EUROPEAN MECHANICAL SCIENCE
e-ISSN: 2587-1110
Research Paper

A Battery Powered on-Chip Peristaltic Pump for Lab-On-A-Chip Applications

Sinan Gucluer[1*]

[1]Department of Mechanical Engineering, Aydın Adnan Menderes University, Turkey

Abstract

In this work, a low-cost and practical peristaltic pump is demonstrated using 3D printing and an open-sourced microcontroller platform. The peristaltic pump is designed to be compatible with polydimethylsiloxane (PDMS) microfluidic chips which are also fabricated through replica molding method using 3D printed molds. A thin layer of PDMS which was bonded on top of the microfluidic chip is designed to have a circular-shaped channel to be aligned with the circular arrangement of steel ball bearings. The entire system is designed to be portable and capable of producing metered fluid flow in small-scale devices. The developed device is characterized to provide adjustable fluid flow control between 1.7 μL/s to 23 μL/s which is suitable for many on-chip applications. Overall, a low-cost, portable and simple-to-fabricate peristaltic pump is presented.

Keywords: microfluidics, peristaltic pump, fluid manipulation, 3D printing

1. INTRODUCTION

In the last two decades, miniaturization has become the driving force to engineer more practical and low-cost systems in medical and technological applications [1]. Especially with the advancement of micro and nanotechnologies, and their simplified and more commonly available fabrication methods, microelectromechanical systems have emerged in many applications of automobile, telecommunication, biomedical instrumentation, and robotics industries [2,3]. There has been also a significant advancement in materials research that is investigating more reliable and low-cost materials for a wide range of applications [4,5]. Both with the new materials and advanced microfabrication tools, miniaturized systems including microfluidics have become available. Microfluidics is a sub-field of the microelectromechanical systems in which the standard micro and nanofabrication tools of the semiconductor industry such as photolithography and wet/dry etching procedures are adopted [6]. After the introduction of the soft-lithography and replica molding techniques, the field of microfluidics has spawned numerous on-chip devices for performing various tasks including sample processing and disease diagnostics [7]. Flow rate control in microfluidic systems is an important requirement for these applications. While the motivation and the goal of this new field are to reduce the cost of testing, analysis and diagnostics in biomedical research, and eliminate the dependence on central facilities, the requirement of peripheral elements such as pumps and flow regulators to drive the microfluidic lab-on-a-chip devices hinders their wide-spread implementation.

Various technologies have been developed to provide fluid manipulation on small scale for lab-on-a-chip applications [8,9]. In a general sense, the microfluidic pumping devices are divided into two groups, namely active and passive pumping platforms [10]. In the passive fluid manipulation scheme including surface tension and capillary-based approaches, no external power sources are required which is one of the main advantages of these types of microfluidic pumps [11]. However, it is not possible to provide any fluid rate control with the passive devices which are also application-specific and do not provide very precise fluid flow control. On the other hand, the active microfluidic pumps use external forces to control the fluid flow which provides fluid flow rate [12]. There are various types of active pumps including magnetic field-driven pumps, acoustofluidic pumps, electric field-driven pumps, laser power pumps, and peristaltic pumps [13]. In the magnetic pumps, an external magnetic field is applied to rotate an internal structure or particle which in turn can actuate fluid in microfluidic channels [14]. Another type of magnetic pump is magnetically actuated cilia that are implementing magnetic cilia beating motion to induce fluid actuation [15]. While it is easy to control the rate and direction of fluid flow in most of the magnetic field-driven pumps, these types of pumps require an exter-

* Corresponding author
Email: sgucluer@adu.edu.tr

European Mechanical Science (2021), 5(4): 201-205
doi: https://doi.org/10.26701/ems.876597
Received: February 8, 2021
Accepted: March 31, 2021

nal magnetic field strong enough to couple with the internal structure and generate rotation/actuation. Besides, the cilia-based magnetic pumps are not very effective to generate a uniform flow rate. Similarly, an electric field is labored in microfluidic fluid pumping by implementing external electric fields, but peripheral systems to generate the external electric field are generally bulky and expensive [16]. Laser beams are utilized in microfluidic platforms to drive fluid motion with precision owing to the resolution and high power of the laser systems [17]. While the laser beam-driven fluid pumps enable such high dexterity in terms of precision, they depend on very complex and costly optical setups which are very problematic for lab-on-a-chip systems that are meant for point-of-care and low-cost applications. A different type of active pump technology, acoustofluidic pumps rely on external equipment such as signal generators and power amplifiers to drive piezo actuators which are coupled to a microfluidic channel to enable fluid pumping [10]. In addition to the external field driven microfluidic pumps, peristaltic microfluidic pumps have also been widely explored as an active means of microfluidic fluidic pumping [18–23]. These devices commonly use a pneumatic actuation of arrays of chambers in a microfluidic channel to drive fluid flow. These devices generally require multilayer microfabrication steps due to their complex geometries [13]. Besides, their corresponding pneumatic actuation mechanisms are usually well-regulated pressurized inert gas supplies. Based on the limitations and shortcomings of the existing microfluidic fluid pumping approaches, there is still a demand for a simple-to-fabricate and easy-to-actuate microfluidic pump which can be adopted in lab-on-a-chip platforms.

Herein, a low-cost microfluidic peristaltic pump is demonstrated. To eliminate the complex fabrication requirements of lithographic mold preparation, the microfluidic molds are printed using a commercial-grade 3D printer. 3D printing significantly reduced the cost of the system which is critical to enable the end product to low-resourced researchers. Polydimethylsiloxane (PDMS) microfluidic channels are prepared with the soft-lithography process. Fluid actuation is enabled by a battery-powered stepper motor which in turn rotates 3.15 mm diameter ball-bearings housed in a 3D printed holder. Overall, the presented microfluidic pump is simple, open-sourced, and versatile in fluid flow rate and direction control. This device can be potentially adopted in numerous lab-on-a-chip applications for enabling battery-powered microfluidic fluid control.

2. MATERIALS AND METHODS

To fabricate the peristaltic pumping device, motor holder, microfluidic chip holder, bearing ball assembly and the entire electronic housing parts (Figure 1a) are designed and printed in a commercial-grade fuse deposition modeling (FDM) type 3D printer (i3 Mega, Anycubic, China). FDM printers are commonly available to consumers due to their relatively low-cost. As for the printing material, polylactic acid which is also referred to as PLA filament with 1.75 mm diameter is used. PLA is an easy to print and relatively safer biodegradable material. An open-sourced slicing software, Cura, is used to generate G-codes for the printer. 100 % infill, 205 °C nozzle temperature, 0.4 mm nozzle diameter, 60 °C print bed temperature, 100 % cooling, 50 mm/s print speed, 0.1 mm layer height, and zig-zag filling pattern are used as the printing parameters. For the microfluidic chip, a master mold is printed using PLA (Figure 1b). Polydimethylsiloxane (PDMS) is mixed with a 10:1 ratio and poured into the PLA mold. After curing at 45 °C for 12 hours, the PDMS microfluidic channel is pealed from the mold, and the inlet and the outlet are punched using a reusable biopsy punch. Then, a 1 mm thick PDMS layer is bonded on top of the PDMS channel using a home-made hand-held high-frequency plasma generator. After the plasma treatment and contact bonding process, the PDMS closed channel is baked at 60 °C for 12 hours for better bonding.

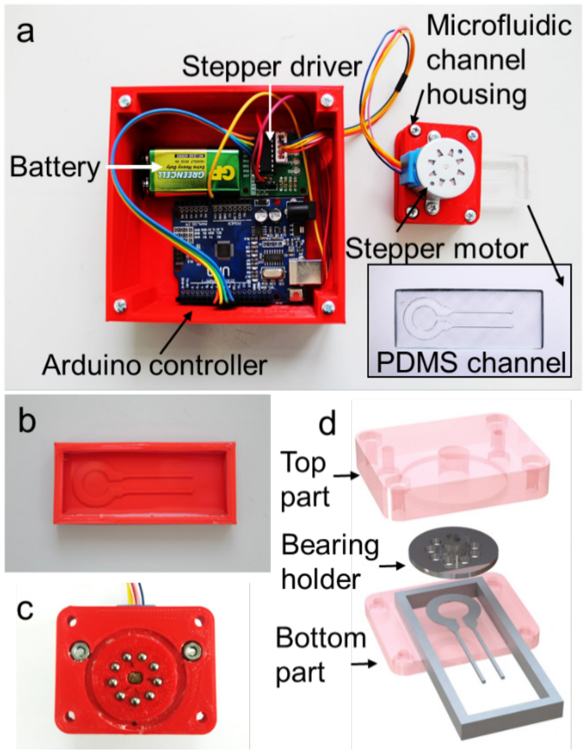

Figure 1. Developed peristaltic pump assembly and schematic. a) Electronic controller unit and the peristaltic pumping sections are shown. The inset is the polydimethylsiloxane (PDMS) channel. b) PLA mold for the PDMS is shown. c) ball bearing housing assembly is shown. d) Microfluidic chip holder and ball bearing housing assembly are schematically depicted.

To drive the fluid, nine 3.15 mm diameter stainless steel ball bearings are placed into a 3D printed housing assembly which is inserted into the shaft of the stepper motor on the reverse side (Figure 1c). The PDMS chip holder and bearing ball housing assembly are designed to be aligned so that the ball bearings have coincided with the circular section of the microfluidic channel (Figure 1d). the microfluidic channel dimensions are given in Figure 2. The circular section of the microfluidic channel is designed to be wider than

doi: https://doi.org/10.26701/ems.876597

the diameter of the ball bearings to include the thickness of the thin PDMS membrane. The main components and the work-flow diagram of the peristaltic pump are shown in Figure 3. The system is composed of an electronic control unit, a stepper motor (28BYJ-48), and a microfluidic chip. The stepper motor is chosen due to its lower cost compared to the other stepper motors including Nema series motors. 28BYJ-48 Stepper motor has 4096 and 2048 steps per revolution for half and full step modes, respectively. In this work, it is used in the half-step mode which results in a 0.0879° step angle. The flatted shaft of the motor made it easier to couple to the 3D printed ball-bearing housing. The stepper motor is controlled by a ULN2003 stepper motor driver controlled by an Arduino Uno board. Arduino boards have become the power-house of open-source hardware and development kits in the last decade due to their low-cost and simple programming requirements. The entire system can be supplied with a 9-volt battery to obtain a portable system. For the characterization of the peristaltic pumping system in the experimental conditions, an adjustable DC power supply is used.

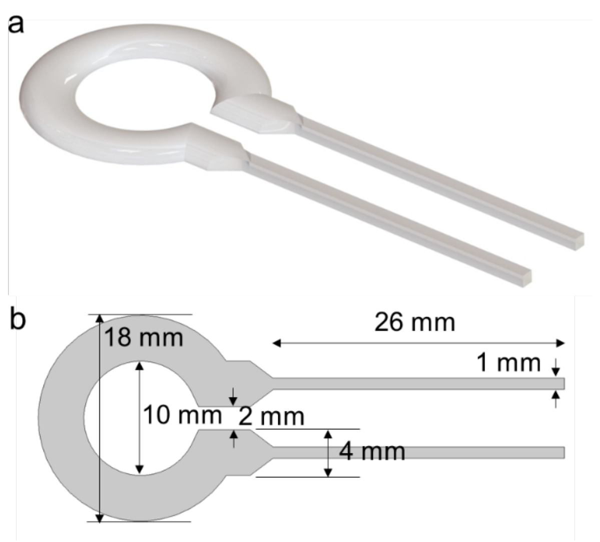

Figure 2. Microfluidic channel dimensions. a) A 3D rendered schematic of the fluidic path. b) actual dimensions of the fluidic channel.

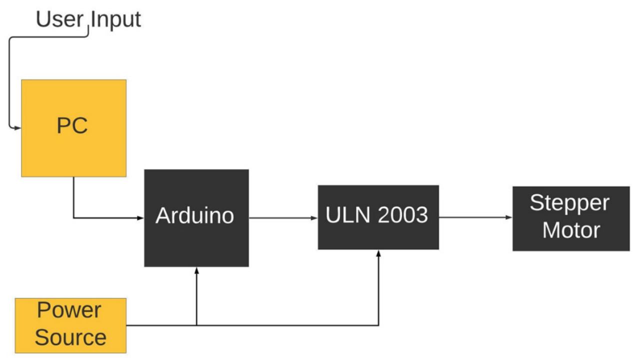

Figure 3. Primary elements and the work-flow of the peristaltic pump.

3. RESULTS AND DISCUSSION

In the working principle of the peristaltic pump, the rotation rate (RPM) of the stepper motor is used as the externally controlled variable to adjust the flow rate. Flow rate direction can also be controlled by reversing the motor rotation direction. RPM and motor rotation direction is controlled by the Arduino board. The programming requirement for these tasks is very simple and easy to adjust for specific needs. Figure 4 includes the simple code for the Arduino for a specific RPM input. The RPM input of this specific style of the stepper motor is adjusted by the rotation step parameter. In the Arduino code, first a commonly used stepper library called “AccelStepper” is included to incorporate the required definitions and commands into the Arduino environment. After that, motor pins are defined concerning the pins of the Arduino board. Arduino and the motor interface type are selected to drive the stepper motor in half-step mode by the predefined interface type 8 in the library. Then, the motor rotation speed (RPM) is defined by the step count per unit time in the “setspeed” command. By changing the step count, stepper motor RPM values from 28 to less than 2 RPM are obtained.

```
// Include the AccelStepper library:
#include <AccelStepper.h>
// Motor pin definitions:
#define motorPin1 8 // IN1 on the ULN2003 driver
#define motorPin2 9 // IN2 on the ULN2003 driver
#define motorPin3 10 // IN3 on the ULN2003 driver
#define motorPin4 11 // IN4 on the ULN2003 driver
// Define the AccelStepper interface type; 4 wire motor in half step mode:
#define MotorInterfaceType 8
// Initialize with pin sequence IN1-IN3-IN2-IN4 for using the AccelStepper library
with 28BYJ-48 stepper motor:
AccelStepper stepper = AccelStepper(MotorInterfaceType, motorPin1, motorPin3,
motorPin2, motorPin4);
void setup() {
// Set the maximum steps per second:
stepper.setMaxSpeed(2000);
}
void loop() {
// Set the speed of the motor in steps per second:
stepper.setSpeed(500);
// Step the motor with constant speed as set by setSpeed():
stepper.runSpeed();
}
```

Figure 4. An example Arduino Uno code to control the stepper motor RPM.

The wiring of the Arduino Uno, stepper motor driver, and stepper motor is also arranged to be compatible with the code. The wiring diagram is shown in Figure 5. In the wiring, IN1, IN2, IN3, and IN4 of the driver is connected to pin 8, 9, 10, and 11 of the Arduino, respectively. The stepper driver and the Arduino board are supplied with a positive voltage (from 5 V to 12 V) and a negative ground terminal. It is also possible to power the Arduino with the PC USB connection, but Arduino one of the ground pins is still required to be connected to the same ground of the stepper driver. Powering both the Arduino and the stepper driver with a portable power source is more desirable for lab-on-a-chip applications in remote sites with limited resources as well as for point-of-care diagnostics. The system presented in this study targets a minimalistic and low-cost approach to enable simplicity and dexterity in fluid manipulation on small scale. Therefore, the components, their assembly, and operation are crafted to serve the needs of developing countries.

doi: https://doi.org/10.26701/ems.876597

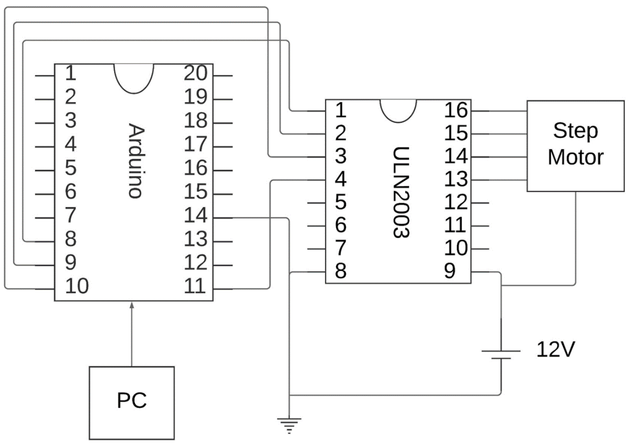

Figure 5. wiring diagram of the electronic components and the stepper motor of the control system.

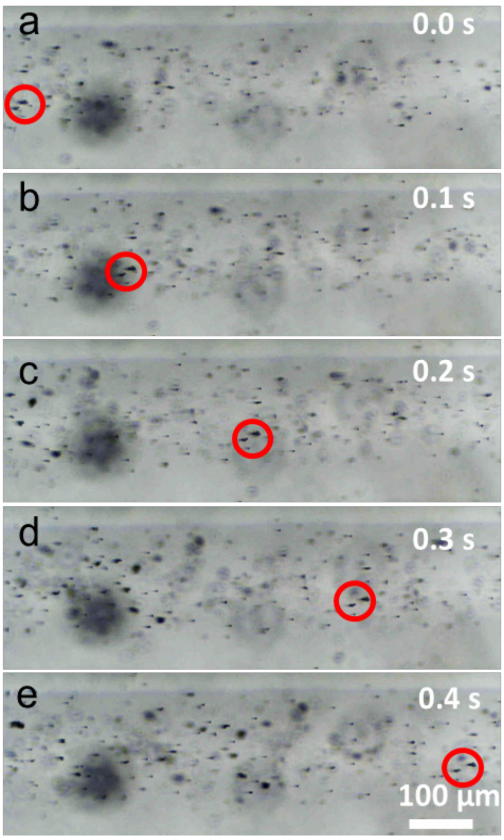

Figure 6. Characterization of flow rate via tracing suspended inclusions in the water medium. a-e) A cluster of polymer particles is tracked within the visible section of the microfluidic channel to calculate the fluid velocity and corresponding volumetric flow rates.

After completing the electronic control unit and the entire system assembly, the peristaltic pump is tested with water and polystyrene bead solution. Polystyrene beads with 5-micrometer diameters are introduced into the water medium to obtain flow characterization markers. The device characterization is completed using 3 different microfluidic chips. For each case, RPM values of the stepper motor are used as the controlled variable to obtain gradually increasing flow rates. The experiments with each RPM value are repeated at least three times for every microfluidic chip. For calculating the volumetric flow rates, polystyrene beads are tracked in collecting videos, and the bead velocities are calculated by dividing the travel distance by the time duration. Finally, the velocity of the fluid flow is multiplied with the inner cross-sectional area of the microfluidic channel to obtain the volumetric flow rate in microliters/seconds. For each RPM value, at least 6 beads are tracked, and the standard deviation of all the repeated experiments is used as the error bars in the data presented. An example of particle tracking and flow characterization is shown in Figure 6. As it is observed from the figure a polymer particle cluster is traced as a function of time. In this example, the calculated flow rate is 1.7 µL/s which corresponds to around 1.5 RPM stepper motor rotational speed.

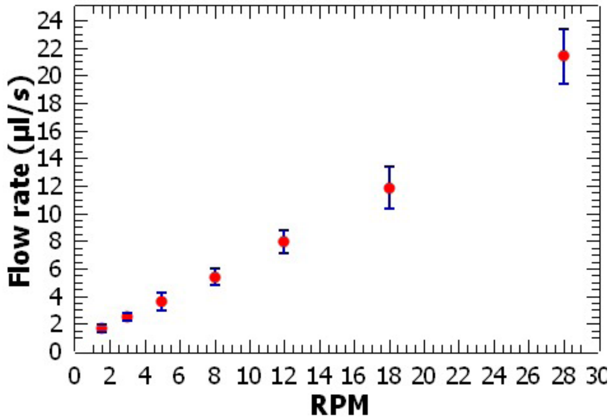

Figure 7. Volumetric flow rate versus rotational speed of the stepper motor. The standard deviations represent at least 18 repeated measurements for each RPM value.

4. CONCLUSION

In this study, a portable and low-cost peristaltic microfluidic pump is presented for providing controllable fluid flow manipulation for lab-on-a-chip applications. Considering the increasing need for more affordable and accessible medical care and disease testing, point-of-care devices and on-chip sample processing are becoming even more important in today's world where global outbreaks and pandemics are rising. To achieve, the goal of rapid testing and diagnostics using the lab-on-a-chip approaches, the fundamental necessity is to provide flow-controlled fluid manipulation in these on-chip devices. The peristaltic pump presented in this work does not rely on complex cleanroom microfabrication methods which are critical for enabling wide-spread adaptation of this device for the researchers in the developing world where resources are limited. A simple 3D printing tool, which is commonly available even as a house-hold item, is implemented to fabricate various parts of the system from a low-cost material. Furthermore, the master mold for the PDMS soft-lithography process is also 3D printed as a single-layer architecture. The PDMS membrane is simply poured into a petri dish and cured. Overall, the entire device fabrication can be done in any laboratory with commonly found items and tools. Besides, the control and the actuation of the peristaltic pump are provided by an Arduino Uno

doi: https://doi.org/10.26701/ems.876597

board and a low-cost stepper motor, respectively. Arduino Uno has wide-spread usage and is adopted by even middle school students to be applied in coding and project building. Finally, the system presented here can be powered by a simple 9 V battery to obtain a portable system that does not rely on peripheral devices or other infrastructures. Characterization of the pumping performance of the device yielded a flow rate interval between 1.7 μL/s to 23 μL/s which is adequate for enabling on-chip bioanalysis and disease testing. With its low-cost, simplicity, and flow control dexterity, this peristaltic pumping system can potentially be adapted to many lab-on-a-chip applications.

REFERENCES

[1] Franke, T.A., Wixforth, A., (2008). Microfluidics for miniaturized laboratories on a chip. Chemphyschem : A European Journal of Chemical Physics and Physical Chemistry. 9(15): 2140–56. doi: 10.1002/cphc.200800349.

[2] Ozcelik, A., (2019). Atomic Layer Deposition (ALD) of Vanadium Oxide Thin Films. Turkish Journal of Electromechanics and Energy. 4(2): 13–8.

[3] Kaynak, M., Ozcelik, A., Nama, N., Nourhani, A., Lammert, P.E.P.E., Crespi, V.H.V.H., et al., (2016). Acoustofluidic actuation of in situ fabricated microrotors. Lab Chip. 16(18): 3532–7. doi: 10.1039/C6LC00443A.

[4] Yıldızhan, Ş., Çalık, A., Özcanlı, M., Serin, H., (2018). Bio-composite materials: a short review of recent trends, mechanical and chemical properties, and applications. European Mechanical Science. 2(3): 83–91. doi: 10.26701/ems.369005.

[5] Serin, H., Yildizhan, S., Ozcanli, M., Akar, M.A., (2018). Micro hardness and water absorption properties of Cotton/Epoxy Bio-Composite. Journal of Biotechnology. 280: S91. doi: 10.1016/j.jbiotec.2018.06.299.

[6] Ayan, B., Ozcelik, A., Bachman, H., Tang, S.-Y.S.-Y., Xie, Y., Wu, M., et al., (2016). Acoustofluidic coating of particles and cells. Lab Chip. 16(22): 4366–72. doi: 10.1039/C6LC00951D.

[7] Zhang, P., Bachman, H., Ozcelik, A., Huang, T.J., (2020). Acoustic Microfluidics. Annual Review of Analytical Chemistry. 13(1): 17–43. doi: 10.1146/annurev-anchem-090919-102205.

[8] Zhang, C., Xing, D., Li, Y., (2007). Micropumps, microvalves, and micromixers within PCR microfluidic chips: Advances and trends. Biotechnology Advances. 25(5): 483–514. doi: 10.1016/J.BIOTECHADV.2007.05.003.

[9] Akkoyun, F., Ozcelik, A., (2020). A Simple Approach for Controlling an Open-Source Syringe Pump. European Mechanical Science. 4(4): 166–70. doi: https://doi.org/10.26701/ems.769837.

[10] Ozcelik, A., Aslan, Z., (2021). A practical microfluidic pump enabled by acoustofluidics and 3D printing. Microfluidics and Nanofluidics. doi: 10.1007/s10404-020-02411-w.

[11] Walker, G.M., Beebe, D.J., (2002). A passive pumping method for microfluidic devices. Lab on a Chip. 2(3): 131–4. doi: 10.1039/b204381e.

[12] Wu, Z., Cai, H., Ao, Z., Nunez, A., Liu, H., Bondesson, M., et al., (2019). A Digital Acoustofluidic Pump Powered by Localized Fluid-Substrate Interactions. Analytical Chemistry. 91(11): 7097–103. doi: 10.1021/acs.analchem.9b00069.

[13] Au, A.K., Lai, H., Utela, B.R., Folch, A., (2011). Microvalves and micropumps for BioMEMS. Micromachines. 2(2): 179–220. doi: 10.3390/MI2020179.

[14] Chen, Z., Noh, S., Prisby, R.D., Lee, J.B., (2020). An implanted magnetic microfluidic pump for in vivo bone remodeling applications. Micromachines. 11(3): 1–10. doi: 10.3390/mi11030300.

[15] Zhang, S., Wang, Y., Lavrijsen, R., Onck, P.R., den Toonder, J.M.J., (2018). Versatile microfluidic flow generated by moulded magnetic artificial cilia. Sensors and Actuators, B: Chemical. 263: 614–24. doi: 10.1016/j.snb.2018.01.189.

[16] Studer, V., Pépin, A., Chen, Y., Ajdari, A., (2002). Fabrication of microfluidic devices for AC electrokinetic fluid pumping. Microelectronic Engineering. 61–62: 915–20. doi: 10.1016/S0167-9317(02)00518-X.

[17] Maruo, S., Inoue, H., (2007). Optically driven viscous micropump using a rotating microdisk. Applied Physics Letters. 91(8): 1–4. doi: 10.1063/1.2768631.

[18] Sönmez, U., Bekin, M., Trabzon, L., (2018). Design and Fabrication of Integrated Microchannel and Peristaltic Micropump System for Inertial Particle Separation. MATEC Web of Conferences. 153: 08002. doi: 10.1051/matecconf/201815308002.

[19] SOBH, A.M., (2008). Interaction of couple stresses and slip flow on peristaltic transport in uniform and nonuniform channels. Turkish J. Eng. Env. Sci. 32(2): 117–23.

[20] Srinivasacharya, D., (2008). The effects of wall properties on peristaltic transport of a dusty fluid. Turkish Journal of Engineering and Environmental Sciences. 32(6): 357–65.

[21] Hayat, T., Abbasi, F.M., (2010). Peristaltic Mechanism in an Asymmetric Channel with Heat Transfer. Mathematical and Computational Applications. 15(4): 621–37.

[22] Elangovan, K., Dar, A.A., (2017). Impact of an inclined magnetic field, heat generation/absorption and radiation on the peristaltic flow of a Micropolar fluid through a porous non-uniform channel with slip velocity. New Trends in Mathematical Science. 3(5): 227–44. doi: 10.20852/ntmsci.2017.198.

[23] Yang, Y.N., Hsiung, S.K., Lee, G. Bin., (2009). A pneumatic micropump incorporated with a normally closed valve capable of generating a high pumping rate and a high back pressure. Microfluidics and Nanofluidics. 6(6): 823–33. doi: 10.1007/s10404-008-0356-7.