Manufacturing and Tests of a Tubular Multilayer Dielectric Elastomer Actuator for an Impedance Pump

J. Chavanne¹, J. Haenni¹, T. Martinez¹, D. Moser¹, A. Walter¹, M. Almanza², F. Clavica^{1,3},
Y. Civet^{1*} and Y. Perriard¹

¹ École polytechnique fédérale de Lausanne (EPFL), Integrated Actuators Laboratory, Switzerland

² École normale supérieure Paris-Saclay, Laboratoire SATIE, France

³ ARTORG Center for biomedical engineering research, University of Bern, Switzerland

Abstract—During the last three decades, the field of electro active polymers (EAP) has seen a rapidly growing interest worldwide. These materials can exhibit many advantages, such as elasticity, high energy density, high dynamics and are often nicknamed artificial muscles. In particular, Dielectric Elastomer Actuator (DEA) are good candidates for soft pump. In this paper, we manufactured and test a tubular actuator further used in a valve-less pump. A mean net flowrate up to 0.11 L.min⁻¹ has been reached using a frequency of 8 Hz, a duty cycle of 0.5 and a voltage of 9.5 kV.

Index Terms— DEA, Flowrate, Impedance pump, Soft actuator.

I. INTRODUCTION

The pump plays significant role in the human body. For example, the heart and the lung are responsible of blood and air transportation in the most efficient way. Meanwhile, Heart Failure (HF) is a devastating disease that affects more than 11 million people in the US and Europe and more than 23 million worldwide [1]. Although heart transplant is a gold standard in highly selected patients suffering from severe heart failure, there is a large need of alternative effective therapies due to the shortage of heart donors. Not all patients need a transplant and the current possibilities to assist the failing heart are expensive and clinically speaking complex procedures. There are mainly two groups of devices: ventricular assist devices (VADs) and aortic-counter pulsation devices. In the first hand, current VADs are based on rotary pumps (axial and centrifugal pumps) characterized by a single rotating element that ensures a constant blood flow [2, 3]. In the second hand, the basic principle of counter pulsation is similar for all devices: it allows an increased coronary blood flow during diastole and unload the left ventricle during systole through rapid balloon deflation just before the aortic valve opens [4]. To this end, for intra-aortic-balloon-pumps (IABPs), a balloon is inflated in the lumen of the descending aorta during the diastolic phase of the heart (which translates into augmentation of the peak diastolic pressure) while presystolic deflation of the balloon decreases the

afterload. Notably, none of the devices that are commercially available are fully implantable [2, 5] and these devices are also used in patients with advanced heart failure. Therefore the need for supportive therapies should also focus on patients with moderate heart failure.

With the development of smart and functional materials, it is possible to propose credible alternative. Among them, DEAs have a simple operating principle, where mechanical stress is induced as result of applied electric field [6]. The activation leads in a decreased thickness and expanded area. In recent years, multilayer configurations have shown increasing relevance in the field of EAPs and promising interests for several applications [7]. However, fabrication is still challenging especially when comes the concern of fully embedded electric field for safety reasons.

In the other hand, impedance pumping technique [8] is a mechanism that generates flow in a compliant tube by repeatedly actuating the tube asymmetrically, without employing any internal valves, blades or other mechanism. The pump uses the reflection of the waves at the interface of the compliant tube and a stiffer one. Constructive interferences induce a non-zero net flow in the tube. The reflection at the interface is created by the change in rigidity between both tubes (Fig. 1). Studies of impedance pumping in a single tube have shown that the flow rate strongly depends on the actuation frequency, as well as on other parameters, such as actuator location and amplitude [9, 10, 11].

This work, therefore deals with the manufacture and tests of a multilayer tubular actuator potentially implantable in human body without risking any electric field in the surrounding tissue thanks to a relevant design for an impedance pump. In the first section, the manufacturing of the tubular dielectric elastomer actuator is presented. Then the static characteristics of the rigid tube and the tubular actuator respectively, are highlighted. The experimental set-up and associated measurements are finally discussed.

This work has benefited from the financial support of the Werner Siemens foundation.

Fig. 1. Typical topology for an impedance pump circuit [12].

II. MANUFACTURING PROCESS

The fabrication of the tubular dielectric elastomer actuator is based on stacking and rolling commercial films from Wacker Chemie™ (Elastosil® film 2030). The key point is a very thin (27 μm) conductive Fabric tape from Laird™ (Figure 2) which allows fully encapsulating the electric field even around the electrical pads and reducing significantly the access resistance. The structure of the DEA is composed of 2 layers of dielectric membrane and three layers of carbone based electrode [13]. The tape Laird™ is integrated in-between the carbone based electrode and the Elastosil® film to well-spread the electrical charges during activation and ensure thus a short time response. Two small shielded electrical wires are soldered directly onto the tape to ensure safe connection. The active area of the stack is 18.6 cm and 3.5 cm large. Then the full stack is rolled into a cylindrical hollow tube with an internal diameter of $D = 30$ mm and a full length of 80 mm. The total length of the tube is divided into two zones of same dimensions, respectively one active and one inactive area. The final device is composed of two membranes and two turns, resulting in a four layers structure. The electrical impedance at rest and low voltage of the tube is equivalent to a resistance of 30 k Ω in series with a capacitor of 1.6 nF (Fig. 3).

Fig. 2. Schematic cross section (a) and top (b) views of the stack (focus on the active area) before rolling.

Fig. 3. Picture of the tubular dielectric actuator at the end of the manufacturing process.

III. ELECTROMECHANICAL CHARACTERISTICS

A custom-made test bench has been built (Fig. 4), to measure the electromechanical behavior of the tubular DEA. The pressure-volume characteristic is obtained by measuring the pressure inside the tube and the volume thanks to a 2D-laser (assuming an axisymmetric deformation) while applying a voltage. The pressure is imposed by a pneumatic set-up composed of a moving piston, a linear motor and a screw nut system.

Fig. 4. Custom made test bench to obtain the pressure-volume characteristic of the Tygon™ tube and the tubular actuator.

First, we have measured the mechanical characteristic of the Tygon™ tube and the inactive DEA to validate our choice of material, in terms of rigidity for impedance mismatching. Fig. 6 clearly shows that the Tygon™ tube is more than 20 times stiffer than the DEA tube, fulfilling thus the requirements for impedance pump.

Fig. 5. Comparison of the Rigid tube (Tygon™) and soft tube (DEA) pressure-volume characteristic.

The DEA tube was then characterized at low frequency (quasi-static), following pressure profile as a triangle wave with maximum pressure reached in 20 seconds for different voltages ranging from 0 kV to 10 kV. As soon as the voltage increases, the DEA tube becomes softer as expected (Fig. 6).

The work provided by the DEA is related to the Pressure – Volume cycle in Fig. 6. Since the pressure is initially imposed and constant, the area is defined around this mean pressure. Then depending on the operating

conditions (voltage profile, i.e. maximum voltage and rising (or falling) edges) the cycles could be more or less large, so could the energy. A possible cycle is highlighted (red circle) in Fig. 6 for an initial pressure of circa 13 kPa.

Fig. 6. Characterization of the DEA for a range of applied voltages. The inactivated DEA (0kV) deforms the least under pressure.

IV. EXPERIMENTAL SET-UP

The DEA is then implemented as a valve-less pump that operates based on the principles of waves propagation and reflection [4]. Waves are generated in our case by the actuation of the DEA. The experimental set-up is composed of a rigid Tygon™ tube and the DEA including both the active zone for creating the deformation and an inactive zone needed for asymmetry. The deformation of the DEA and the flow are measured respectively with the 2D-laser and a flowprobe (Transonic, ME16PXL)) fitting the rigid Tygon™ tube. This avoids disturbing the created flow. The initial pressure could be adjusted thanks to a luer™ connector and is measured during the whole experiment (Baumer™, PBMN-258 1 2R A21 44621). The tubes are filled with DI water.

Fig. 7. Picture of the experimental setup, corresponding to the schematic of Fig. 1.

V. MEASUREMENTS

To generate a unidirectional mean flow, the DEA is periodically activated following a trapezoidal profile with a rising time and falling time of 15 ms. The frequency

and the duty cycle are varied to study the variation of the flow. Fig. 8 and Fig. 9 show the flow and pressure obtained for a pumping frequency of 8 Hz and a duty cycle of 0.5 with an activation voltage of 9.5 kV. The net mean flow is approximately of 0.11 L.min⁻¹.

Fig. 8. Example of a flow curve for a pumping frequency of 8 Hz and a duty cycle of 0.5.

Fig. 9. Example of a pressure curve for a pumping frequency of 8 Hz and a duty cycle of 0.5.

Fig. 10 shows that the net flow depends strongly on the actuating frequency and the duty cycle. This is similar to what has been observed in the literature [9, 10, 11]. The biggest mean flowrate reaches 0.086 L.min⁻¹ and is obtained with an actuating frequency of 8 Hz and a duty cycle of 0.7.

Fig. 10. Net mean flow in the loop according to the frequency activation and different duty cycles. The initial pressure was set to 13 kPa.

As expected, the initial pressure plays a significant role in the performance of the system (Fig. 11). The higher the pressure is, the bigger is the flowrate. This is linked to the intrinsic deformation of the DEA we could see in Fig. 6. At higher pressure, the radial deformation, and thus the modified volume is bigger compare to lower pressure. The maximum net mean flow could reach up to $0.11 \text{ L} \cdot \text{min}^{-1}$ for an initial pressure of 15 kPa.

Fig. 11. Net mean flow in the loop according to the frequency activation and different initial pressures in the loop. The duty cycle of the input signal is 0.5.

VI. CONCLUSIONS

We have manufactured a tubular DEA taking into account safety considerations for further use in human body. Tests (Activation profile and voltage, frequency, location of DEA versus the soft tube...) have been performed to validate the approach of a DEA based impedance pump, also known as Liebau pump. We have obtained a net mean flowrate of $0.11 \text{ L} \cdot \text{min}^{-1}$. This shows that dielectric elastomer technology could bring new perspectives for impedance pump where bulky and rigid actuators are always used. This pumping mechanism could thus pave the way to soft alternatives for medical application.

It is clear that the impedance pump mechanism relies on a huge number of parameters such as elasticity, frequency, duty cycle and impact location inherent to the pumping principle in a complicated manner. But the DEA itself is also designed for very specific working conditions (pressure, voltage). That is why, an analytical model, compared to the empirical studies proposed here, would be very helpful to investigate deeply the phenomena in such complex system. For example, implementing a model which uses electrical transmission lines (to our knowledge, this has been never investigated in the literature) could allow to precisely describe the propagation and reflection of pressure waves.

ACKNOWLEDGMENT

We thank Felix Wey for the pictures.

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