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Optimization of piezo-fiber scanning architecture for low voltage/high displacement operation

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ABSTRACT

Piezo-based fiber scanning probes have emerged as low-cost and compact tools for various optical imaging modalities, allowing access to tissue sites that are hard to reach. These instruments exploit scanning of a fiber optic cable via a piezoelectric element, which is driven at the mechanical resonance of the extended fiber piece. However, the dynamics of the piezo-scanning structure is often neglected, resulting in an inefficient electromechanical conversion. This work presents a methodology, together with experimental evidence, to collectively optimize the geometries of the piezo-scanner and the extended fiber optic cable to achieve maximum displacement for a given drive voltage. Our findings suggest that matching the individual resonances of the fiber optics cable and the piezo-scanner alone, leads to optimum electromechanical conversion efficiency. Simulations, circuit model, and experimental results reveal more than x2 improvement in the achieved fiber displacement when piezo and fiber resonances are matched, as opposed to the unmatched (i.e., when piezo element length is varied approximately by ±20% from its optimal value) case. Besides offering lower power consumption for the actuation of the piezo-element, our findings paves the way for safer (electric shock-free) minimally-invasive procedures using the piezo-based fiber scanning probes, which is crucial for patient safety.

1. Introduction

Thanks to their compact geometry and simple architecture, piezo-scanning based fiber probes have been exploited in confocal imaging [1], optical coherence tomography [2], multiphoton imaging [3], and imaging of Raman scattering [4]. Their size advantage enables accessing tissue sites that are difficult to reach, while providing high-resolution images that reveal pathological information. In this regard, piezo-scanning fiber probes have been widely utilized in biological studies such as calcium imaging of the rat cerebellum [5], epithelial imaging [6], monitoring collagen morphology in the cervix [7], minimally invasive imaging of the mouse lung and colon tissues [8], brain imaging in freely moving animals [9]. With their superior form factor, piezo-scanning fiber probes have also been involved in commercial applications, such as compact display systems [10,11].

In a piezo-scanning fiber probe, light is mapped onto the tissue that is exiting the extended optical fiber, which is encircled and actuated by a cylindrical piezoelectric tube. The piezo-tube, which converts applied voltage to mechanical movement, is driven at the mechanical resonance of the extended fiber to create the desired scan pattern. Owing to its four quadrant electrodes surrounding the piezo-tube, various 2D tissue scan patterns (spiral or lissajous patterns) can be achieved for imaging and microsurgery [12,13]. Furthermore, combination of orthogonally placed piezo-sheet cantilevers has been exploited to establish a raster scanning pattern, which cannot be achieved with conventional piezo-tube based fiber scanning [14].

Actuating the piezo-element (tube or cantilever) at the mechanical resonance frequency of the extended fiber optic cable, results in a high fiber tip displacement, for a much lower displacement of the piezo-element free-end. Yet, the dynamics of the piezo-scanning structure is often neglected, i.e. the relationship between the geometry and resonant frequencies of the piezo-element and the extended fiber for optimized electromechanical coupling has not been treated. A high degree of electromechanical coupling not only ensures achieving high Field-of-View (FOV), but also acts as a safeguard towards electric shock-free interventions (for worst-case scenarios, where electrical connections to the endoscopic device gets detached from the piezo-element, or get stripped off, resulting in an electrical contact with the tissue) with the piezo-scanning fiber probe. Besides safety and power consumption issues, lowering

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the actuation voltages will further lower the cost of the electronic driver circuitry that accompany the probe.

2. System description and modeling

Fig. 1(a) illustrates the basic concept of a simplified piezo-scanning unit, consisting of a piezo cantilever and an extended optical fiber that is capable of 1-D scanning of the light beam. It is worthy to note that a 1-D treatment of a cascaded piezo-cantilever and extended fiber system is directly applicable to practical miniaturized imaging scenarios, mentioned in the introduction part, exploiting both piezo-cylinders and two orthogonally placed piezo-cantilevers: i) For a piezo-tube actuator (cylindrical piezo-element) the 2D scan pattern can be decomposed into two orthogonal components, each of which can be modeled with a resonator system as explained below, and ii) for imaging probes utilizing two-orthogonally placed cantilevers, one axis (fast-axis) operates in resonance while the other is driven off-resonance (slow-axis). The introduced model in Fig. 1(a) would then be relevant to the fast-axis operation. In this regard, our aim is to form the basis through introducing the mechanical models of each orthogonal direction.

Fig. 1(b) shows the first two mechanical vibration modes, obtained by finite-element-analysis (FEA) of the piezo-scanning unit. In the first vibration mode, the movement direction of the fiber cable is in phase with the piezo-cantilever, whereas it is out-of-phase for the second mode. Although operation of fiber-scanning exploiting several vibration modes of the extended fiber cable has been utilized in the literature for display [10], and endoscopy [15] applications, higher order modes of the combined piezo/fiber system hve not been treated, to the best of our knowledge. This paper investigates the optimization of the geometries of the piezo-scanner and the fiber optic cable to achieve maximum displacement achieved for both modes for a given actuation force. Our findings, summarized in the next section, reveal that optimum electromechanical conversion efficiency is achieved when fundamental resonance frequencies of the individual piezo-cantilever (f_p) and individual extended fiber cable (f_f) are matched. In other words, f_p and f_f refer to the fundamental resonance frequencies of piezo-cantilever (without the presence of the extended fiber optic cable) only and the extended fiber cantilever (without the presence of the piezo-cantilever) only.

Fig. 2(a) illustrates the mechanical equivalent model of the cascaded resonance system, where the piezo-element constitutes the first resonator, connected to the extended fiber optic cable that forms the second resonator. In the model; k_1 and k_2 represent the spring constants, m_1 and m_2 represent the masses, and b_1 and b_2 represent the damping experienced by the piezo-element and the fiber optic cable, respectively. Although, piezo-structures possess considerable non-linearity at high drive voltages, where models have been introduced using non-linear capacitors (stiffness elements) and resistors (dampers) [16,17], our major aim here is to decrease drive voltage towards patient-friendly optical interventions. Thus, as indicated in the experimental setup part, we have utilized very low voltages (250 mV) where we do not expect any significant hysteresis. We have further varied the actuation voltage to values <250 mV and observed a linear dependence of the scan length with applied voltage. In this regard, we have considered modeling with linear circuit elements only.

In this study, we exploited both i) the T-coupler for modeling the connection between two resonators as depicted in Fig. 2(a and b) [18] and ii) direct connection of both resonators without any coupling elements as in Fig. 2(c and d) [19]. The T-coupler model, composed of three spring elements (having $\pm k_{12}$ values), not only connects both resonators via set of springs, but also has an impact on the stiffness values of both resonating structures. In the T-coupler, the negative valued spring constants (Fig. 2(a and b)) effectively reduce the spring constants of both resonators and the positive valued spring constant accounts for the velocity (corresponding to "current" in the circuit model) difference between the resonators. As k_{12} is a complicated function of both resonator parameters, we use the stiffness values of the T-coupling model as a fit parameter to match FEA results. Fig. 2(c) depicts the equivalent circuit model, showing the piezo element (1st resonator), T-coupler, and the extended fiber optic cable (2nd resonator).

The parameters for the circuit model above can be calculated as follows; the mass of both structures (m_1, m_2) can be calculated based on volume – density product of the structure (for both piezo-cantilever and the fiber). The spring constants (k_1, k_2) can be calculated based on well-known stiffness formulation for typical cantilever (piezo-cantilever) and circular cross-sectioned elements (fiber optical cable) [20]:

$$k_1 = E_p W_p h_p^3 / 4L_p^3 \quad (1)$$

$$k_2 = 3\pi E_f d_f^4 / 64 L_f^3 \quad (2)$$

where E_p and E_f are the Young's moduli of the piezo-cantilever and the silica fiber, W_p, h_p, and L_p are the width, thickness and length of the piezo-cantilever, d_f and L_f are the diameter and length of the extended optical fiber respectively. For the circuit models, the damping values (b_1, b_2) are inserted into the model based on the experimentally measured quality factors for the first and second mechanical mode of the system. Upon using an long cantilever (L_p = 20-mm, such that f_p is approximately equal to 1st resonance frequency of the system) to separate 1st and 2nd mechanical modes of our piezo/fiber configuration, one can assume that the quality factor of the first mode can be approximated to be equal to the quality factor of the piezo-cantilever. Similarly, the quality factor of the second mode can be approximated to be equal to the quality factor of the extended fiber. We have also assumed that the individual quality factor of the piezo-cantilever is constant within the length-range used in this study. The damping coefficient can then be deduced using [20]:

$$b_{1,2} = \frac{\sqrt{k_{1,2} m_{1,2}}}{Q_{1,2}} \quad (3)$$

where Q_1 and Q_2 are the quality factors of the 1st and 2nd resonances of the overall system, respectively. For a long cantilever (L_p = 20-mm), there is minor interaction between the two mechanical modes, leading to resonance frequencies of the combined system (1st and 2nd mode frequencies) to approximately match f_p and f_f. Thus, the frequency response, experimentally acquired at L_p = 20-mm, and the observed resonance frequency of the 1st mode can be specifically utilized to extract the effective Young's modulus of the bimorph piezo-cantilever (E_p) using [20]:

$$f_p = \frac{1}{2\pi}\sqrt{\frac{k_1}{m_{1\_eff}}} = \frac{1}{2\pi}\sqrt{\frac{E_p W_p h_p^3}{4L_p^3 \rho_p}} \quad (4)$$

where, m_{1_eff} is the effective mass, and ρ_p is the effective density of the piezo bimorph. The density was calculated as 7455 kg/m^3 based on weight measurement of the cantilever weight with a microscale and dimensions with a surface profiler. Using Eq. (4) and the measured 1st resonance frequency (see Results section) we deduce the effective Young's modulus to be 49.5 GPa, which was then introduced to both FEA simulations and analytical calculations.

The models introduced in Fig. 2(b and d) are both composed of three equivalent impedance elements (Z_1, Z_2, Z_3), where the overall impedance is a series combination of Z_1 with parallel combination of Z_2 and Z_3

($Z_{TOTAL} = Z_1 + Z_2 || Z_3$). The transfer function: H(s), i.e. the amount of displacement at the fiber tip (integral of the current through Z_3):