the last decades has greatly increased the possibilities of development of new smart structures [47–49] with the ability, not possible for the traditional systems, to adapt to external conditions variations. Because of their speed of response, low power consumption and high operating bandwidth the piezoelectric materials are, among the smart materials, the most promising ones for active vibration control [50–53] and MEMS applications [54]. However the research in the latter field mainly focuses on the energy harvesting [55–58] or MEMS tweezers for manipulating micro-objects and microassembly [59–61].

This paper presents the design of a novel piezoMEMS tweezer for the analysis and characterization of soft materials. Each jaw can be built as a sandwich composite beam and activated by an electric field.

The tweezer is supposed to be force controlled because the piezoelectric materials produce a stress proportional to the applied electrical field. Furthermore, a sensor is supposed to be integrated into the structure for displacement control. This gives rise to improve the analysis of the cell properties. In fact, the displacement-controlled actuator is able to identify the beginning of a rupture during straining or softening, while a force-controlled system maintains a constant force regardless of the required displacement [62]. The action of the piezoelectric material has been modeled by the Pin Force Model [63].

By applying symmetric electric fields to the composite jaws, they will bend in opposite directions allowing the gripping of the soft tissue.

A new mathematical model to measure the stiffness, the equivalent Young's modulus, and the viscous damping coefficient of the soft tissue has been developed. The model has been tested on three different soft test specimens and the results were in good agreement with those obtained by COMSOL finite elements code.

2. The Adopted Piezo-Mems Microgripper

The purpose of this paper is to develop a theoretical model of a piezo-MEMS microgripper, which can be used to characterize a grasped sample tissue. Therefore, the actual fabrication process of this instrument will not be herein considered. However, for the sake of completeness, a selection of possible materials and technological processes is briefly recalled.

In the last decades, several actuation methods have been proposed to induce motion in MEMS devices such as electrostatic, thermal and piezoelectric. The electrostatic devices are widely adopted but the piezoelectric MEMS's offer some attractive advantages: lower power consumption, broader bandwidth and approximately ten times lower voltages to obtain the same given displacement [64]. Furthermore the piezoelectric materials can be manufactured using the same MEMS conventional technologies and for this reason these materials have been preferred to develop piezo-MEMS devices in the last decades. Typical applications include vibration energy harvesters [65,66], resonators [67], capacitors [68], micro sensors/actuators [69,70], micromachined ultrasonic transducer [71], gyroscopic sensors [72], microlens [73,74], 1D and 2D micro-scanners [75,76].

The piezoelectric materials can be gathered into two groups: ferroelectric (lead zirconate titanate, PZT compounds) and non-ferroelectric (ZnO and AlN). The piezoelectric characteristics (piezoelectric coefficient, Q factor, dielectric constant, etc.) rely on the crystalline structure. In fact ZnO and AlN thin-films show wurtzite structure that entails lower piezoelectricity if compared to PZT materials (perovskite structure). Nevertheless ZnO and AlN exhibit large mechanical stiffness, high Q factor and do not require a polarization process so they can be attractive for sensors applications. PZT thin-films provide high piezoelectric properties, lower cost and stability against temperature but require a poling process before using the piezo-MEMS device [77]. The electric field poling direction depends on the functional configuration of the piezo-MEMS.

Usually the piezoelectric MEMS actuators/sensors are based on cantilever structures and the number of active layers identifies their working configuration:

- unimorph (one piezoelectric layer coupled with an inactive structural layer) or
- bimorph (a structural layer sandwiched by two active layers).

When a PZT bimorph bending beam and the transverse piezoelectric effect (d_{31}) are considered, the PZT layers must be poled in opposite directions in order to maximize the bending action. Then the electrodes of the PZT layers in contact with the structural layer share the same electric potential, whereas the outer electrodes share the same opposite sign potential (see Section 3). The design and fabrication process of piezo-MEMS devices have been extensively explored for the above mentioned unimorph and bimorph cantilever configurations using the aforementioned materials:

- AlN [67,68,71];
- ZnO [70]; and
- PZT [66,69,75,78].

Various technologies can be applied to deposit piezoelectric thin-films such as

- pulsed laser deposition (PLD);
- chemical vapour deposition (CVD);
- screen printing;
- sol-gel and
- radio frequency sputtering.

Usually the sol-gel and sputtering methods are the preferred ones both for research and commercial production because they allow the piezoelectric thin-film to be homogeneously deposited on large Si wafers [79]. The beam structural layer can be metal-based or silicon-based and the etching processes could be accomplished relying on conventional techniques (RIE, D-RIE, ECR).

A schematic view of the target piezo-MEMS tweezer is qualitatively depicted in Figure 1. The system can be obtained by using a multilayer wafer that can be built by using the above mentioned techniques. At the end of the process, the whole microsystem is composed of two bimorph beams (a)-(b)-(c) (see Figure 1). The two bimorph beams are supported by the handle layer (d). The specific steps of the process (deposition, etching, exposure, etc.) will depend on the peculiar materials and technology selected for the construction. In the case under study, the system is conceived in such a way that the mask geometry could be quite elementary for any etching step.



Figure 1. A schematic view of the microsystem: outer layers with a same voltage V_0 (**a**); structural material for the beams (**b**); internal layers with voltage V_i (**c**) and structural layer for the handle (**d**).

Another possible layout is represented in Figure 2, where two new layers (e) have been added with respect to the previous example of fabrication. Layers (a), (b), (c), and (d) have the same function as described for the previous layout. The second design is better for the operational aspects because a clamping tooth has been added for each jaw. However, these two teeth are more difficult to obtain

during the process because they require at least two more deposition layers (**e**) and also a more complex series of intermediate etching-deposition steps that depend on the selected materials.



Figure 2. An alternative layout for the microgripper: outer layers with a same voltage V_0 (**a**); structural material for the beams (**b**); internal layers with voltage V_i (**c**) and structural layer for the handle (**d**).

3. Modeling Piezoelectric Actions on the Microbeams

Concerning the piezoelectric actions, many studies [63,80–82] have confirmed that, under certain assumptions (piezoelectric plates are perfectly bonded to the structure, the ratio between their thickness and the thickness of the beam is very low and their inertia and mass negligible with respect to those of the beam) the stresses applied to the beam can be considered as they were concentrated at the piezoelectric plate ends. Moreover, if the electric field to the upper plate is opposite in sign to the one applied to the lower plate, such action is equivalent to a flexural moment (see Figure 3) applied to the end of the beam of magnitude equal to:

$$M_a(t) = \frac{\psi}{6+\psi} E_a c T_a T_b \Lambda(t) \tag{1}$$

with

$$\begin{cases} \Lambda(t) = \frac{d_{31}}{T_a} V(t) \\ \psi = \frac{E_b T_b}{E_a T_a} \end{cases}$$
(2)



Figure 3. Piezoelectric action (PIN force model).

By activating the piezoelectric plates pairs on the two jaws-beams of the tweezer in opposite manner the two moments M_a will have opposite sign and the beams will bend in opposite direction allowing the gripper to grasp the soft tissue, as depicted in Figure 4.



Figure 4. PiezoMEMS grip of the soft tissue.

3.1. Static Model

The value of the stiffness K_{ST} and of the equivalent Young's modulus E_{ST} can be established by a static test with a constant voltage V_0 applied to the piezoelectric plates: $V(t) = V_0$. In this case, the bending of the beams will cause the compression of the soft tissue and consequently a reaction force F_{ST} will be applied from the soft tissue to the beams, through the clamp teeth (see Figure 5).



Figure 5. Scheme for the calculation of the tip displacement.

Considering that the Euler Bernoulli model is appropriate for the present case because of the high length-to-thickness aspect ratio of the beams, the tip deflection:

$$w_L = \frac{M_a L^2}{2E_b I_b} - \frac{F_{ST} L^3}{3E_b I_b} , \qquad (3)$$

is due to a combination of the loads F_{ST} and M_a . Therefore, the first load

$$F_{ST} = \frac{3E_b I_b}{L^3} \left(\frac{M_a L^2}{2E_b I_b} - w_L \right) \tag{4}$$

can be obtained by measuring w_L .

Due to the high value of the axial stiffness of the clamp teeth, the deflection w_L can be considered equal to the axial displacement of the soft tissue (see Figure 5) so that the axial strain

$$\epsilon_{ST} = \frac{2w_L}{L_{ST}} \tag{5}$$

is calculated.

Under the assumption of linear elastic behavior for the soft tissue (ST), the relationship between the normal stress σ_{ST} applied to the cross section of the soft tissue and ϵ_{ST} can be written as:

$$\sigma_{ST} = \frac{F_{ST}}{S_{ST}} = E_{ST} \frac{2w_L}{L_{ST}} \tag{6}$$

from which the equivalent Young's modulus

$$E_{ST} = \frac{F_{ST}L_{ST}}{2S_{ST}w_L} \tag{7}$$

is evaluated. Moreover, the stiffness

$$K_{ST} = \frac{2E_{ST}S_{ST}}{L_{ST}} \tag{8}$$

is calculated given the dimensions of the ST sample.

3.2. Dynamics Model

To determine the value of the viscous damping C_{ST} , a dynamic test has been conceived by applying a variable harmonic excitation voltage $V(t) = V_0 sin(\omega t)$.

By referring to δL_{int} , δL_{a} , δL_{a} and δL_{ST} as the virtual works of internal, inertial, actuator and ST forces, respectively, and to δL_{inm} as the virtual work of the inertial forces of the clamp teeth, the virtual work principle can be written as:

$$\delta L_{int} = \delta L_{in} + \delta L_a + \delta L_{in_m} + \delta L_{ST} \tag{9}$$

where (the quantities over signed by a tilde are virtual quantities):

$$\delta L_{int} = E_b I_b \int_0^L \frac{\partial^2 w}{\partial x^2} \frac{\partial^2 \tilde{w}}{\partial x^2} dx$$

$$\delta L_{in} = -\rho S \int_0^L \frac{\partial^2 w}{\partial t^2} \widetilde{w} dx$$

$$\delta L_a = M_a \frac{\partial \tilde{w}}{\partial x} \Big|_{x=L}$$

$$\delta L_{in_m} = -m \frac{\partial^2 w_L}{\partial t^2} \widetilde{w_L}$$

$$\delta L_{ST} = F_{ST} \widetilde{w_L}$$

(10)

with (see Figure 6):

$$F_{ST} = K_{ST} w_L + C_{ST} \dot{w}_L \tag{11}$$



Figure 6. Scheme for the calculation of the stiffness and viscous damping of the soft tissue.

The viscous damping coefficient C_{ST} can be identified by measuring the amplitude $|w_L|$ in correspondence of the *i*-th vibration mode of the structure that has been excited by means of the piezoelectric elements, where the flexural modes of the beams are obtained as

$$\varphi(x) = B_1 \sin(\lambda x) + B_2 \cos(\lambda x) + B_3 \sinh(\lambda x) + B_4 \cosh(\lambda x)$$
(12)

with

$$\begin{cases} \varphi(0) = 0 \\ \frac{\partial \varphi(x)}{\partial x} \Big|_{x=0} = 0 \\ \frac{\partial^2 \varphi(x)}{\partial x^2} \Big|_{x=L} = 0 \\ EI \frac{\partial^3 \varphi(x)}{\partial x^3} \Big|_{x=L} = K_{ST} \varphi(L) - \omega^2 m \varphi(L) \end{cases}$$
(13)

By substituting Equation (12) in (13), a system of four equations in four unknowns (B_1 , B_2 , B_3 , B_4) is obtained. The eigenfrequencies are obtained setting to zero the determinant of the matrix coefficient (with $\omega^2 = \lambda^4 \frac{EI}{\rho S}$) and then the eigenmodes can be calculated.

The *i*-th flexural mode can be excited by applying a potential function

$$V(t) = V_0 \sin(\omega_i t) \tag{14}$$

where $\varphi_i(x)$ and ω_i are the *i*-th flexural mode and its related frequency, respectively. In these conditions the displacement w(x, t) and the virtual displacement $\tilde{w}(x, t)$ can be written as:

$$\begin{cases} w(x,t) = A_i(t)\varphi_i(x) \\ \widetilde{w}(x,t) = \varphi_i(x) \end{cases}$$
(15)

By substituting Equations (15) and (10) in (9), the following equation is obtained:

$$E_b I_b A_i(t) \int_0^L \frac{\partial^2 \varphi_i(x)}{\partial x^2} dx = -\rho S \ddot{A}_i(t) \int_0^L \varphi_i(x)^2 dx - m A_i(t) \varphi_i(L)^2 + M_a \frac{\partial \varphi_i(x)}{\partial x} \Big|_{x=L} - K_{ST} A_i(t) \varphi_i(L)^2 - C_{ST} \dot{A}_i(t) \varphi_i(L)^2$$
(16)

assuming (see Equation (1)):

$$M = \rho S \int_{0}^{L} \varphi_{i}(x)^{2} dx + m \varphi_{i}(L)^{2}$$

$$C = C_{ST} \varphi_{i}(L)^{2}$$

$$K = E_{b} I_{b} \int_{0}^{L} \frac{\partial^{2} \varphi_{i}(x)}{\partial x^{2}}^{2} dx + K_{ST} \varphi_{i}(L)^{2}$$

$$Q = \frac{\psi}{6+\psi} E_{a} c T_{b} d_{31} \left. \frac{\partial \varphi_{i}(x)}{\partial x} \right|_{x=L} V_{0}$$
(17)

and therefore Equation (16) becomes:

$$M\ddot{A}_{i}(t) + C\dot{A}_{i}(t) + KA_{i}(t) = Q\sin(\omega_{i}t)$$
(18)

By neglecting the transient part (see Equation (15).1), the amplitude of the free end displacement is:

$$\left|w_{L}\right| = A_{i_{f}}\varphi_{i}(L) \tag{19}$$

where

$$A_{i_f} = \frac{Q}{C\omega_i} \,. \tag{20}$$

Finally, the damping coefficient of the soft tissue

$$C_{ST} = \frac{Q}{\left|w_L\right|\omega_i\varphi_i(L)} \tag{21}$$

can be found from Equation (17.2).

4. Results and Discussion

To validate the proposed model, numerical simulations have been done. The dimensions and the material characteristics are summarized in Table 1:

Table 1. Beams, piezoelectric plates and end masses specifications; lengths are expressed in µm.

	Material	Length	Thickness	Width	Young's mod. [GPa]	Density $[kg/m^3]$
Beams	Silicon	750	2	80	170	2329
Piezoel. pl.	AlN	750	$2 * 10^{-2}$	80	350	3300
Clamp teeth	Silicon	36	7.5	80	170	2329

In this work, the FEM results have been chosen as the reference values. Three typical soft tissues (liver, muscle and uterus) of known characteristics [83], have been considered. The value of the beam tip displacement, obtained by the FEM simulations, has been included in the mathematical model to calculate the equivalent Young's modulus and the viscous damping coefficient. The ST sample stiffness can be calculated by means of Equation (8). The values of the Young's modulus and the viscous coefficients reported in the literature have been compared with the ones calculated by the new method. Because of the symmetry of the structure with respect to its mid-plane (see Figure 7) only the upper part has been considered in the simulations.

In Table 2 the results of the static simulations, obtained by the above described procedure, have been reported. It is possible to observe that the model results are in good agreement with the real values with a percentage error less than 6% in all the cases.



Figure 7. Scheme used for the simulations.

Table 2. Comparison between the actual Young's modules and those obtained by the model; E_{mm} is the Young's modulus obtained by the mathematical model.

	E [kPa]	E _{mm} [kPa]	Err %
Liver	10	9.47	5.3%
Muscle	20	19.43	2.8%
Uterus	30	29.64	1.2 %

To obtain the viscous damping coefficient, dynamics simulations are necessary. As described above, the chosen strategy consists in exciting, by the piezoelectric plates, the *i*-th mode of the structure, in order to obtain the amplitude of the free end displacement $|w_L|$ and then in including this in the model.

A comparison between the eigenfrequencies obtained by the COMSOL FEM code and the proposed model is reported in Table 3

Table 3. Comparison between the eigenfrequencies obtained by the COMSOL FEM code and the proposed model.

Liver					
	FEM	Model	Err%		
$f_1[Hz]$	15,523.50	15,856.40	2.14%		
$f_2[Hz]$	30,701.40	29,487.60	3.95%		
$f_3[Hz]$	82,116.70	75,668.50	7.85%		
$f_4[Hz]$	162,877.10	151,620.00	6.91%		
Muscle					
	FEM	Model	Err%		
$f_1[Hz]$	18,768.20	18,686.20	0.44%		
$f_2[Hz]$	35,437.60	34,681.70	2.13%		
$f_3[Hz]$	82,674.80	76,297.90	7.71%		
$f_4[Hz]$	162,995.20	151,781	6.88%		
Uterus					
	FEM	Model	Err%		
$f_1[Hz]$	20,066.80	19,721.70	1.72%		
$f_2[Hz]$	40,190.2	39,660.70	1.32%		
$f_3[Hz]$	83,255.70	77,038.00	7.47%		
$f_4[Hz]$	163,118.00	151,949.00	6.85%		

In the FEM simulations the first mode has been chosen to excite the structure with the values of the electrical potential reported in Table 4.

Table 4. Electrical potential used for the simulations.

	$V_0[V]$
Liver	8
Muscle	5
Uterus	5

By neglecting the initial transient part, the axial displacements for the various soft tissues have been reported in Figure 8.



Figure 8. Axial displacement for the various soft tissues.

Finally, the comparison between the actual viscous damping coefficients and those obtained by the model results have been highlighted in Table 5.

Table 5. Comparison between the effective viscous damping and those obtained by the model.

	C[Ns/m]	$C_{mm}[Ns/m]$	Err%
Liver	$3.07 * 10^{-5}$	$2.94 * 10^{-5}$	4.2%
Muscle	$5.09 * 10^{-5}$	$4.98 * 10^{-5}$	2.2%
Uterus	$7.14 * 10^{-5}$	$7.26 * 10^{-5}$	1.7%

A good agreement between the relative coefficients is also observed with a percentage error always less than 5%.

5. Conclusions

In this paper a novel piezo MEMS tweezer for soft materials characterization has been proposed. The tweezer mechanical structure is compatible with the known fabrication processes. A new mathematical model to calculate the stiffness, the equivalent Young's modulus and the viscous damping coefficient of the soft tissues is suggested. The method has been tested by comparing its results with Finite Element

Analysis based on experimental data from the literature. The two sets of data are in good agreement with a difference less than 6% in all the considered cases.

Author Contributions: F.B. has designed and developed the mathematical model. A.R. and F.B. carried out the numerical simulations. N.P.B. has reviewed the theoretical approach and the compatibility of the structure with some technological fabrication process.

Conflicts of Interest: The authors declare no conflict of interest.

Abbreviations

The following abbreviations are used in this manuscript:

- C damping coefficient
- *C*_{ST} damping coefficient of the soft tissue
- *d*₃₁ piezoelectric coefficient
- *E_a* Young's modulus of the piezoelectric material
- E_b Young's modulus of the beam
- E_{ST} equivalent Young's modulus of the soft tissue
- F_{ST} force applied from the soft tissue to the tweezers
- I_b inertia moment of the beam
- *K* stiffness coefficient
- K_{ST} stiffness coefficient of the soft tissue
- *L* beam length
- *m* mass applied at the end of the tweezer
- *M_a* piezoelectric bending moment
- M mass coefficient
- S_{ST} cross-sectional area of the soft tissue
- ST soft tissue
- *T_a* piezoelectric thickness
- T_b beam thickness
- V electric potential applied to the piezoelectric plates
- w transverse displacement
- w_L transverse displacement of the free end
- \tilde{w} virtual transverse displacement
- $\phi_i(x)$ i-th flexural mode of the beam
- $ho \qquad {
 m mass density of the beam}$
- ω_i natural frequency of the i-th flexural mode

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