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A hydrogel-based intravascular microgripper manipulated using magnetic fields☆

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ABSTRACT

This study presents a magnetic hydrogel-based microgripper that can be wirelessly manipulated using magnetic fields. The proposed device can move freely in liquids when driven by direct current (dc) magnetic fields, and perform a gripping motion by using alternating current (ac) magnetic fields. The device is fabricated from a biocompatible hydrogel material that can be employed for intravascular applications. The actuation mechanism for gripping motions is realized by controlling the exposure dose on the hydrogel composite during the lithography process. The preliminary characterization of the device is also presented. The measurement results show that the gripping motion reached a full stroke at approximately 38 °C. By dispersing multiwall carbon nanotubes (MWCNT) into the material, the overall response time of the gripping motion decreases by approximately 2-fold. Device manipulations such as the gripping motion, translational motion, and rotational motion are also successfully demonstrated on a polyvinyl chloride (PVC) tube and in a polydimethylsiloxane (PDMS) microfluidic channel.

1. Introduction

Cardiovascular diseases have become increasingly common worldwide. The coronary arteries are critical vessels for supplying the heart with nutrients, and coronary artery anomalies often cause cardioplegia and death [1,2]. Therefore, blood vessel therapy has recently become a popular medical practice. Intravascular surgery is one of the possible methods of blood vessel therapy [3,4]. In general, intravascular surgery requires assistance from microdevices to deliver diagnostic and therapeutic modalities [5,6], and several types of untethered microdevice have been developed.

Typically, untethered microdevices scavenge energy from the environment and convert that energy into mechanical energy for inducing locomotion by using certain principles [7]. Donald et al. proposed an untethered microrobot, operated using electrostatic force. The device was equipped with a curved steering arm that was mounted on an untethered scratch drive actuator. The proposed device could be remotely controlled to travel through complex paths [8]. Fukuta et al. presented a micromachined pneumatic actuator that can be employed for air-jet planar micromanipulation, and also proposed a pull-in voltage minimization method for reducing the voltage required for electrostatic actuation [9]. Erdem et al. proposed a microrobot that was propelled by cilia-like thermal bimorph actuator arrays. Groups of cilia were controlled independently for generating planar motion with three degrees of freedom [10]. Hu et al. proposed a hydrogel-based microrobot that was optothermally actuated using laser-induced bubbles. The proposed device did not consist of solid materials, but rather employed a gas bubble in a liquid medium for physically manipulating objects [11].

Recently, magnetically driven microdevices have attracted attention because they can be wirelessly driven and provide a relatively large actuation force [12–19]. In general, they could be operated in magnetically transparent media, such as air, vacuum, conducting liquids, and non-conducting liquids. Frutiger et al. proposed microrobots that were operated using wireless magnetic micro-actuators. The microrobot could be actuated using alternating current (ac) magnetic fields, which directly transformed into mechanical actuation without requiring intermediate conversion by using an electronic circuit [20]. Leong et al. proposed a mass-producible, tetherless microgripper. The locomotion of the device could be manipulated magnetically, and the gripping motion could be triggered by controlling the temperature [21]. Jiang et al. proposed a ball-shaped microrobot with rolling capabilities. Driven by magnetic fields, the device could freely roll on 3D surfaces in air, water, or silicon oil [22]. Tottori et al. proposed a magnetic helical microswimmer, fabricated using 3D direct laser writing. The device was capable of performing steerable corkscrew motions

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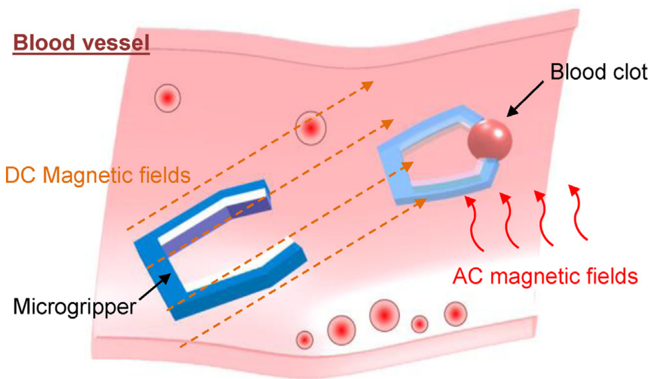

Fig. 1. The schematic of the proposed microgripper. The device can move freely in liquid when driven by dc magnetic fields, and perform gripping motions by using ac magnetic fields.

in water [23]. Kim et al. proposed a micromachined magnetic hydrogel cell carrier fabricated using a single ultraviolet (UV) exposure. The primary application of the proposed device was the active separation of cell carriers from the original solution [24]. Bergeles et al. developed servoing magnetic intraocular microdevices, which were proposed for applications such as targeted drug delivery and the retinal vein cannulation procedure. An algorithm that localizes the micromachined device based on the shape of devices has also been proposed [25].

For intravascular surgery, it is desirable to use microdevices to perform clinical actions such as drug delivery, sensing, and surgery. Certain studies have employed soft materials that require simple fabrication processes, and have demonstrated the reversibility of shape changing in response to stimuli [26–32]. However, because of challenges in fabrication and manipulation, most microdevices can only be operated as either a freely movable unit or an end effector. In this study, a hydrogel-based microgripper is proposed that can be wirelessly actuated for translational, rotational, and gripping motions by using direct current (dc) and ac magnetic fields [33]. The proposed device, which is made of a biocompatible hydrogel material, is suitable for intravascular applications. In addition, the microgripper can be fabricated using a simple lithography technique.

The remainder of this study is organized as follows: the operational principles and design are presented in Section 2. The proposed fabrication process of the microgripper is described in Section 3. The measured results of the fabricated microgripper and the discussion are presented in Section 4. Finally, Section 5 draws the conclusions.

2. Design

Fig. 1 shows the schematic of the proposed microgripper. The device can move freely in liquids when driven by dc magnetic fields, and perform gripping motions by using ac magnetic fields. Potentially, the proposed device can grip an object such as a blood clot in a blood vessel for intravascular therapy. In addition, the proposed device, which is made of a hydrogel material, is suitable for intravascular applications that require biocompatibility.

Fig. 2(a) shows the operational principle of the device locomotion. Fe_3O_4 nanoparticles (NPs) and multiwall carbon nanotubes (MWCNTs) were dispersed in the thermoresponsive hydrogel. Because of the dispersed Fe_3O_4 NPs in the hydrogel, the movement of the device can be wirelessly controlled by applying dc magnetic fields. As shown in Fig. 2(a), without applying external magnetic fields during the fabrication process, the nanoparticles are randomly dispersed in the pre-gel solution of hydrogel. When applying an external magnetic field during process, the superparamagnetic nanoparticles form chain-like nanostructures along the direction of the applied magnetic field [15]. Hence, the fabricated hydrogel-based microgripper possesses a specific magnetic axis, which is expected to enable a more precise manipulation. When the direction of the applied magnetic fields is changed, the device, which has a specific magnetic axis, can rapidly rotate or realign along the direction of the fields.

Fig. 2(b) shows that the gripping motion can be realized by the bimetallic hydrogel composite with layers of different cross-links. Hydrogel polymers contain pendent benzophenone units that allow the tuning of cross-links by using irradiation doses [26]. By controlling the exposure dose on the hydrogel composite during the lithography process, the fabricated hydrogel with layers of different cross-links can induce different shrinking responses at lower critical solution temperatures. By applying ac magnetic fields, the Fe_3O_4 NPs are heated because of the Néel and Brownian relaxation process, which in turn induces the internal temperature elevation of the hydrogel matrix [27]. Because of the non-homogeneous shrinking responses in the hydrogel structure, the temperature elevation creates an internal stress gradient, causing the deformation of the structure, thus generating the gripping motion. In addition, by dispersing MWCNT molecules in the hydrogel, the composite exhibits a shorter thermal response time because of the enhancement of the mass transport of water molecules [28–30].

The magnetic force (F_m) exerted on the microgripper for a translational motion (in x-direction) is proportional to the gradient of the magnetic field [34]:

$$F_m = V_b M \nabla B_x \tag{1}$$

where V_b is the volume of the magnetized object with a uniform magnetization M, and B_x is the magnetic field in x-direction.

As the microgripper moves in x-direction driven by the magnetic field gradient, the device also experiences a drag force acting by the fluid. The typical form of the drag force for an object immersed in a fluid can be written as [35–37]:

$$F_d = \frac{1}{2} C_d \rho_f A v_m^2 \tag{2}$$

where ρ_f is the density of the fluid, v_m is the relative velocity of the object with respect to the fluid medium, A is the cross sectional area of the object, and C_d is the drag coefficient. In laminar flow regime, the drag coefficient and drag force for a sphere shape can be approximated as:

$$C_{d,shpere} = \frac{24}{Re} = \frac{24\eta}{\rho_f v_m D}; \quad F_{d,sphere} = 3\pi D \eta v_m \tag{3}$$

Similarly, for a circular disk parallel to the flow, the drag coefficient and drag force can be written as [35–37]:

$$C_{d,disk} = \frac{13.6}{Re} = \frac{13.6\eta}{\rho_f v_m D}; \quad F_{d,disk} = 1.7\pi D \eta v_m \tag{4}$$

where ($Re = \rho_f v_m D/\eta$) is the Reynolds number, and D is the diameter of the object. Also, the cross sectional area in (2) is:

$$A = \frac{\pi D^2}{4} \tag{5}$$

Since the microgripper is neither a sphere nor a circular disk, we may employ the similar form shown in (3) and (4), and write the drag force for the microgripper as:

$$F_{d,gripper} = S\pi D_G \eta v_m \tag{6}$$

where D_G is the length of the gripper, and S is the shape correcting constant of the microgripper. The discussion of the upper and lower limits of S is described in Appendix.