

# Evaluation of a Dielectric Elastomer Actuator Made of an Organogel Fabricated by 3D Printing

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**Abstract**—In this paper, we fabricated a 3-dimensional (3D) organogel by using 3D printing and evaluated its properties. DEAs have been widely used in various fields because of their high energy density and fast response. DEAs consist of a dielectric elastomer and flexible electrodes, and their strain is influenced by the relative permittivity and Young's modulus of the elastomer. Organogel has high relative permittivity and low Young's modulus compared to other dielectric elastomers. Thus, DEAs made of an organogel can generate significant strain. This characteristic is ideal for applications such as organ models and tactile presentation devices. However, it is difficult to fabricate complex-shaped organogels by using conventional fabrication methods. 3D printing is one of the most effective methods for fabricating complex shape. Therefore, we used a 3D printer to realize complex-shaped DEAs made of an organogel.

## I. INTRODUCTION

In recent years, dielectric elastomer actuators (DEAs) have been attracting significant attention because of their high driving frequency and high energy density [1]. DEAs consist of a dielectric elastomer film and flexible electrodes as shown in Fig. 1. When electrical voltage is applied to the DEA, the DEA is stretched in the in-plane direction by Maxwell stress. Maxwell stress  $\sigma$  can be written as following eq. (1).

$$\sigma = -\epsilon_r \epsilon_0 \left(\frac{V}{d}\right)^2 \quad (1)$$

In this equation,  $\epsilon_0$  represents the permittivity of vacuum,  $\epsilon_r$  is the relative permittivity of the dielectric elastomer,  $V$  is the applied voltage, and  $d$  is the thickness of the dielectric elastomer. Assuming that DEA is parallel plate capacitor and Young's modulus of the dielectric elastomer is constant during actuation, the strain of DEA  $S_t$  can be expressed as eq. (2) when the Young's modulus of the dielectric elastomer is  $E$ .

$$S_t = \frac{\epsilon_r \epsilon_0 V^2}{E d^2} \quad (2)$$

When DEAs are applied in the field of biomimetics and haptics, large strain is needed. The strain of DEAs can be made larger by applying a higher voltage, using a thinner dielectric elastomer film, and selecting a dielectric elastomer material having higher relative permittivity and lower Young's modulus. However, increasing the voltage and decreasing the thickness of the elastomer also raises the risk of dielectric breakdown [2]. Thus, using materials with high relative permittivity and low Young's modulus is essential to achieve large strain.

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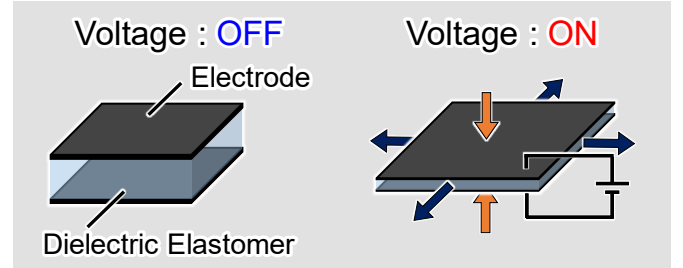


Figure 1. Drive principle of dielectric elastomer actuator

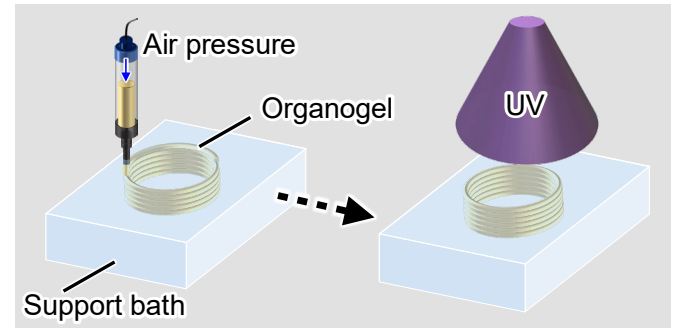


Figure 2. Fabrication process of a 3D-shaped organogel

Generally, acrylic or silicone materials are used as dielectric elastomer materials in DEAs [3]. However, strain of DEAs using these materials is relatively low. Recent research by Li Shi [4] has demonstrated significant strain improvement in DEAs by using an organogel. The organogel has high relative permittivity and low Young's modulus compared to acrylic or silicone materials. Thus, by using the organogel, strain of DEAs can be enlarged.

Our group has studied fabrication methods of three-dimensional (3D) shaped DEAs made of the organogel by using a dip-molding method [5]. The dip molding method is suitable for fabricating relatively simple structures, but it is not suitable for fabricating dense or complex shapes. 3D printing is superior to conventional methods in fabricating complex 3D shapes. By using a 3D printer to fabricate complex shapes with the organogel, we think that we can realize fabricating organ models having realistic structures or tactile presentation devices with various 3D structures.

In this study, we used direct ink writing (DIW) 3D printing to fabricate a 3D-shaped organogel as shown in Fig. 2. The organogel is a light-curing material. Thus, light-curing 3D printing methods are required. As famous light-cured

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3D printings, there are stereolithography (SLA) and DIW. SLA has problems in removing support structures when fabricating complex shapes. On the other hand, DIW uses a support material that can be easily removed by washing. Thus, DIW was considered the most suitable method for fabricating 3D-shaped organogels. Therefore, we fabricated a 3D-shaped organogel by using DIW 3D printing. As experiments, we evaluated the basic properties of the organogel fabricated by 3D printing. Furthermore, we fabricated 3D-shaped organogel by using 3D printing and drove a DEA made of an organogel.

## II. BASIC EVALUATIONS OF AN ORGANOGELS

We first explain preparation method of the molded and 3D printed organogel. Next, we will evaluate the electrical properties, focusing on relative permittivity, and mechanical properties using a tensile testing machine. These electric and mechanical properties of the molded organogel and the 3D printed organogel are compared.

### A FABRICATION OF ORGANOGEL MEMBRANE

To prepare the organogel precursor solution, first, organic solvents, ethylene carbonate (Kanto Chemical, 14086-00, Japan) and propylene carbonate (Kanto Chemical, 32455-00, Japan), are mixed in a volume ratio of 4:1. Next, monomers, 2-ethylhexyl acrylate (Kanto Chemical, 14524-00, Japan) and 4-acryloylmorpholine (Tokyo Chemical Industry, A0841, Japan) are added to the mixture in a volume ratio of 1:1. Then, a photopolymerization initiator, 1-hydroxycyclohexyl phenyl ketone (Tokyo Chemical Industry, H0617, Japan) and the crosslinking agent, N,N'-methylenebis(acrylamide) (Kanto Chemical, 25787-30, Japan), are mixed with 0.132 wt% and 0.2 wt% of monomers, respectively. Next, we explain a fabrication process of molded organogel. The organogel precursor solution is poured into a glass mold capable of transmitting ultraviolet (UV) light, by using a syringe. The glass mold consists of aluminum plates as spacers, glass, and silicone release film (ID Create Inc, J0L-A4, Japan). The flat-shaped organogel is crosslinked by irradiating UV light for two hours using an UV light exposure system (Sunhayato, BOX-S3000, Japan). Finally, we explain fabrication process of organogels by using DIW 3D printer (Bio X Gen 2, CELLINK, Sweden). The precursor solution from the syringe was extruded into the support material to form square shape. Then, the flat-shaped organogel was irradiated for two hours using the UV light exposure system same as the mold process.

### B EVALUATION OF BASIC MATERIAL PROPERTIES

We evaluated the material properties, specifically the relative permittivity, stretchability, and Young's modulus. We compared these properties of the molded and the 3D printed organogel with horizontal and vertical stacking directions.

First, we evaluated the relative permittivity of the molded and 3D printed organogels using an impedance analyzer (HIOKI, IM3570, Japan). The thickness of the molded organogel elastomer was 850  $\mu\text{m}$ , and the 3D printed organogel was 500  $\mu\text{m}$ . Carbon grease (MG Chemicals, 8481-1, Canada) was applied as electrodes on the elastomer by using a brush. The testing voltage was set at 1  $V_{rms}$ , and the frequency range was 4-100 Hz. The results of the relative permittivity are shown

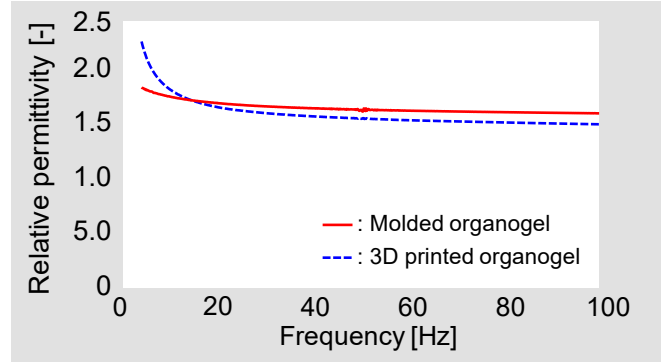


Figure 3. Relative permittivity of organogels

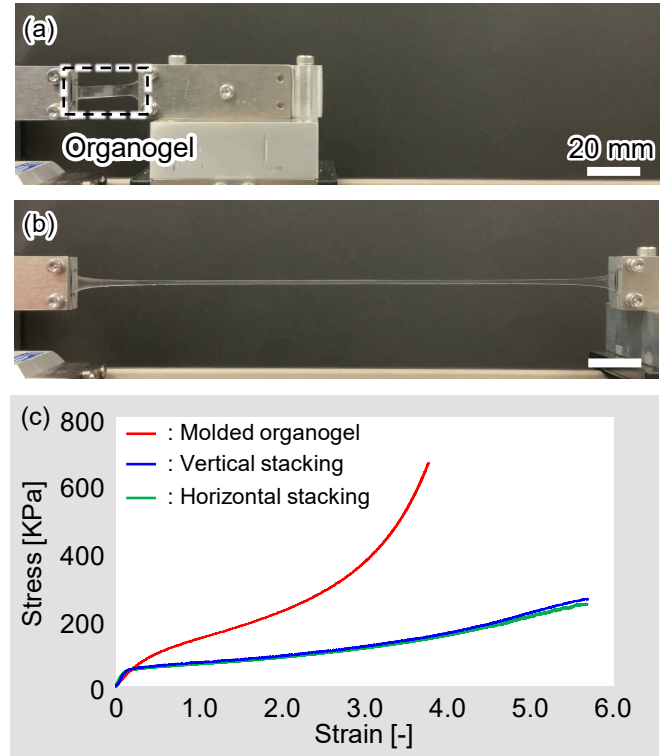


Figure 4. Tensile testing of an Organogel  
(a) before stretching the gel, (b) after stretching the gel,  
(c) Stress- strain curves of organogels

in Fig. 3. it was found that the relative permittivity decreases as the frequency increases. This is likely due to the dielectric polarization of the organic solvent and uncrosslinked monomers within the organogel being dependent on frequency. At frequency between 4 Hz to 20 Hz, the relative permittivity of the 3D printed organogel was particularly large. This is likely due to the formation of defects and voids caused by extrusion or overexposure. The relative permittivity of the molded organogel ranged from 2.4 to 1.5, while that of the 3D printed organogel was 1.8 to 1.6 between 20 Hz and 100 Hz.

Next, tensile tests were conducted using a tensile testing machine to evaluate a stretchability and Young's modulus as shown in Fig. 4. Each organogel was processed into a dumbbell-shaped No. 8. form. Figure 4 (c) shows results of the tensile tests as stress-strain curves. The molded organogel was stretched approximately 3.8 times of its original length. 3D printed organogels, regardless of stacking direction, were stretched approximately 5.7 times. Additionally, Young's

modulus of the molded organogel was approximately 220 kPa, while the 3D printed organogels, regardless of stacking direction, was approximately 500 KPa. No significant differences were observed between stacking direction of the 3D printing. However, the Young's modulus of the 3D-printed organogel was higher than molded one. The increase in Young's modulus of the 3D printed organogel is thought to be because of an overexposure.

### III. 3D PRINTING OF ORGANOGEL

#### A. Fabrication of a support material and organogel ink

Support materials are essential in DIW 3D printing to fabricate complex 3D shapes. Support materials are required to maintain the shape of a printed ink. To realize 3D printings of soft materials such as organogels, Bingham fluids are used as a support material in DIW 3D printing [6]. Bingham fluids are non-Newtonian fluids that exhibit both solid-like and liquid-like behaviors depending on an applied stress. When external stress exceeds their yield shear stress:  $\gamma$ , they flow like liquid. However, when an external stress does not exceed their yield shear stress, they behave as a solid. In DIW, Bingham fluids mainly consist of soft droplet dispersions [7],[8]. To maintain the shape of the printed ink, the yield shear stress  $\gamma$  of support materials needs to be sufficiently greater than the body forces  $\rho gh$ , where  $\rho$  is density,  $g$  is gravity, and  $h$  is depth of printed ink. If the yield shear stress  $\gamma$  of the support material is insufficient relative to the body force, the support materials cannot maintain the shape of the printed ink as shown in Fig. 5 (a). Conversely, when the yield shear stress is adequate, the support material can maintain the shape of the printed ink as shown in Fig. 5 (b). However, even under the stable conditions of  $\gamma$  and  $\rho gh$ , the viscosity of printed ink and interfacial tension between the printed ink and the support material affect the accuracy of 3D printing. Low viscosity of the printed inks and high interfacial tension can cause the deformation or breakout of 3D-printed structures before they solidify as shown in Fig.5 (c). Therefore, high yield shear stress, high viscosity, and low interfacial tension contribute to stabilizing the shape of the printed inks as shown in Fig.5 (d).

To prepare a support material [9], first, we mixed pure water and glycerin (Kanto Chemical, 18469-1A, Japan) with silicone oil (Shin-Etsu Chemical Co., Ltd., KF-96L-5cs, Japan) and surfactant (Tokyo Chemical Industry, S0060, Japan). Next, we emulsify them by using a homogenizer (AS ONE Corporation, S-203, Japan) to create silicone droplets. Finally, we increase the concentration using a centrifuge (Kubota Shoji Co., Ltd., Model 3740, Japan). Figure 6 (a) shows the fabricated silicone support material. This silicone support material is suitable for 3D layering of organogel due to their low solubility in each other. However, the non-polar nature of the support material and the polar nature of the organogel cause high interfacial tension and unstable extrusion. This issue was resolved by increasing the viscosity of the organogel precursor solution. To increase viscosity of the precursor solution, the precursor solution was irradiated by UV light (Asahi Spectra Co., Ltd., UD-1754, Japan) for 30 seconds before adding the crosslinker.

#### B. Fabrication of a 3D-shaped organogel

By using the support material fabricated in section III A, we

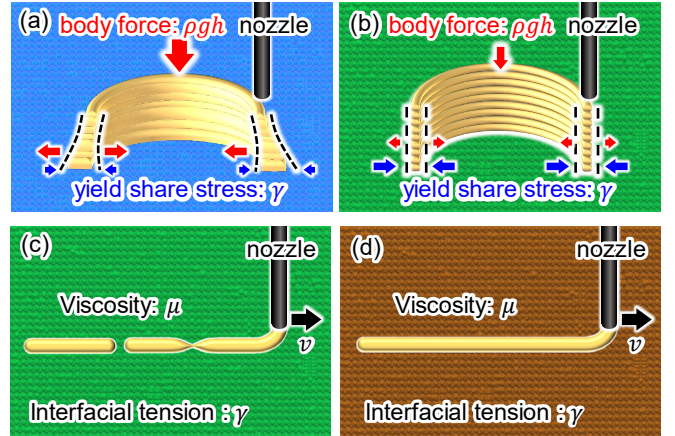


Figure 5. The relationship between yield shear stress, shape of printed inks, viscosity, and interfacial tension  
(a) high yield shear stress, (b) low yield shear stress  
(c) low viscosity and high interfacial tension,  
(d) high viscosity and low interfacial tension

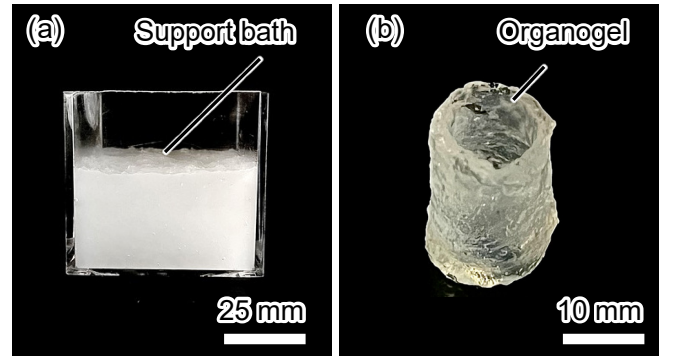


Figure 6. Fabrication of a 3D-shaped organogel  
(a) support material, (b) 3D-shaped organogel

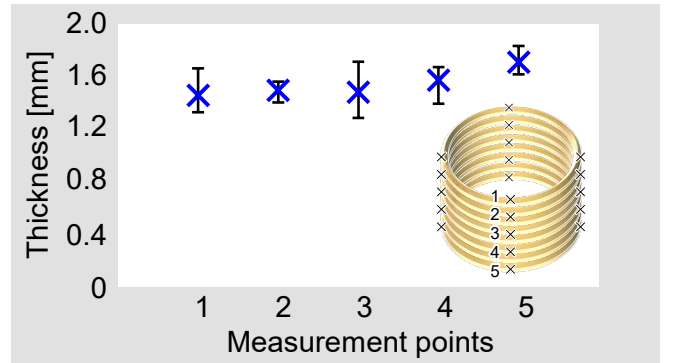


Figure 7. Thickness of the organogel

fabricated a tubular structure having diameter of 10 mm as an example of 3D-shaped organogel. As the printing conditions, we set air pressure as 100 kPa, nozzle speed as 1.0 mm/s and we repeat writing the circle for 35 times with 0.41 mm z displacement. We succeeded in fabricating a 3D-shaped organogel as shown in Fig. 6 (b). Next, we measured the thickness of the organogel at intervals of 3.5 mm from the top to the bottom, as shown in Fig. 7. the average thickness was  $1.5 \pm 0.067$  mm and the variation were  $\pm 4.3\%$ . It was founded that the film thickness increases from the top to the bottom due to insufficient yield shear stress of the support material. To address this issue, it is necessary to increase the yield shear stress. Improving the yield shear stress can be achieved by

reducing the particle size of the silicone droplets and increasing the concentration of the silicone droplet dispersion. Therefore, increasing the homogenization time and centrifugation time is considered an effective method to enhance the yield shear stress of the support material.

#### IV. DRIVE OF A 3D-PRINTED ORGANOGEL DEA

##### Experimental System of Balloon-Type DEA

To evaluate the actuation capability of the 3D printed organogel, we constructed an experimental system of a balloon-type DEA by using the flat-shaped organogel fabricated in chapter II [10] using water pressure as shown in Fig. 8. In this experimental system, an elastomer is attached to an acrylic pipe connected to a water reservoir by a hose, and carbon grease is applied as electrode on a surface of the elastomer. this experimental system can apply pre-stretch to the elastomer using water pressure.

##### B. Drive and evaluation of a DEA made of an organogel

When we applied a 5kV sine wave voltage to the 3D-printed DEA, the DEA was successfully driven in the out-of-plane direction, as shown in Fig. 9. Next, we measured the displacement of the DEA. Figure 10 (a) shows the displacement measurement method. To measure the displacement, we took pictures of ON and OFF states of the DEA, as shown in Fig. 9. We detected the edge of the DEA by using visual analysis and measured distances from the center of the DEA to the measurement point in ON and OFF states. The displacement was determined from the difference in these distances. The results of the displacement measurement are shown in Fig. 10 (b). As shown by the plot, applying the voltage leads to out-of-plane stretching of the DEA with a maximum displacement of approximately 1.1 mm.

#### V. CONCLUSION

In this study, we conducted basic evaluation of 3D-printed organogel. We evaluated the material properties of 3D-printed organogels were not different with the layering direction in DIW 3D printing. However, it was different from the molded organogel. Currently, this difference is thought to be because of a condition of photocuring. Additionally, we fabricated 3D-shaped organogel using DIW 3D printing and succeeded in driving a DEA made of flat-shaped organogel by using DIW 3D printing. In the future, we plan to improve the fabrication accuracy by enhancing the yield shear stress of the support materials and drive the 3D-shaped DEA made of organogel using DIW 3D printing.

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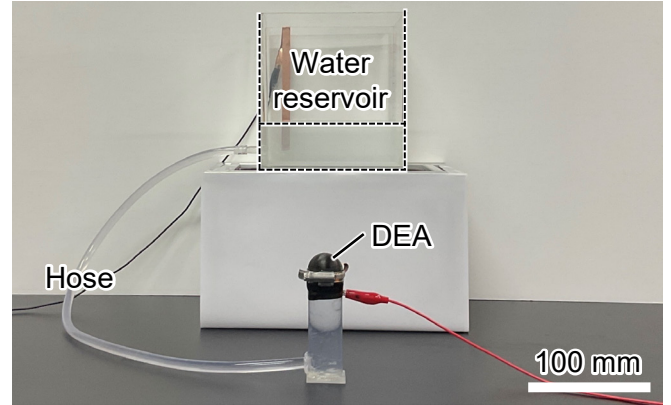


Figure 8. Experimental system of a DEA with water pressure

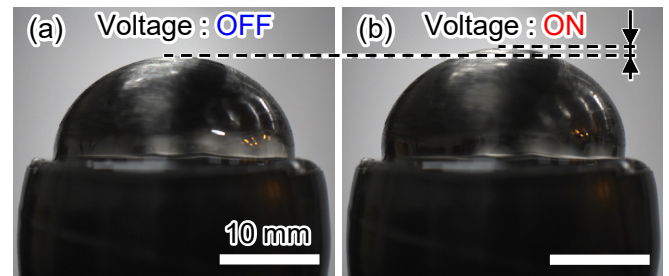


Figure 9. Driving of a DEA made of an organogel (a) voltage OFF, (b) voltage ON

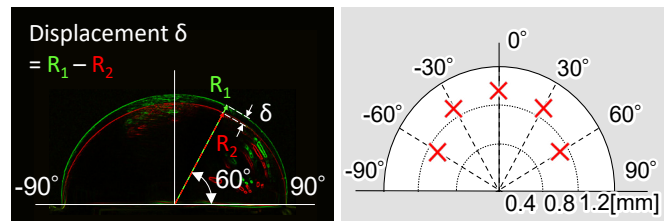


Figure 10. Displacement of the organogel DEA (a) measurement method, (b) displacement graph

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