BULK-METAL-BASED MEMS FABRICATED BY MICRO-ELECTRO-DISCHARGE MACHINING

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Abstract—**This paper presents the recent development of micro-electro-mechanical systems (MEMS) and devices realized by planar micromachining of metal foils. New types of cardiac stents including sensor-integrated antenna stents, a micromachined Kelvin probe with an integrated actuator, an intraluminal flow sensor cuff, and a mechanically/chemically robust capacitive pressure sensor are reported. Micro-electro-discharge machining (μEDM) and the modified processes were used for the fabrication of the devices with mechanical or electromechanical functionality. The cost-effective manufacturing is potentially available with the use of the batch mode μEDM technology that employs lithographically fabricated microelectrode arrays.**

Keywords: MEMS; Bulk metals; Micro-electro-discharge machining; Stent; Wireless.

I. INTRODUCTION

The advancement of micromachining techniques has led to the evolution of micro-electro-mechanical systems (MEMS). These techniques are typically based on semiconductor manufacturing processes, which offer significant advantages such as batch manufacturing of end products and monolithic integration with microelectronics. Surface micromachining has been used to construct fairly complex microstructures, but their structural geometries are two-dimensional, which often limits their mechanical abilities. This constraint has been addressed by use of bulk micromachining techniques that involve both etching and deposition processes. For etching, anisotropic wet etching and deep reactive ion etching have been widely used to create three-dimensional (3D) geometries in MEMS. However, these processes are mainly applicable only to silicon. As for deposition, electroplating has been used to form 3D metallic microstructures, but practical materials are also limited to a few metals and their alloys. Diversifying bulk materials is a key to achieve new functionalities and higher performance in MEMS, extending their application fields.

Micro-electro-discharge machining (μEDM) is a bulk micromachining technique that is applicable to virtually any electrically conductive material. The technique is capable of creating complex 3D mirostructures with features as small as 5 μm and aspect ratios up to 20-30. It involves the sequential discharge of electrical pulses between a microscopic electrode and the workpiece while both are immersed in dielectric oil [1]. Although it has been commercially used for applications such as ink-jet nozzles and magnetic heads for digital VCRs [2], traditional μEDM is limited in throughput because it is a serial process based on the use of single machining tools, which are typically cylindrical tungsten elements with 5-300 μm diameter. This also limits precision because the electrodes themselves are individually shaped by using a μEDM technique, wire electro-discharge grinding [3], which may cause variation in the electrode shape. It was previously demonstrated by the authors that these constraints could be addressed by using lithographically formed arrays of planar microelectrodes (Fig. 1) [4]. The parallelism, or, batch mode process in μEDM offers not only opportunities to manufacture micro devices and components from a variety of bulk metals with high throughput and precision but also compatibility with other planar microfabrication techniques based on lithography processes, which are the mainstream of MEMS manufacturing.

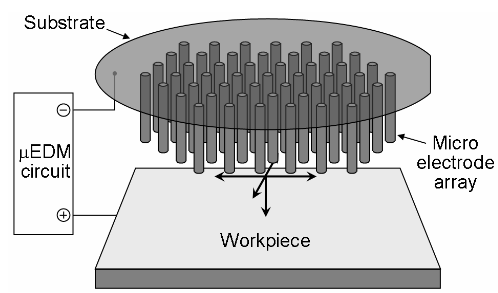

Fig. 1: Batch μEDM with high-aspect-ratio microelectrode arrays fabricated by a LIGA process.

The effort was extended to address the constraint in MEMS, i.e., lack of diversity of bulk materials by applying the μEDM technology. The approach was to use planar metal foil as starting material for the fabrication, which permits the benefit of the parallelism to be exploited, offering high throughput and repeatability. The following section presents the devices developed through the approach.

II. μEDM MEMS BASED ON BULK METALS

The application of μEDM in the effort involved both purely mechanical and electromechanical devices. The μEDM technique can be directly applied to the fabrication of mechanical devices or components by patterning metallic materials. The stent described in Section *A* is included in this group. For the development of electromechanical devices, dielectric portions need to be incorporated in μEDM structures in order to form electrical circuitry in the structures. This challenge was addressed through the development of the devices presented in Sections *B-E*, i.e., antenna stent, micro Kelvin probe, electromagnetic flow sensor, and capacitive pressure sensor.

A. MICROMECHANICAL STENT

Stents are mechanical devices that are chronically implanted into arteries in order to physically expand and scaffold blood vessels that have been narrowed by plaque accumulation. The vast majority of stents are made by laser machining of stainless steel tubes [5], creating mesh-like walls that allow the tube to be expanded radially upon the inflation of an angioplasty balloon. The use of μEDM is another option for cutting metal microstructures. The batch machining approach mentioned above potentially enables the technology to be a promising method for the stent manufacturing.

The fabrication approach in this effort was to μEDM 50-

µm-thick stainless steel foil into a planar structure that could be slipped over an angioplasty balloon, plastically deforming it into a cylinder shape when deployed [6]. The planar pattern had two longitudinal side-beams, which were connected transversely by expandable cross bands, each of which contained identical involute loops (Fig. 2a). In the manner identical to commercial stents, the deployment of the stent was emulated by inflating an angioplasty balloon that was threaded through the planar structure such that the transverse bands alternated above and below it. Figure 2b shows an SEM image of the expanded stent with the balloon removed. The final diameter was 2.65 mm in this case.

Figure 3 shows the measured result of radial stiffness test for the developed stent with type 304 stainless steel. A commercial stent with type 316L stainless steel of thickness varying over 90-130 µm (Guidant Co, IN, USA, Multilink Tetra^{TM}) was also tested for comparison. The test indicated that the developed stent had almost the same radial strength even though its walls were only 50-µm thick. (Note that the mechanical properties of types 304 and 316L of stainless steel are almost identical.) The radial stiffness was similar when the loading was applied at two extreme orientations, i.e., perpendicular to the original plane of the pre-expansion planar microstructure and parallel to the plane [6].

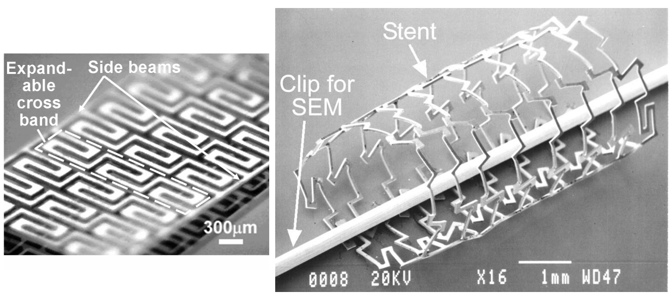

Fig. 2: (a: left) A 7-mm-long planar stent sample as cut by µEDM from stainless steel foil; (b: right) an expanded state of the planar structure.

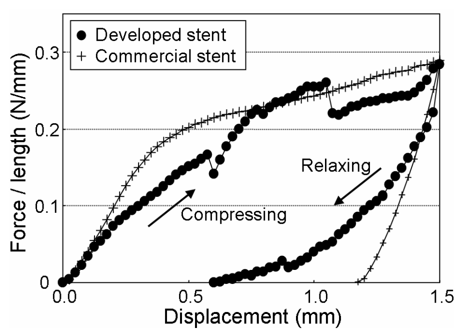

Fig. 3: Measurement of the radial stiffness of a developed stent and comparison to a commercial stent with similar diameter and twice the thickness.

B. Antenna Stent with Microsensors

After stent implantations, re-narrowing (restenosis) often occurs. To determine the status, patients are required to take x-ray angiography periodically, which is an invasive procedure that inserts a catheter to inject contrast dye and cannot be taken frequently. The failure is still a concern even with the recent availability of drug-eluting stents. Wireless monitoring of cardiac parameters such as blood pressure and flow can provide advance notices of such failures (Fig. 4). Toward this end, the planar fabrication approach was utilized to develop a method that automatically transformed the electrical characteristics of a stent during balloon angioplasty, allowing the stent to be a helical-shaped antenna (stentenna) [7]. The planar design of the stent permits the combinational use of lithography-based micromachining techniques for direct fabrication of sensors on the stent as well as the integration of separate micromachined sensors fabricated by the techniques with the stent. This effort took advantage of the latter benefit.

The planar microstructure for the stentenna had a series of cross bands that had involute contours, similar to those in the mechanical stent in Section *A*, with a bridge to a longitudinal beam at the center of the device. The involute bands formed dual inductors, whereas the beam was a common electrical node. Two micromachined capacitive pressure sensors [8] were connected across the common line and the inductors, implementing dual inductor-capacitor (LC) tank configuration (Fig. 5a). The resonant frequency of the tank, which depends on local pressure or flow rate, was wirelessly interrogated through an external antenna that was magnetically coupled to the stentenna. In contrast to the mechanical stent, the stentenna needed to play an electrical role. To warrant this functionality, the fabricated device was coated with Parylene-C^{TM} for electrical protection while granting biocompatibility to the device. The stentennas were then deployed inside a silicone-based mock artery using a standard angioplasty balloon, resulting in a helical shape with inductance of approximately 110 nH (Fig. 5b).

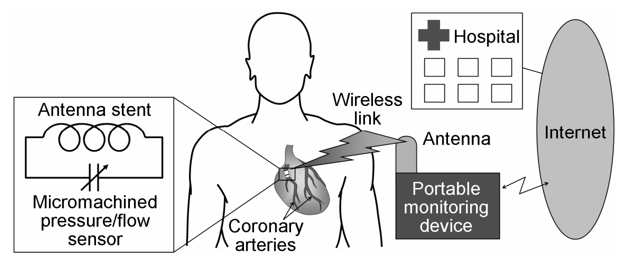

Fig. 4: A wireless cardiac monitoring system and its link to a database permitting physicians to review the sensed parameters.

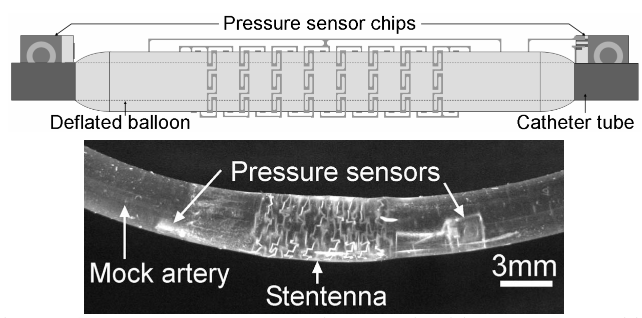

Fig. 5: (a: upper) A stentenna integrated with two sensor chips mounted on a deflated angioplasty balloon; (b: lower) a deployed device with the balloon removed.

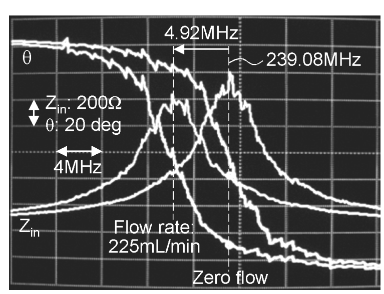

Fig. 6: Measured amplitude of input impedance in an external coil shifted near 239 MHz due to flow change.

Wireless tests with varying flow rate in a fluidic set-up demonstrated that an impedance peak was shifted down by increasing flow rate as shown in Fig. 6. The measured frequency shift indicated a reduction of 9-31 KHz per mL/min. increase in the flow range over 370 mL/min. [7]

C. *Micro Kelvin Probe with Integrated Actuator*

Kelvin probes are used to measure the contact potential difference (CPD) between materials, which cannot be measured directly using a voltmeter. One of the major applications is the characterization of solid-state devices. A probe is placed above the surface of a sample in close proximity, and an AC current is generated by dithering the gap where a CPD-induced charge is built up. The bias voltage which nulls the current indirectly determines the CPD.

The micromachined probe developed by µEDM included an actuator that provided the axial dither motion and a lead transfer beam for the probe (Fig. 7a) [9]. An electrothermal bent-beam actuator offered the dither motion with amplitude in the 10-µm range with drive voltages of a few volts [10]. The low voltage was important to minimize the coupling of the drive signal to the sense probe, while the large displacement permitted the dithering frequency and amplitude to be varied to suit the needs of the measurement.

The device needed an isolation plug that mechanically coupled the probe to the actuator while electrically and thermally decoupling them from each other. A large width of isolation was desired to minimize the capacitive feedthrough of the drive signal as well as the thermal noise from the actuator. The incorporation of dielectric components was achieved by a modified µEDM process shown in Fig. 8. The starting material was commercially available 30-µm-thick foil of MetGlas 2826MB, an alloy of primarily Ni and Fe (Fig. 8a). First, a traditional step was performed to define the shape of the epoxy plug in the workpiece (Fig. 8b). A two-part epoxy was applied onto the machined workpiece to fill the plug and then cured (Fig. 8c). A lapping step was performed for both sides of the workpiece to remove the excess cured epoxy (Fig. 8d). The rest of the features were defined by µEDM so that they were aligned to the epoxy plug, which was released by cutting along its edges (Fig. 8e). Finally, the finished part was attached to a glass substrate for testing (Fig. 8f). The fabricated device shown in Fig. 7b was used for non-contact sensing of pH of liquid inside microfluidic channels [9].

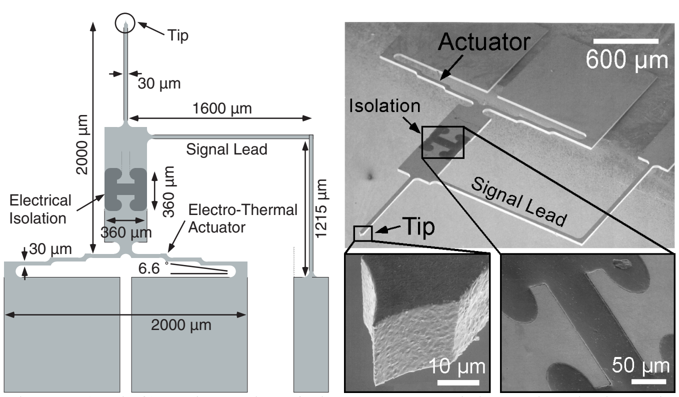

Fig. 7: (a: left) Schematic of the µEDM Kelvin-probe device; (b: right) a fabricated device bonded to a glass substrate.

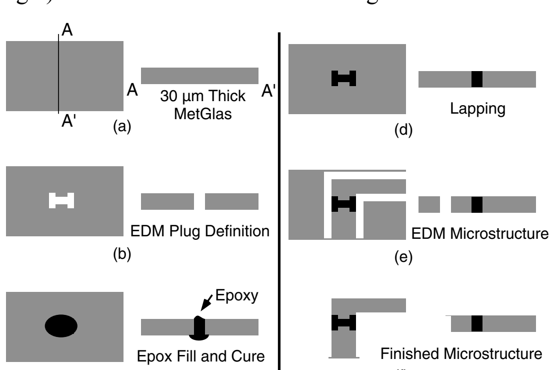

Fig. 8: The modified µEDM fabrication sequence.

D. *Intraluminal cuff for electromagnetic flow sensing*

The planar-to-cylindrical reshaping technique used for the stent fabrication was applied to the development of an intraluminal cuff for electromagnetic (EM) sensing of flow [7, 11]. The EM detection [12, 13] offers several attractive features such as direct and linear relationship between the output and flow, less dependence on cross-sectional flow profile, and mechanical robustness due to no moving parts used. EM flow sensors typically have two electrodes located on inner walls of the fluid channel. In the presence of a magnetic field, a voltage proportional to the flow velocity is developed between the electrodes.

The planar design of the ring cuff had a pair of meander bands comprised of 50 µm-wide beams, electrode plates, and two dielectric links that mechanically tied the bands but electrically insulated them from each other (Fig. 9a). This pattern was µEDMed in 50-µm-thick stainless steel foil, and then all the surfaces except front-side planes of the electrodes were coated with an insulating layer. (Without this, spatial averaging would reduce the voltage.) The dielectric links (of epoxy in this case) were created by a process similar to that for forming the isolation plug in the Kelvin-probe device shown in Fig. 8. The planar structure was mounted on a deflated balloon of a standard angioplasty catheter so that one of the bands was located above the balloon whereas the other band was below it. Figure 9b shows a device that was expanded inside a silicone tube with 3-mm i.d.. The device in a wired set-up showed a response that linearly and symmetrically increased or decreased depending on the orientation of a magnetic field applied externally (Fig. 10). The signal reading for this device was also extended to a wireless implementation using the stentenna [7, 14].

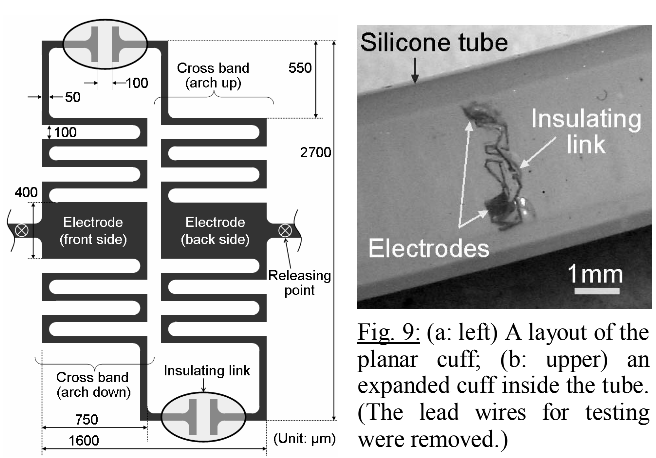

Fig. 9: (a: left) A layout of the planar cuff; (b: upper) an expanded cuff inside the tube. (The lead wires for testing were removed.)

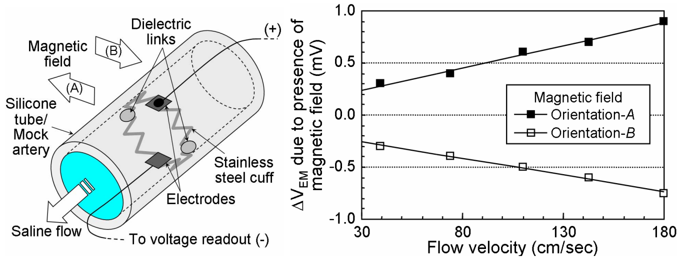

Fig. 10: (a: left) The fluidic measurement set-up with magnetic field of ~0.25 T applied at different orientations; (b: right) the measured results.

E. Cavity/Diaphragm-Less Capacitive Pressure Sensor

Micromachined capacitive pressure sensors typically use an elastic diaphragm with fixed edges and a sealed cavity in between the diaphragm and the substrate below. Since this configuration relies on the deflection of a relatively thin diaphragm against a sealed cavity, in some applications there is a concern for the robustness of the diaphragm and leaks in the cavity seal. To achieve mechanical robustness and simplify the structural configuration, it was aimed to eliminate the need for diaphragms and cavities from the sensor structure. This was approached by the configuration that consisted of two micromachined metal plates with an intermediate polymer layer [15]. Use of polymeric material soft enough to deform in a target pressure range allowed the thickness of the polymer, or capacitance of the parallel plate capacitor, to be dependent on hydraulic pressure that surrounded the device. The operation was demonstrated by the device with micromachined stainless-steel electrodes defined by µEDM and a liquid-phase polyurethane that was applied and solidified between the electrodes (Fig. 11a). The pressure monitoring was demonstrated by measuring frequency shifts in the LC tank that was fabricated by winding a copper coil on the sensor and bonding the terminals to the electrodes (Figs. 11b and 12). The combination of the selected materials potentially permits the direct use in corrosive or biological environment. As demonstrated in this effort, the use of µEDM promotes proper choice of materials that are compatible with particular environments for MEMS fabrication. This potentially allows us to circumvent constraints and problems associated with packaging of the devices, broadening application opportunities for MEMS.

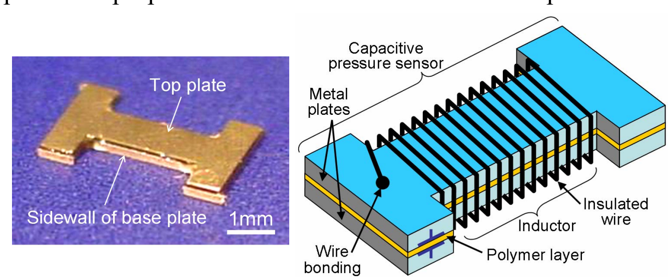

Fig. 11: (a: left) A polyurethane/stainless-steel capacitive pressure sensor fabricated by µEDM; (b: right) an LC tank form of the device.

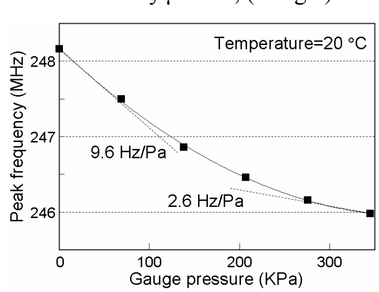

Fig. 12: Pressure vs. frequency measured with the LC tank device.

III. Conclusion

This paper presented recent study on the use of bulk metals and the µEDM technology for the development of MEMS devices. The effort demonstrated that the use of µEDM was effective to fabricate micromachined devices with both mechanical and electrical functionalities. The large material base of the technology enabled us to select appropriate engineering materials with particular characteristics such as plasticity, robustness, chemical inertness, and biocompatibility as well as cost-effectiveness with the use of commonly available stock metal foil. The result suggests that with the availability of the batch mode technique, µEDM can be a unique tool for the development and manufacturing of new types of MEMS that cannot be achievable with conventional MEMS fabrication methods.

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