Additive Manufacturing Applications at the Microscale

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Additive manufacturing (AM) technology has been researched and developed for almost three decades. Microscale AM is one of the fastest-growing fields of research within the AM area. Considerable progress has been made in the development and commercialization of new and innovative microscale AM processes, as well as several practical applications in a variety of fields. However, there are still significant challenges that exist in terms of design, available materials, processes, and the ability to fabricate true three-dimensional structures and systems at a microscale. For instance, microscale AM fabrication technologies are associated with certain limitations and constraints due to the scale aspect, which may require the establishment and use of specialized design methodologies in order to overcome them.

Keywords: additive manufacturing ; microscale ; design ; materials ; processes

1. Actuator Applications

In the field of micro and miniature actuators, microscale AM is indicated as a promising fabrication solution, and it promotes the production of micromachines with complex geometry using monolithic approaches, which would otherwise require a combination of advanced micro-subtractive manufacturing methods and, usually, assemblies with a large number of components. Recent review papers present and discuss the use of AM processes in the design and fabrication of microelectromechanical systems (MEMS) actuators, biohybrid actuators, and piezoelectric systems as relevant applications. In [1][2], the authors investigated the recent developments and achievements regarding the most widely used 3D printing technologies for MEMS fabrication and discussed their challenges and potential. Several papers [3][4] presented recent advancements in the field of small-scale soft robotics and actuators using AM, while other researchers [5][6] examined the application of 3D printing for the fabrication of piezoelectric actuators; finally, the authors of [7][8] discussed an approach regarding the fabrication of biohybrid actuators using AM.

The fabrication of micro-grippers is an indicative example revealing the importance of micro–AM. Accurate tip displacements, which are as small as 20 μm, are necessary for handling and pick-and-place, and the sterile handling of sensitive parts, which is a common procedure in the biology and health sector environments. The actuation principle may be piezoelectric, magnetic, or electrothermal. Shao et al. fabricated magnetically active 3D microstructures using a high-resolution micro-continuous liquid interface production process (μCLIP), combining 3D-printed centimeter-sized samples with sub-75 μm fine features [9]. The magnetic photopolymerizable resin that was used maintains high solid loading (30 wt % Fe_3O_4 nanoparticles), improves the surface properties by reducing the stair-like surface roughness, and accelerates the fabrication process. In another study [10], the authors used the same method (μCLIP) to fabricate a 3D printed magnetically driven triple-finger micro-gripper, and tested its efficiency using a 300 μm diameter microsphere, both in air and in deionized water. The printing process involved the soaking of the part in acetone to remove the residual liquid resin (2 min), then its transfer into ethanol and ultrasonic cleaner (5 min), and finally, after drying (30 min), the specimens were post-cured in UV light of a 405 nm wavelength (10 min). Daniel et al. [11] fabricated a chevron-type electrothermal actuator, using the material extrusion-based manufacturing of a shape memory polymer composite. Using a resistivity of 1.8 Ωcm and an operational voltage as low as 3 V, they accomplished 100 μm tip displacement, which was computationally and experimentally investigated. Their main computational finding was that the grippers can be actuated quickly (3–5 s) with voltages as low as 5 V, but they recover slowly (60–100 s). Experimentally, higher voltages were required for actuation; a tip displacement of up to 77–117 μm was achieved in 5 s with an operational voltage of 17.5–19.5 V. In [12], Tyagi et al. used a custom-built syringe-based extrusion 3D printer to fabricate bilayer micro-actuators, driven by hydrogels, down to a size of 300 × (1000 ÷ 5000) μm^2, with a minimum thickness of 30 μm. The printing resolution was 25 μm in the *x-y* plane; the rate of the lateral motion of the stage was ~2.5 mm/s and the air-dispensing pressure was 50–65 psi. The printing ink consisted of dissolved Hydromend D4 (hydrogel) in ethanol at a concentration of 20%. Lantada et al. [13] presented the development process of geometrically complex micro-vascular shape-memory polymer actuators by laser SLA, using a shape-memory epoxy that could change its shape as an effect of temperature increase. They presented two proof-of-concept applications: an active micro-claw with inner vasculatures of different cross-sections and an active spring with inner vasculatures of different cross-sections. In order to assess the effect of temperature on the closing of the gripper and the compression of the spring, they heated the prototypes with water flow (80 ℃) running through the micro-vasculatures. In [14], Kozaki et al. presented the design of a microgripper for handling spheroid microstructures, mounted on a glass capillary. They used a top-down micro-stereolithography setup, based on a 405-nm blue laser developed in their previous study [15]. The photo-curable polymer used is a mixture of acrylate resin and a photopolymerization initiator, polymerization inhibitor, and blue light absorber (wavelength 405 nm). The mixture was mixed, degassed (2000 rpm and 5 min for each mode), and stirred for 24 h at 60 rpm in a ball mill. The nominal diameter of the micro-gripper tip was 300 μm, while the effective force could reach values of between 0.01 and 0.04 N and the tip displacement varied between 20 and 80 μm, respectively. Alblalaihid et al. [16] demonstrated the application of a sputter-coating process for the deposition of metallic layers on polymer components and validated their approach for the fabrication of a micro-gripper device. They used a 3D projection micro-stereolithography (PMSL) system. The gripper was thermally actuated and the tip displacement, in this case, was in the range of 10–180 μm, depending on the applied potential.

2. Soft Robotics Applications

AM enables the fabrication of complex, freeform, and smart structures [17]. Besides micro-grippers, which, in most cases, maintain a rigid-type behavior during their operation, flexible actuators or soft robotics yield another application of microscale AM. Almeida et al. [18] designed an actuation mechanism for robotic micro-tweezers, based on a 3D-printed nylon flexure and a piezo-bimorph actuator, targeting the desired manipulation range from 100 μm to 1 mm. Bas et al. [19] designed miniature inflatable bending actuators, consisting of ultra-fine fibers (diameter of between 1 and 50 μm) and a soft elastomer matrix able to exhibit diverse movements. They used melt electro-writing (MEW) technology to create the prototypes. Their actuators, with a length of 10–15 mm and an inner diameter of 1 mm, can reach their full range of motion within ~20 ms without exploiting snapping instabilities or material non-linearities. Joyee and Pan [20] fabricated a fully 3D-printed multi-material, multi-modal functional soft monolithic robot, composed of polymer and magnetic particle-polymer composites. The fabrication process was magnetic field-assisted projection stereolithography (M-PSL), capable of fabricating smart particle-polymer composites layer by layer. A photocurable flexible resin was used as the base material for 3D printing, while the magnetic nanoparticles (10 nm in nominal diameter) contained 60–80 wt % iron oxide. The maximum bending deformation was 5.2 mm on the z-axis and the maximum deflection in the xy plane was 146°. Schaffner et al. [21] reported a 3D-printing platform for the seamless digital fabrication of pneumatic silicone actuators, exhibiting programmable bioinspired architectures and motions with spatial resolutions in the range of 300 μm. They used viscoelastic silicone inks, resulting in elastomers with variable stiffness after polymerization. Sinatra et al. [22] introduced a novel fabrication strategy for nanofiber-reinforced soft micro-actuators with 30 μm feature sizes. The design and manufacturing of composite polydimethylsiloxane (PDMS)/nanofiber actuators using soft lithography and rotary jet spinning are described. Among the examined parameters were the lamina design and fiber orientation on the actuator curvature, mechanical properties, and pressurization range. Composite actuators displayed a 25.8% higher maximum pressure than pure PDMS devices. Furthermore, the best nanofiber-reinforced laminates tested were 2.3 times tougher than the control PDMS material, while maintaining comparable elongation. Xavier et al. [23] presented the design and direct 3D printing of novel omnidirectional soft pneumatic actuators using SLA. They used an elastic resin and FDM with a soft thermoplastic polyurethane (TPU), achieving multimodal actuation including bending, extension, and contraction motions under positive, negative, or differential pressures. The printing time for a single actuator using the SLA method was 6 h and 40 min while the printing time using the FDM method was approximately 29 h and 20 min. In [24], Zhang et al. presented a generic process flow for the systematic and efficient tailoring of the material formulation and key processing parameters for the digital light processing-based 3D printing of miniature pneumatic actuators for soft robots. They printed various miniature pneumatic robots with an overall size of 2–15 mm and a feature size of 150–350 μm. They used a commercially available UV-curable elastomer, to which was added 30 wt % epoxy aliphatic acrylate (EAA), leading to a reduction in Young's modulus and an increase in failure strain. All the specimens were post-cured for 10 min. Ge et al. [25] presented the design of a bottom-up digital light processing (DLP) 3D printer system (385 nm UV light source, 50 μm normal resolution) and the fabrication of multiple-size soft pneumatic actuators integrally, with fast speed and high precision. Their experiments demonstrated that the printer could print objects with features as small as 87.5 μm. They also presented the design and fabrication of a soft pneumatic gripper containing three micro pneumatic actuators with 0.4-mm-wide square air channels, as well as 0.2-mm-thick chamber walls.

3. Biomedical and Microfluidics Applications

Microscale AM has been efficiently used in biomedical engineering, including many microfluidic applications, which can also be treated as a separate category. The methods, potential, challenges, and limitations of microscale AM in biomedical engineering have been reported in recent review studies. The applications of 3D printing in the health and pharmaceutical sectors have been thoroughly investigated over the last few years and can be tracked in the following review papers [26][27][28]. According to the authors of [28], the applications can be divided into the following categories: 3D disease modeling, pharmaceutical products, organ printing, and patient-specific in situ implants. Other possible applications include drug-delivery devices [29][30][31], the fabrication of microneedles [32][33], microfluidic devices and biomedical micro-devices [34][35], and the fabrication of tissues [36][37].

Drug delivery applications incorporate design solutions characterized by microscale features where AM has been successfully incorporated. In [38], Joyee and Pan proposed the design of a 3D-printed soft robot capable of multimodal locomotion. Utilizing computer aided design and computer aided engineering (CAD-CAE) tools for the design, they printed the robot via a novel magnetic field-assisted projection stereolithography (M-PSL) technique. This soft robot is capable of bi-directional bending in the *xy* plane and *z*-direction and consists of anterior and posterior legs that contain a drug. The maximum dimensions of the robot in width and height is 5 mm × 5.5 mm, while the drug is released from a 200 μm hole.

In the field of microfluidics, Coltelli et al. [39] combined microfluidics, AM, and electrostatic actuation to design artificial muscles capable of generating up to 33 Mpa stress and 10–20% strain. Their design consists of arrays of rectangular cavities arranged accordingly, filled with conducting material inside a bulk dielectric volume. They suggest that the microfluidic devices are AM-fabricated in such a way that the channels would form wiring when filled with conducting fluid, while the bulk core would serve as the dielectric and as the force-transfer medium. The non-flexed lateral size of the electrode plates was kept at 400 μm × 400 μm. The non-flexed plate thickness was kept at 100 μm for each plate and the non-flexed separation between paired plates within the same micro-capacitor was kept at 100 μm. The accuracy of the SLS printing was kept at 100 μm. In [40], the authors illustrated the direct fabrication of a 3D complex microchannel design using AM, for the continuous mixing of micro/nano-particles with biomolecules. The fabrication process was conducted using the DLP method. After the 3D printing stage, the part was removed and washed with IPA (70% ethanol and water), blow-dried with pressurized air, and, finally, cured under UV light for 120 s. The cross-section of the trapezoidal channel had a width of 600 μm and heights of 80 and 130 μm. Another example of a design of microfluidic MEMS was presented in [41], where the authors proposed a micro-extrusion 3D printing system that contained integrated pick-and-place functionality. The case study was the fabrication of microfluidic-based 3D MEMS (three-dimensional microelectromechanical systems) that contain orthogonal out-of-plane piezoelectric sensors and actuators, using additive manufacturing.

Miniature pumps are very critical components in the health sector. In [42], Thomas et al. fabricated a 3D-printed electromagnetically actuated microfluidic pump, capable of generating a 2.2 μL/min flow rate of biofluid. An FDM process with 100 μm-layer resolution was used to deposit polylactic acid on a plastic filament. Taylor et al. [43] fabricated a multi-material miniature diaphragm pump for the creation and maintenance of a low vacuum from atmospheric conditions, using PolyJet printing. The output surface was assessed in terms of roughness, giving values of R_a in the order of some microns (~2–3 μm), while the R_z values were close to layer thickness (~16–18 μm), which was considered acceptable. The stroke of the pump was 2.5 μm. In [44], a low-cost (~$120), open-source peristaltic pump was constructed with a combination of 3D-printed parts and common hardware. The pump was capable of producing flow rates of up to 1.6 mL min^{-1}.

In the field of microneedles (MNs), Economidou et al. [45] fabricated a hollow MN MEMS system for controlled transdermal drug delivery. They fabricated hollow cone-shaped MNs with a base diameter of 1000 μm, a tip diameter of 100 μm, and a height of 1000 μm using SLA and, afterward, integrated the MNs onto the MEMS. The hollow cones featured a wall thickness of 100 μm and the internal bores had a diameter of 800 μm at the cone base. The MNs were fabricated using an SLA 3D printer, followed by curing for 60 min under 40 ℃ UV radiation. The authors observed smooth surfaces on the MNs (no “stair-stepping” effect), as an outcome of the printing method they selected. In [46], the authors provided the capabilities of FDM low-budget printers (using PLA printing material) to print the non-transparent and closed internal microfeatures of in-plane linear, curved, and spiral microchannels with a diameter of less than 0.5 mm (i.e., linear, curved, and spiral channel profiles) and varying cross-sections. The surface roughness of each microchannel configuration was measured and was found to be in the order of some microns (~0.5–3 μm). In addition, each configuration was tested in terms of leakage flow. Caudill et al. [47] designed and printed microneedle arrays utilizing a three-dimensional (3D)-printing technique called continuous liquid interface production (CLIP). Besides pyramidal MNs, the design involved faceted MNs with horizontal grooves, leading to an increase in surface area and, thus, better vaccination properties. The MNs were 700 μm in height and 500 μm in width and were printed in a 10 × 10 array on a 10 mm × 10 mm patch for vaccine delivery. Chen et al. [48] proposed a novel 3D AM method, known as magnetorheological drawing lithography (MRDL), to efficiently fabricate bio-inspired MNs imitating the honeybee’s stinger. With the assistance of an external magnetic field, a parent MN (20 μm tip width) was directly drawn on the pillar tip, and tilted micro barbs (5 μm tip width) were subsequently formed on the four sides of the parent MN. The fabrication process of the parent MN was conducted by means of insertion and, afterward, the removal of a copper pillar inside a pool filled with curable magneto-rheological fluid (CMRF) under an external magnetic field. Micro barbs were formed later, on the curved surface of the parent MN. Compared with a barbless microneedle, the micro-structured barbs enabled the bio-inspired microneedle to be easily inserted into the skin, with difficult removal. In [49], the authors used a commercially available stereolithographic 3D printing, which was assessed regarding its microscale fabrication properties, in order to fabricate sharp MNs (12 × 12 array, in total, 144; 30 min per patch) with a tip radius of approximately 15 μm. In another study [50], a microneedle mold fabrication technique using a low-cost desktop SLA 3D printer was presented, and the fabrication of needles with high-aspect ratios and tip radii of 20–40 μm took place.

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