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A scalable syringe-actuated microgripper for biological manipulation

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ABSTRACT

Recent developments in the functionality of micro-electromechanical systems (MEMS), particularly for medical and biological applications, have led to an increasing demand for micromanipulation devices. This paper describes the design, fabrication, and testing of a family of pneumatically driven microgrippers which can be scaled to handle millimetre to nanometre compliant and non-compliant objects, with the potential to control gripping forces. In contrast to conventional actuation methods including piezoelectric, magnetic and thermal, pneumatic actuation has the advantages of large power density, the potential for force control, low cost and simplicity. The reported prototypes were fabricated using straightforward processes, in contrast to previously reported pneumatically actuated manipulators. The overall aim of the work is to demonstrate a family of low-cost, polymer based micro grippers that can be actuated manually using pneumatic forces (e.g. via a syringe). A version of the device has been successfully fabricated using laser micromachining and assembled to give an output force of up to 8 ± 0.01 mN. The pneumatic actuation was implemented in such as way that it can open the jaws of the micro gripper in a precisely controlled way, demonstrated on a prototype for handling various compliant objects smaller than 200 μm in diameter. Finite Element Analysis (FEA) was used to calculate the gripping force, and the results compared with the experimental measurements. The scaling of the demonstrator and its reverse actuation to increase the gripping force are discussed on the basis of the FEA.

1. Introduction

A number of microdevices with dimensions from 1 μm to 1 mm for the industries of electronics, information technologies, optics, and bio-technologies have been developed recently. Consequently, the development of microgripper technologies to fulfil the need in handling the microdevices is urgently required [1,2]. Microgrippers are complex micro-electro-mechanical-systems (MEMS) [3] and can be categorized according to their actuation mechanism: electrostatic, shape memory alloy (SMA), magnetic, and piezoelectric, each one of which has its advantages and disadvantages.

Electrostatic actuation generates a satisfactory amount of output force [4] but is difficult to operate in ion-rich liquids (e.g. body fluids). SMA actuators, while producing a high force and displacement, are problematic in liquid environments due to heat loss associated with the high surface-to-volume ratio of microdevices [5]. Furthermore, the displacements of SMAs are hard to control that is because of their thermomechanical nonlinearities [6]. Piezoelectric actuation offers high speed and good motion resolution [7] but actuation displacements are limited [8]. Due to the disadvantageous scaling of magnetic fields, magnetic microactuators have low force output and their performance is also limited because the thermal dissipation by the conductive materials with significant currents passing through. Also, magnetic fields can pose problems in the biological environments [9].

Fluidic microactuators, with their high force and power densities, have been studied extensively for manipulating precise amounts of liquid within microfluidics and micro total analysis systems (μTAS) [10]. The actuation can be either pneumatic or hydraulic, and can therefore operate in liquid environments for biological manipulation. However, there are relatively few studies on developing this type of actuator even though they have the potential to produce among the highest force and power densities at the microscale [10]. Relatively straightforward fabrication processes also make these actuators attractive as a design option.

This paper reports the design of a pneumatically actuated microgripper capable of scaling to the level of manipulating a single biological cell. The main design challenges are to provide control over the opening and closing displacements and the gripping force, which is achieved by a compliant gripper design and two-way actuation.

2. Microgripper design

The device design comprises two main parts; the actuation mechanism and the gripper arms, as shown in Fig. 1. The

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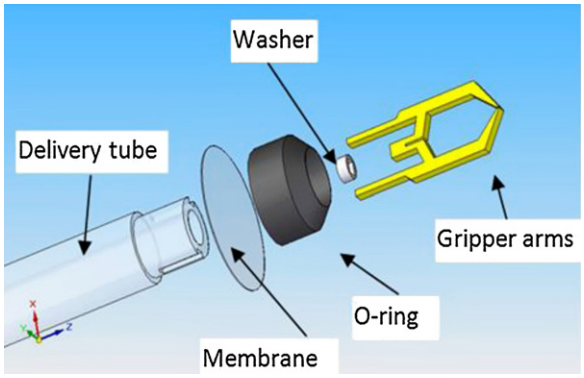

Fig. 1. Conceptual design.

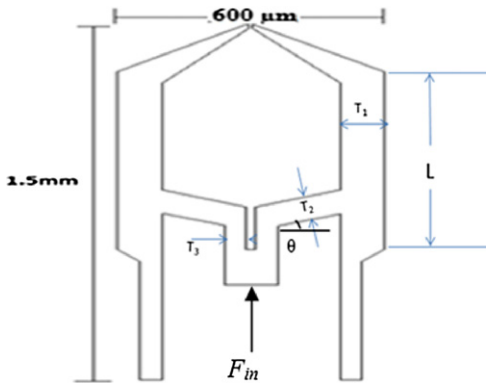

Fig. 2. Design variables of the microgripper.

actuator is essentially a flexible membrane suspended across one end of a delivery tube and applies a force to the base of the gripper when a pneumatic pressure is applied to it. The gripper consists of two fixed legs and two compliant arms, actuated by a central platform to which the actuation force is applied, as shown in Fig. 2. The sizes of biological cells are usually in the range from 5 to 150 μm, so the outer dimensions were chosen at the higher end to demonstrate the design principles while maintaining scalability. To assess the performance, five design variables were investigated using Finite Element Analysis (FEA); thickness of gripping arm (T1), thickness of linkage arm (T2), thickness of actuation arm (T3), length of gripping arm (L), and inclination angle of linkage arm (θ). These were varied by ±25% around a set of reference dimensions; T1 = 1 mm, T2 = 1 mm, T3 = 0.5 mm, L = 5 mm and $\theta = 20^{\circ}$

2.1. Actuation mechanism

The actuation is achieved by using a flexible membrane fixed to one end of a 'delivery' tube. The membrane will expand and contract as air pressure in the tube rises and falls, by means of a pneumatic source. A schematic diagram of the pneumatic actuation is illustrated in Fig. 3. The resulting change in vertical displacement of

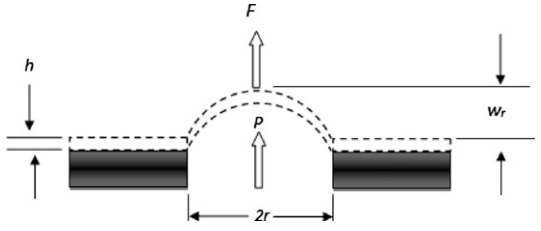

Fig. 3. Schematic diagram of pneumatic actuation.

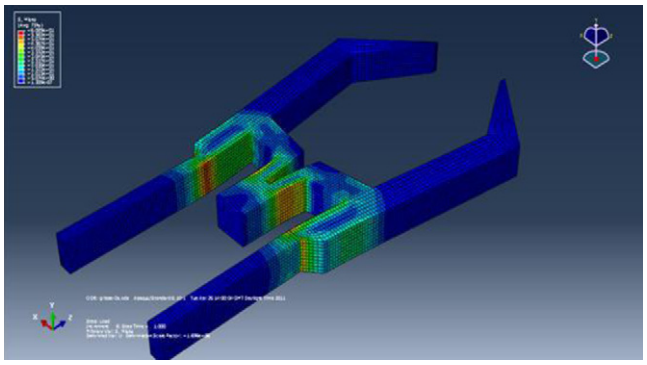

Fig. 4. Stress distribution over the microgripper.

the membrane will provide the means of actuation for the gripping arms.

2.2. Finite element analysis

The microgripper design was analyzed using finite element in ABAQUS 6.10. The analysis used the properties of PMMA which exhibits a good elastic response (Young's Modulus of 1.8 GPa and Poisson's ratio of 0.35) at low strain rates [11]. A seed size of 0.045 was used when meshing the part. The model legs were constrained and the actuation force increased until the tensile strength of the PMMA was reached at some point in the model. The distribution of stress over the microgripper is shown in Fig. 4. Fig. 5a and b shows the effect of the sensitivity analysis to T1, T2, T3 and L, and to θ, respectively.

As might be expected, T1 has the largest effect on opening as it controls the bending stiffness of the part of the gripper arm below the point where the hinge joins; it also will affect the passive gripping force for a given degree of jaw opening. The angle θ also has a significant effect as it affects the bending length of the gripper arm.

The results in Fig. 5a and b were re-plotted as in Fig. 6 to observe the curve of stiffness of the gripper for each design variable. As can be seen, the curve of the stiffness of the gripper is almost linear for all the parameters, whereas it shows a higher order dependence for T1.

3. Fabrication and assembly

A gripper was fabricated to the reference dimensions using a maskless laser microfabrication procedure. It was cut from 1 mm thick polymethyl-methacrylate (PMMA) sheet using a TROTEC Speedy 300^{TM} CO_2 laser cutter operating with the power and speed adjusted to 9 W and 0.6 mm/s, respectively, using several passes to ensure a clean and precise cut without burning. Under these conditions, the cut can be produced with an edge precision of 50 μm. The diameter at one end of the delivery tube was reduced and slots made in its wall to allow the gripper arms to be fixed to the outside of the delivery tube and over the top of the membrane. A combination of a washer and rubber O-ring was used to form an air tight seal around the membrane and hold the gripper arms in place, epoxy resin bonding the gripper arms to the membrane and also sealing the washer and O-ring to the delivery tube. Upon completion of the assembly a syringe was attached to the other end of the delivery tube to act as a pneumatic air source as shown in Fig. 7.

4. Calibration of output force

An important indicator of the performance of the device is its output force. A new approach illustrated in Fig. 8 was used to

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