Miniaturized soft centrifugal pumps with magnetic levitation for fluid handling

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INTRODUCTION

Centrifugal pumps account for more than 80% of the world's pump production (1) and have the capability of offering continuous flow and handling large volumes of fluids at high rotation speeds, large flow rates, and low maintenance requirements. They are one of the widely used mechanical systems to deliver liquids for various applications in industry, household, medicine, and scientific research. They are also essential components in many biomedical systems, such as ventricular assist devices (2–4), extracorporeal membrane oxygenation (ECMO) machines (5–8), and dialysis machines (8–10), to supply continuous and quantified flows, such as blood, nutrient, and insulin to millions of patients who are suffering from cardiovascular diseases and chronic kidney diseases. However, centrifugal pumps are predominantly constructed by mechanically rigid components that involve magnetically driven rotors coupled with impellers with shafts (11–13). They are much heavier and larger than many diaphragm-based pumps whose miniaturization has already been achieved by many previous researches (14–16).

Realization of wearable and miniature centrifugal pumps may notably change the lifestyles of patients by allowing them to perform daily activities while receiving treatment. In addition, optimized mechanical properties that are compatible with the soft mechanics of body tissues and organs can enable ergonomic adaption of the pumps with body contour. However, even the smallest and most advanced centrifugal pump (e.g., HeartMate 3) is still 200 g in weight, 5 cm in diameter, and 3.4 cm in height (17–19). Despite these parameters showcasing great technical achievement, they are still considerably large for long-term applications. Miniaturization and weight reduction of centrifugal pumps are very challenging when considering their complex structures but are very tempting in terms of their capability to handle large volumes of fluids for wearable applications. Soft materials (20–23) and flexible electronics (24–27) may yield mechanical systems with lightweight and excellent compatibility to biotissues (28–30). Despite lots of demonstrations in skin patches (31–33), soft robots (34–36), and implantable devices (37–39), fabricating conventional mechanical pumps with soft materials has seldomly been achieved, not to mention a rotational flexible structure in a liquid environment for fluid delivery.

Here, soft magnetic levitation micropumps (SMLMs) have been developed for wearable and implantable applications. Each of the pumps contains a unibody of an origami magnetic rotor and a silicon impeller, all of which are driven electromagnetically through a flexible printed circuit board (PCB) circuit. Additional magnetic membranes are adopted in the pumps to confine the position and the rotation of the impellers through the magnetic levitation effect, enabling a shaftless design and abrasionless operation. The implementation of soft materials enables SMLMs to be the smallest and lightest centrifugal pumps that are suitable for wearable and implantable applications. They are only 0.3 to 12.8 g in weight and 0.3 to 11.7 cm³ in dimension, offering a rotation speed of up to 1000 rpm and a flow rate of 30.6 ml/min at 0-mm water-level difference with an operating temperature of less than 37°C. They have been used to pump liquids with various viscosities ranging from 1 to 6 centipoise (cP), and their use in assisting dialysis, blood circulation, and skin temperature control has been demonstrated. Implantation of the devices into rats for more than 30 days has indicated excellent biocompatibility with no organ damage. The unique levitation design prevents direct contact between impellers and the surrounding structures, allowing elimination of friction force and mechanical wear. The development of SMLMs not only suggests the possibility to replace mechanically rigid and rotational components with flexible materials and circuits but also indicates the potential to achieve fully flexible artificial organs that may revolutionize health care and improve the well-being of patients.

RESULTS

The fundamental concept of SMLMs is depicted in Fig. 1A. They can be used to replace conventional fluid pumping equipment for renal dialysis, ECMO, ventricular assistance, and heart failure treatment in combination with peripheral components, such as oxygenators...
and dialyzers. In addition, by wearing directly on skin or daily clothing, the SMLMs can realize fluid delivery for microenvironment regulation. A typical SMLM consists of six major components, including a magnetic rotor, an impeller, six electromagnetic stators, two magnetic confiners, a fluidic chamber, and a flexible PCB circuit (fig. S1). The design, fabrication, and assembly processes have been shown in detail in text S1 and fig. S2. The magnetic rotor that is made of origami magnetic membranes with enhanced magnetism and definable magnetic polarities (40–43) forms a unibody with a silicone impeller for fluid propulsion. The stators constructed by six electromagnetic coils are connected into three phases (phases A, B, and C) and controlled by pulse-width modulation (PWM) signal generated by the flexible PCB circuit (fig. S3). The operation of the SMLM is realized by the combined effects of magnetic levitation and magnetic field coupling between the stators and the magnetic rotor (Fig. 1B). The top and bottom magnetic confiners have opposite magnetic polarities to the magnetic rotor, resulting in levitation of the magnetic rotor in a liquid environment and oscillation of the magnetic rotor between the upper and the lower boundaries to maintain dynamic equilibrium. A magnetic cylinder and a silicone cone are attached to the magnetic rotor to further centralize and stabilize its rotation. The magnetic force between the stators and the magnetic rotor leads to counterclockwise rotation of the impeller. Once the magnetic rotor is rotated by 15°, the coil phases will be switched immediately to guarantee continuous and stable rotation of the SMLM. A detailed switching strategy shown in fig. S4 indicates that a full switching cycle of the three-phase coils leads to 90° rotation for the magnetic rotor.

To adapt to different application scenarios such as wearing and implantation, multiple SMLMs with different dimensions and weights were prepared (Fig. 1C). The smallest SMLM (named as the small SMLM), which is useful for implantation in small animals, is
only $9 \times 9 \times 4 \text{ mm}^3$ in dimension and 0.32 g in weight, the medium pump is $18 \times 18 \times 6 \text{ mm}^3$ in dimension with a weight of 1.9 g, while the largest SMLM (named as the regular pump) is $36 \times 36 \times 9 \text{ mm}^3$ in dimension and 12.8 g in weight. The flexible PCB circuit is $33 \times 25 \text{ mm}^2$ in dimension and only 0.66 g in weight. The soft, miniaturized, and lightweight configurations of the SMLMs allow them to be readily wearable on a human body without interfering with the physical motion of the wearer (Fig. 1D).

Characterizations of magnetic and electromagnetic components

The essential component of an SMLM is the magnetic rotor. A typical magnetic rotor is $700 \mu \text{m}$ in thickness with eight teeth. It can be formed by cutting a circular soft membrane made of a composite of Nd$_2$Fe$_{14}$B microparticles and polydimethylsiloxane (PDMS) into a petal shape followed by alternate teeth folding and uniaxial magnetization (Fig. 2A). The mechanical properties of similar magnetic membranes have been characterized in previous publications (41–43). The origami approach allows programmable magnetic polarities within a single membrane and improved magnetic field density of 55 mT because of edge effect (Fig. 2B and fig. S5) (41–44). In addition, the magnetic rotor is highly flexible to withstand repeated bending, stretching, and twisting (Fig. 2C).

The fine-tuning of magnetic polarities and complex magnetic polarity design can be readily achieved by different folding approaches (42, 43). Compared with the one-step process to achieve complex magnetic polarities in the magnetic membrane, rigid rotors are very challenging to achieve, as fabrication of irregular rigid magnets and assembly of rigid magnets are very difficult. In addition, it is well known that blood cells will be subjected to the hemolysis effect in a flow field with large shear stress. Thus, the hemolysis of blood cells is one primary concern in the rotation pumps, such as centrifugal pumps and axial flow pumps. Compared with rigid materials, such as stainless steel, flexible and soft materials, such as silicone and polyurethane, have demonstrated reduced hemolysis rates when in contact with blood (45). In addition, the high rotation speeds also increase the chance of cell disruption because of the impact of rigid materials on the soft cell bodies. Furthermore, soft materials have also been used as coating layers for the housing and the impeller of centrifugal pumps to improve surface smoothness and reduce wear resistance (46, 47). Thus, soft rotor and soft propeller may reduce cell disruption by offering soft buffer layers to reduce cell damage.

The capability of different coils in generating various magnetic flux densities has also been investigated. For coils with a fixed outer diameter of 8 mm and a height of 0.6 mm, the magnetic flux density increases steadily with current (Fig. 2D). A coil with a wire diameter of 0.04 mm has a flux density of 121.5 mT at 0.8-A driving current, while that of 0.1 mm is only 32.5 mT. The larger flux density of coils with small wire diameters is due to larger numbers of turns. In addition, the flux density also increases with the outer diameter of the coils (Fig. 2E). Various coil geometries can be applied to construct SMLMs with different sizes. When considering balance between the heat generation during operation and magnetic flux density, typical SMLMs contain coils with a diameter of 8 mm, a height of 0.6 mm, and a wire diameter of 0.06 mm.

Performance of SMLMs under various operation parameters

Further investigation of the performance of SMLMs under varying operation parameters was conducted at rotation speeds ranging from 0 to 1000 rpm (movie S1 and fig. S6A). To evaluate the capacity of a single SMLM, flow rates at different rotation speeds with water level differences ranging from −30 to 30 mm were measured (Fig. 3A). A maximum flow rate of 30.6 ml/min at a water level difference of 0 mm can be achieved for a rotation speed of 1000 rpm. The maximum flow rate is only five to eight times lower than that required in the bulky clinic systems (48, 49). In addition, a flow rate of 54 ml/min can be achieved by decreasing the water level difference to −30 mm.
The flow rates of the SMLM have been compared with other miniaturized soft pumps (50–61), indicating the superiority of the SMLM present in this work (Fig. 3B). A more detailed comparison of major features between different pumps has been provided in table S1 (50–75). The pumps based on electrohydrodynamics, dielectric elastomer, and piezoelectrics are small in size and low in power consumption, but they require high working voltage and offer low flow rates. Three commercial centrifugal micropumps (HeartMate, DuraHeart, and HeartWare) have large flow rates, but they are also bulky and heavy with fully rigid components. The SMLMs here demonstrate excellent balance in flow rate, working voltage, scalability, operation temperature, and weight, making them preferable for implantable and wearable applications.

Water level differences of a single pump and pumps connected in serial or parallel were measured under varying rotation speeds (Fig. 3, C and D; movie S2; and fig. S6B). For a single pump, the water level difference increases from 9 to 30.7 mm with the rotation speed from 500 to 1000 rpm. When two pumps are connected in series, the water level difference nearly doubles, reaching 58.7 mm at 1000 rpm (Fig. 3C), while the flow rate reduces slightly to 25.1 ml/min at the water level difference of 0 mm (Fig. 3D). A similar water level difference of 30.3 mm and a slightly higher flow rate of 33.7 ml/min

**Fig. 3. Characterization of the pumping capability of SMLMs.** (A) Changes in flow rate with water level differences measured at various rotation speeds. (B) Comparison of different types of micropumps created by soft materials with the SMLM in this work in terms of weights and flow rates. (C) Changes in water level difference with rotation speeds for SMLMs in single, series, and parallel configurations. (D) Changes in water level difference with flow rates for SMLMs in single, series, and parallel configurations. (E) Comparison of flow rates of an SMLM when pumping different fluids, such as water, blood, fat emulsion, and glucose solution with various viscosities. (F) Power consumption of the single SMLM at different rotation speeds. (G) Temperature changes on the top and the bottom surface of the SMLM when working continuously for over 30 min. Photo credit: Mingxing Zhou and Xian Huang, Tianjin University.
can be observed in the parallel connection (Fig. 3, C and D). These changes of water level differences and flow rates through connecting more pumps in series or in parallel suggest the approach to achieve different pumping capabilities through introducing different combinations of pumps to accommodate various demands.

The SMLMs can be used to pump biologically and medically important fluids such as water, blood, glucose solution, and fat emulsion with viscosities from 1 to 6 cP (Fig. 3E). Despite the decreasing flow rates with viscosities, the pump exhibits a flow rate of 15.5 ml/min for viscous fat emulsion, suggesting the potential of the pumps for delivering wide ranges of drugs. The SMLM consumes 1.15 W when operated at 500 rpm and 4.1 W at 1000 rpm (Fig. 3F), suggesting that the pumps may continuously operate at full speed for 9 hours with support from an ordinary power source with a capacity of 10,000 mAh.

Besides the regular pump, the small pump and the medium pump have also been characterized (movie S3). The experimental results in fig. S7A indicate that the small pump can rotate up to a speed of 500 rpm with a power consumption of 1.5 W at this speed (fig. S7B). However, because of the low rotation speed, the small pump cannot effectively pump water. In contrast, the medium pump can reach a rotation speed of 1400 rpm (fig. S7C) with a power consumption of 2.8 W (fig. S7D). The maximum water level difference obtained by the medium pump is 18 mm (fig. S7E and movie S4) with a flow rate of 9.16 ml/min at a water difference of 0 (fig. S7F).

The design parameters of the medium and small pumps were primarily obtained by scaling down the parameters of the regular pump. However, simply scaling down the parameters certainly cannot guarantee the optimum functions of the smaller pumps, as some nonlinear factors such as weight, spatial spacing, folding capability, and flow interaction may influence the performance of the pumps. In addition, pumps in smaller dimensions demand miniaturization of electronic components such as coils and circuits together with the minimized structures, bringing more challenges in the accuracy of fabrication and the tolerant errors in assembly. Furthermore, when the pumps become smaller, it is more complex to predict and adjust rotation and levitation of the rotors. For larger pumps, they may require thicker magnetic rotors, which are difficult to achieve through origami approaches by folding the membranes. Thus, despite that, scaling the parameters for the regular pumps to achieve larger and smaller pumps can be the starting point of the design. However, more complex optimization processes are still required to enhance the performance of the pumps.

Further experiments have been conducted to demonstrate the capability of the SMLMs in operating at different tilt angles. The investigation has both measured the capability of the SMLMs to start at different tilt angles and maintain rotation at different tilt angles. The results show that the pump can start at a maximum tilt angle of 180° for low rotation speeds between 100 and 900 rpm (movie S5). However, the pump can only start to rotate at a maximum tilt angle of 90° when the rotation speed is 1000 rpm (movie S6). After completion of the starting process, the pumps can work at tilt angles between 0° and 180° at rotation speeds ranging from 100 to 900 rpm. In contrast, the tolerance range of tilt angles reduces to 0° to 120° when the pump is rotating at a higher speed of 1000 rpm (Fig. S8 and movie S7). The underlying reason may be caused by the imbalance of magnetic force, gravity, lift force, and buoyancy at a higher rotation speed. Current configurations of the pumps allow the pumps to operate at broad tilt angles, indicating the effectiveness of the magnetic confiners in maintaining the rotation of the rotors.

The SMLMs have been able to withstand small deformation caused by bending, compression, and stretching. There is no physical contact between the internal rotor and external fluidic chamber because of the magnetic levitation effect. Thus, pressure applied directly on the chamber may first cause deformation of the chamber, while the rotor remains undeformed. Theoretically, the package of the pump can be bent upward (fig. S9A) and downward (fig. S9B) to a level with a radius of curvature of 55 mm. Afterward, the rotor may contact with the chamber wall and stop rotating. In actual experiments, an SMLM was fixed on a poly(methyl methacrylate) plate to study the influence of bending, striking, and stretching during its operation (movie S8). The results indicate that the SMLM can maintain a constant rotation speed of 1000 rpm with a maximum radius of curvatures of 62.5 and 57.5 mm, respectively, when bending upward and downward under a force of 14.27 N. Other deformations caused by compression and stretching only influence the package of the pumps, while the rotor can still operate under such deformations. At the same time, less than 1.25% changes in the resistance of the three-phase coils can be observed, resulting in insignificant changes in the driving current and the rotation speeds. Although different pressures will be applied to the human body in practical applications, it is difficult to cause bending of the SMLMs to a radius of curvature of 57.5 mm when implementing them as wearable or implantable systems.

Heat generated during the operation of the pumps was also characterized. The temperature of the system was only 34°C after operating at a full speed of 1000 rpm for 30 min and maintained stably afterward (Fig. 3G). This temperature is lower than the body temperature, suggesting that the pumps will not cause any discomfort to human skin. The temperature observed on the top surface rose from 27.3°C to 29.6°C after working for 1 min and then dropped to 27°C after 30 min. Because of the close contact between the coils and the fluid, the heat can transfer from the coils to the flowing fluid, which eventually brings the heat outside the system. The minimized heat accumulation within the pump is favorable for wearable application, leading to stable operation of the pump for a long period of time.

To study the influence of extreme temperatures on the operation of the SMLMs, an SMLM was fixed on a Peltier module that offered a temperature range from −15°C to 60°C when the rotor in the SMLM rotated at a fixed speed of 1000 rpm. Temperature values of three selective points labeled as M1, M2, and M3 are shown in fig. S10A and movie S9. When the Peltier module controlled the temperature to change from room temperature to 60°C or −15°C and returned to room temperature, the pump can operate during the entire process. Very small temperature changes are observed on point M3 because of continuous water flow that accelerates heat dissipation and exchange (fig. S10, B and C). As the SMLMs are based primarily on PDMS, which has a thermal expansion coefficient of 3.1 × 10⁻⁴°C⁻¹, when the SMLMs are operating within an environmental temperature ranging from −15° to 60°C, the changes in dimension will be less than 0.0013% when the temperature changes by 75°C. As the rotor is magnetically levitated inside the chamber of the SMLM, the small volume changes have little influence on the rotor to maintain constant speeds. In addition to the volume changes, different environmental temperatures inevitably change the resistance of the magnetic coils and the viscosity of the pumping fluids. The driving
circuit may need to integrate more functions such as constant current driving and flow rate monitoring in the future design to tackle the issues. In practical applications, as the pumps intimately contact with the skin or body tissues, it is expected that the fluctuation of the temperature may be very minimum.

**Dynamic measurement of the magnetic rotor**

The motion of the magnetic rotor is governed by the equilibrium of multiple forces, such as gravity, buoyancy, drag force, lift force, and magnetic force, leading to unique dynamic motion that involves rotation, vertical vibration, and procession. The vibration of the magnetic rotor and the wobble of the magnetic cylinder were measured respectively using a noncontact displacement sensor (Fig. 4, A and B). The measured results of the magnetic rotor vibration exhibit a decreased vibration amplitude from 0.78 to 0.09 mm with an increased rotation speed from 200 to 1000 rpm (Fig. 4C and fig. S11, A to C), suggesting that the magnetic rotor becomes gradually stable and has the tendency to rotate along a vertical axis (Fig. 4E). The trajectories of point $c_1$ at various rotation speeds are shown in Fig. 4F. At low rotation speeds of 200 and 400 rpm, no vibration occurs in the $z$ direction, suggesting that the bottom cone directly contacts with the bottom surface, and the magnetic rotor is not in the magnetic levitation state because of low lifting force (fig. S12A). However, large circular motion with procession radii of 0.31 and 0.3 mm caused by the wobble was observed. This phenomenon compares similarly to the rotation of a spinning top, which undergoes procession and rotation at the same time and wobble at low speeds. However, as the rotation speed increases, the increase in hydrodynamic pressure and more stabilized magnetic levitation force enable the magnetic rotor to gradually align upright and vibrate dynamically in the $z$ direction (fig. S12B), resulting in more stable and smooth rotation. Consequently, the vibration amplitudes decreased from 0.11 mm at 600 rpm to 0.004 mm at 1000 rpm, while the procession radius decreased to 0.15 mm at 1000 rpm.

To better visualize the rotation of the magnetic rotor, the coordinates of the dynamic trajectory of the measuring points ($Q_1$ and $N_1$) and fixed point ($c_1$) were calculated (detailed in text S2). The Euler angle or inclination angle ($\theta$) becomes smaller with increased rotation speeds, indicating that the rotor tends to rotate along a vertical axis (Fig. 4E). The trajectories of point $c_1$ at various rotation speeds are shown in Fig. 4F. At low rotation speeds of 200 and 400 rpm, no vibration occurs in the $z$ direction, suggesting that the bottom cone directly contacts with the bottom surface, and the magnetic rotor is not in the magnetic levitation state because of low lifting force (fig. S12A). However, large circular motion with procession radii of 0.31 and 0.3 mm caused by the wobble was observed. This phenomenon compares similarly to the rotation of a spinning top, which undergoes procession and rotation at the same time and wobble at low speeds. However, as the rotation speed increases, the increase in hydrodynamic pressure and more stabilized magnetic levitation force enable the magnetic rotor to gradually align upright and vibrate dynamically in the $z$ direction (fig. S12B), resulting in more stable and smooth rotation. Consequently, the vibration amplitudes decreased from 0.11 mm at 600 rpm to 0.004 mm at 1000 rpm, while the procession radius decreased to 0.15 mm at 1000 rpm.

**Fig. 4. Dynamic measurement of SMLMs.** (A) A schematic of displacement measurement of the magnetic rotor teeth and the magnetic cylinder by displacement sensors observed on the x-z plane of a three-dimensional ordinate system. The dashed purple lines represent the actual trajectory of the magnetic rotor. (B) A diagram of the top surface of the magnetic cylinder observed on the x-y plane. The dashed purple lines represent the actual trajectory of the magnetic cylinder. (C) Changes in the vibration amplitude of the magnetic rotor teeth with a rotation speed from 0 to 1000 rpm. (D) Changes in the wobble amplitude of the magnetic cylinder with a rotation speed from 0 to 1000 rpm. (E) Calculated inclination angle ($\theta$) of the magnetic rotor at different rotation speeds. (F) Calculated trajectory of the top center point ($c_1$) of the cylinder at different rotation speeds. (G) Back-EMF detected in phase A coils at a rotation speed of 1000 rpm. (H) Induced EMF generated in phase A coils at various rotation speeds. (I) Induced EMF in three-phase coils at a rotation speed of 1000 rpm.
The induced electromotive force (EMF) voltages due to the coupling between the magnetic rotor and the coils can be used as the signal for synchronizing the driving circuit or directly captured when using the pump reversely as a generator. Back EMF (back-EMF) voltage of less than 2.4 V was generated as compared with a driving voltage of 7.2 V (Fig. 4G). Through the detection of back-EMF, the timing of phase switching can be adjusted through zero-crossing detection to optimize the control program, replacing coercive phase switching and improving energy efficiency. When using the SMLM reversely as a hydroelectric generator, induced EMF generated in the coils under varying rotation speeds was measured. The induced EMF generated in single-phase coils (phase A) changes in the form of a sinusoidal wave and increases from 0.021 to 0.045 V with the rotation speed from 600 to 1000 rpm (Fig. 4H). Meanwhile, induced EMF voltage in three-phase coils (phases A, B, and C) under the rotation speed of 1000 rpm shows consistent and similar sinusoidal waves (Fig. 4I). With the assistance of an external transformer and a rectifier, the output from the SMLM can be directly converted to DC voltage.

**Simulation of SMLMs**

The properties of SMLMs have been simulated from different aspects. The distribution of the magnetic field was first simulated using COMSOL at an initial state (AB phase on) with a rotation speed of 1000 rpm. The simulation results for SMLMs are shown in Fig. 5. The side view and the top view of the distribution of B surrounding an SMLM are shown in (A). Simulation results of total magnetic force applied on the magnetic rotor in the vertical direction are shown in (B). Simulation results of torques on magnetic rotor teeth generated by the coupling between the stators and the magnetic rotor are shown in (C). Distribution of flow velocities in three SMLM designs at a rotation speed of 1000 rpm are shown in (D). The flow rates of three SMLM designs as a function of rotation speeds are shown in (E). Distribution of differential pressure in three SMLM designs are shown in (F). The differential pressures of three SMLM designs as a function of rotation speeds are shown in (G).
angle of 0° (Fig. 5A). Repulsion occurs between the magnetic rotor and the confiners in the vertical direction as well as the top confiner and the cylinder in the radial direction. Mutual repulsion in these two directions facilitates magnetic levitation of the magnetic rotor in the chamber with no physical contact to the surrounding walls. The total magnetic force applied on the magnetic rotor in the vertical direction ranges from −0.02 to 0.04 N (Fig. 5B). The torques of the eight teeth coincide highly with each other with a 45° phase difference (Fig. 5C). Although the torque of each tooth becomes negative at certain angles, the total torque of the magnetic rotor remains positive with a maximum value of 0.45 mN/m, driving the magnetic rotor to rotate unidirectionally.

The contour of velocity and static pressure distribution of the SMLM with different blade numbers and impeller outlet angles were also investigated by simulation (Fig. 5, D to G). The static pressure and the flow velocity distribution of the SMLM with a rotation speed of 1000 rpm were simulated. The velocity and static pressure increase gradually along the radial direction and reached their maximum values at the periphery of the impeller because of the work done by the impeller, resulting in energy conversion from kinetic energy to static pressure energy in the volute (Fig. 5, D and F). The impeller with eight blades and an outlet angle of 60° offers a maximum flow rate of 134 ml/min at the volute outlet and maximum differential pressure of 1.4 Pa between the impeller inlet and the volute outlet (Fig. 5, E and G, and fig. S13). Increasing the number of blades and the outlet angles can improve the work capacity of the impeller and enhance the flow rate and the water level difference. Thus, the impeller with eight blades and an outlet angle of 60° is considered as the optimized design because of its highest flow rate and differential pressure. The simulated flow rate is four times larger than the experimental results (Fig. 3A), possibly due to the wobble and vibration of the rotor and the impeller in practical operation, leading to extra energy loss and low conversion efficiency.

Applications of SMLMs
To demonstrate the applications of SMLMs in health care, the capability of the SMLMs in accelerating diffusion rates was first demonstrated. The diffusion rates of iodine through the membrane of dialysis bags with and without pumping were first compared. The diffusion rates were determined by the time the starch in the dialysis bag achieves a stabilized blue color due to the reaction between the iodine and the starch (fig. S14A). The time to reach stabilization is 85 min when pumping the solution with 1000 rpm rotation speed as compared with 115 min without pumping (Fig. 6A). The 26% improvement is a result of an increased molecule exchanging rate across the dialysis membrane. Urea concentrations within a dialysis bag were monitored using a urease biosensor to mimic the situation...
of renal dialysis (fig. S14B). Because of the large concentration gradient caused by constantly renewing the dialysis solution, the urea concentration reached a stable value of 25 mmol/liter after pumping for 1.5 hours, which was twice faster than without pumping (Fig. 6B), indicating the effectiveness of pumping in reducing the concentration of toxins in biofluids. When pumping blood, the SMLM exhibited a minimized hemolysis rate, which was less than 3.2% at 1000 rpm (Fig. 6C), meeting the requirements of the international standards (eligibility criteria: <5%, ISO 10993-4). These experiments using different pumping solutions indicate the potential applications of the SMLMs in wearable biomedical applications that require constant pumping of medicine, dialysate, and blood.

The SMLMs can also be used to adjust the microenvironment of the human body by pumping cooled water (15°C) throughout the body (Fig. 6D). Temperature changes of selective points on skin (S1 to S4) and on the tubes (M1 to M4) are shown in Fig. 6E. It took approximately 6 min for the temperature to reach stabilized values. The temperature changes in M1 to M4 increased from 4.2° to 7.5°C because of different transport distances from the outlet of the pump to the tube. The temperature changes on S1 to S4 that were typically between 1° and 2°C were determined by the density of the surrounding tubes. The results suggest that it is possible to integrate multiple SMLMs directly on skin or in daily clothing to tackle severe environments and adjust body temperature.
In vivo biocompatibility

To demonstrate the biocompatibility of the SMLMs as wearable devices, a biocompatibility test of the SMLM on the skin was conducted. Parylene-C (3 μm in thickness), which is a U.S. Food and Drug Administration–approved material with a USP XXII, class VI biocompatibility rating, served as a conformal coating layer for the SMLM. A medical dressing (Tegaderm, 3M Medical) was used to fix the SMLM on the skin (fig. S15). A piece of PDMS, which is also known to be a biocompatibility material for skin, was placed next to the SMLM as a reference. Results after wearing the pump for 1 day on skin showed no obvious signs of maceration, blister, and skin stripping, indicating great biocompatibility of the SMLMs as wearable devices.

Biocompatibility studies were conducted by implanting SMLMs into the abdominal cavity of rats (Fig. 7A). No obvious abdominal bulge and discomfort were observed throughout the process. Changes in the food intake and the body weight of rats implanted with the SMLMs were compared with those in the control groups (Fig. 7, B and C). The experimental groups and control groups exhibit similar trends in the food intake and the body weight. The steady increase in the body weight indicates no negative effect from the implantation. In addition, the results of the complete blood count and blood chemistry tests provide a comprehensive understanding of the health of the rats. According to the P values (P > 0.05) of the independent samples t test, the average counts of white blood cells, red blood cells, platelets, lymphocyte, neutrophilic granulocyte, and the levels of hemoglobin show no significant differences between the experimental group and the control group throughout a study period of 4 weeks (Fig. 7D). Enzymes and electrolytes that serve as important indicators in the blood for organ-specific diseases also fall within the confidence intervals of control values. For example, alanine aminotransferase, urea nitrogen, aspartate aminotransferase, and creatinine were all at normal levels, indicating the absence of disorders in the liver, kidney, and spleen, and good overall health, respectively (Fig. 7E).

Histopathologic evaluation of tissues obtained from a control rat and a rat implanted with the SMLM for 4 weeks reveals the absence of inflammation, ischemia/tissue necrosis, or other architectural/histologic abnormalities in organs such as spleen, kidney, and liver in both rats (Fig. 7F). The normal daily behavior and health conditions of major organs after SMLM implantation suggest the possibility of in vivo medical treatment using SMLMs for diseases such as congestive heart failure. The SMLMs have demonstrated the capability to replace large fluid pumping equipment for wearable or implantable fluid handling.

Additional biocompatibility tests of complete systems that integrated with SMLMs, flexible PCB circuits, and coil cell batteries were also conducted. These systems were first encapsulated with parylene C (3 μm in thickness) and then implanted into the abdominal cavity of rats. PDMS cubes with the same size were also encapsulated with parylene-C and implanted into other eight rats to serve as control groups (fig. S16A). The results of the complete blood count and blood chemistry tests show no significant differences between the experimental group and the control group throughout the studies (fig. S16, B and C). Histopathologic evaluation of tissues obtained from a control rat and a rat implanted with the SMLM for 2 weeks reveals the absence of inflammation, ischemia/tissue necrosis, or other architectural/histologic abnormalities in organs (fig. S16D).

DISCUSSION

Miniaturized centrifugal pumps that are suitable for long-term wearable fluid pumping in various biomedical applications have been presented. These pumps that are 0.3 to 12.8 g in weight and 0.3 to 11.7 cm³ in dimension are constructed by soft materials and structures that are mechanically compatible with human bodies. The operation of the pump features magnetic levitation achieved through interaction between origami magnetic membranes and soft magnetic confiners, offering minimized surface abrasion and mechanical friction that are common in conventional centrifugal pumps. The magnetic rotors in the pumps can rotate at a speed up to 1000 rpm driven by three pairs of coils on flexible PCB circuits, leading to a maximum flow rate of 30.6 ml/min. They can pump various fluids with medical importance, such as blood, glucose solution, water, and fat emulsion, and are excellent in biocompatibility during long-term implantation. These soft centrifugal pumps that combine flexible electronics and soft materials are among the smallest and lightest rotational mechanical pumps. They may benefit wearable or implantable therapeutic systems for hemodialysis and cardiopulmonary failure, leading to innovative medical treatment approaches that can be conducted in a daily living environment.

Compared with the performance of the pumps, realization of magnetic levitation and rotation of the soft impeller is the priority of this study. Characterizations of the pumps have suggested potential directions to improve their performance in the future. The power consumption of the SMLMs may be reduced by introducing magnetic membranes with higher filling ratios of Nd₂Fe₁₄B nanoparticles to offer increased magnetic force when interacting with the electromagnetic field. In addition, the design of the rotor and the impeller may be optimized to enable a larger surface area to interact with the electromagnetic field. Furthermore, the gaps between the rotor and the electromagnetic coils can be further reduced by improving the assembly techniques of the pumps and tuning the structures of the magnetic confiners. The current driving circuit does not involve monitoring back-EMF or hall sensors, resulting in ineffective switching control of the electromagnetic coils. According to the features of monitored back-EMF, an improved circuit can be developed to achieve higher rotation speeds and larger flow rates with reduced power consumption. It can be expected that fully Soft magnetically levitated centrifugal pumping systems with high tolerance to tilt angles, high rotation speeds, high flow rates, and low power consumption can be eventually realized to replace current bulky mechanical pumps for wearable and implantable applications.

The stability of the SMLMs in response to different tilt angles may be improved by design with more magnetic confinement structures and closed impeller structures. A brief diagram and simulation results of the proposed device are shown in fig. S17, in which the magnetic polarities of individual components in the diagrams have been indicated by arrows. The mutual repulsion effect of the magnetic confiners and the rotor may lead to balance in force during the operation of the pump at various tilt angles. Simulation results about the distribution of the magnetic field within the improved design also confirm the concept of a more compact and magnetically confined pump (fig. S17). Besides the well-designed magnetic polarities in the pump, further investigation about the filling ratio of the magnetic particles in the membranes to achieve different repulsion forces and force balance during the operation of the pump may also be needed. As the realization of the new design as well as
the following animal experiments are still ongoing, the result of the improved SMLM system will be reported in a follow-up paper.

**MATERIALS AND METHODS**

**Fabrication of SMLMs**

All soft magnetic membranes, confiners, and cylinders were constructed by molding composites mixed with PDMS (SYLGARD 184, Dow Corning Corp) and NdFe₁₄B microparticles (Magnequench Co. Ltd.) at the weight ratio of 1:5. A magnetic membrane was then cut by a mold made of stainless steel to obtain a magnetic rotor with eight teeth. Teeth with odd numbers were folded inward, while teeth with even numbers remained unfolded. The folded magnetic rotor membrane was then magnetized by a magnetizer (ME-1225, Magele Technology Co. Ltd.), followed by unfolding the odd numbers of teeth to yield the magnetic rotor membrane with defined magnetic polarities. The impeller was achieved by molding through a stainless steel mold with predefined patterns. The fluid chamber was fabricated by molding individual components such as the top cover, the volute chamber, and the bottom base using PDMS. During the fabrication of the bottom base, the flexible PCB circuit with six coils was partially sealed inside the bottom base. These components can be bonded together through oxygen plasma treatment. The entire assembly process has been shown in movie S10.

**Simulation of SMLMs**

The simulation of the SMLM was performed using finite element analysis software COMSOL (COMSOL Inc.). For the simulation of the magnetic field distribution and torques of the magnetic rotor, the remanence of magnetic rotor was set as 0.3 T. Each coil was 400 in turns and had a wire diameter of 0.06 mm, an inner diameter of 1.2 mm, and an external diameter of 7 mm. The working coils were driven by a current of 0.2 A. To simulate the flow rate and the differential pressure, the diameters of the inlet and outlet were all set to 3 mm, and the pressure of the inlet and outlet were both set to 0 Pa. Other dimensions of the model were consistent with the actual device.

**Operation of SMLMs**

PWM driving signal was generated using an Arduino software program, and the voltage was supplied by an external DC power source (fig. S11). The speed was controlled by changing the frequency of the PWM signal. The amplitude of the PWM signal should be adjusted to avoid the situation of speed loss.

**Diffusion measurement**

Dialysis bags (7000D, YA1076, Solarbio Life Science Co. Ltd.) were used in both iodine and urea diffusion. The dialysis bags were first cut into small segments 4 cm in length and treated with 2% (w/v) sodium bicarbonate and EDTA (1 mmol/liter) (pH 8.0). For the diffusion measurement of the iodine solution, the beaker was filled with iodine solution (1 mmol/liter), while the dialysis bag was filled with 0.5 wt% starch solution. The color of the starch solution was recorded every 5 min with and without pumping new iodine solution into the beaker. For the diffusion measurement of the urea, an electrochemical sensor for monitoring urea was first prepared. Polyaniline (PANI) was electropolymerized by dipping a commercial Au working electrode into a solution containing 0.25 M aniline (A801023, Shanghai Macklin Biochemical Co. Ltd.) and 1 M HCl, followed by applying a potential sweeping from −0.2 to 1 V versus a commercial Ag/AgCl reference electrode at a scan rate of 0.1 V/s. The electrochemical sensor was then inserted into the dialysis bag that has been added with urease, which catalyzed the hydrolysis of urea to change the pH value in the dialysis bag. Concentration changes of the urea in the dialysis bag were detected by measuring the open-circuit potential of the PANI electrode versus the Ag/AgCl electrode using an electrochemical workstation (CHI760E, CH Instruments Inc.).

**Characterization of SMLMs**

A gauss meter (CH-1800, Beijing Cuihai Jiacheng Magnetic technology Inc.) was used to measure magnetic field intensity above the center of the coils with different wire diameters and currents. To measure the distribution of the magnetic field of the magnetic rotor teeth, the magnetic rotor membrane was first fixed on a rotation table, and then the gauss meter was placed above the teeth to record the magnetic field density while the rotation table was rotating. The average flow rates for SMLMs in single, series, and parallel configurations were all determined using a digital balance to read the mass of the pumped fluids for a time duration of 20 s. The temperature of the working SMLM was monitored by a thermal camera (Fotric 111, Fotric Inc.) over a measurement period of 30 min. For the measurement of the vibration amplitude of the magnetic rotor teeth and the wobble amplitude of the cylinder, two noncontact displacement sensors (LK-H025, Keyence Co. Ltd.) that were fixed according to the method shown in Fig. 4A measured the displacement of the teeth and cylinder of the SMLM at a measurement frequency of 5 kHz. Back-EMF in single-phase coils was detected by an oscilloscope when the magnetic rotor rotated at speeds of 600, 800, and 1000 rpm, respectively. For induced EMF detection, the magnetic rotor was first fixed on the commercial motor. The induced EMF voltage was then detected when the commercial motor rotated at the speed of 1000 rpm. In the hemolysis demonstration of the SMLM, 10 ml of sodium chloride solution was prepared for both the experimental group and the negative control group, while 10 ml of deionized water was prepared for the positive control group. Then, 0.2 ml of blood cells pumped out from the SMLM was added into the experimental group. Regular blood cells (0.2 ml) were added into negative and positive control groups, respectively. Three groups were all incubated at 37°C for 1 hour followed by centrifugation. Last, the supernatant of the three groups was taken out and measured by an ultraviolet (UV) spectrophotometer (EU-2600, Shanghai Onlab Instrument Co. Ltd.) at 545 nm. The hemolysis rate was determined by the ratio of the absorbance difference between the experimental control group and the negative control group to the absorbance difference between the positive control group and the negative control group. For human body temperature regulation, the SMLM was attached to the arm. Cold water at 15°C was pumped into the SMLM. A thermal camera fixed above the arm was used to monitor the temperature of the skin and the tube.

**Animal experiments**

All animal studies were conducted in Tianjin Institute of Radiation Medicine with accreditation number SYXX (M) 2014-0004 under the ethic and operation guidelines required by the institute. Twenty male Sprague-Dawley rats (10 rats for the control group and 10 rats for the implantation group) with weights between 250 and 290 g were provided by the institute. The miniaturized SMLMs with a
dimension of 9 × 9 × 4 mm³ and a weight of 0.32 g were sterilized by UV radiation overnight before implantation. The SMLMs were implanted into the abdominal cavities of the rats. The body weight and the food intake of the rats were monitored twice a week. Four weeks after the implantation of SMLMs, organs and blood were extracted from the rats. Complete blood count and blood chemistry tests on the blood samples were conducted at Tianjin Institute of Radiation Medicine. All explanted organs were stored in conical metal-free tubes at −20°C or in conical tubes with 10% buffered formalin for histology studies.

SUPPLEMENTARY MATERIALS

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REFERENCES AND NOTES

1. P. Girdhar, O. Moniz, S. Mackay, in Practical Centrifugal Pumps, P. Girdhar, O. Moniz, S. Mackay, Eds. (Newnes, 2005), pp. 1–17.
2. S. Saito, T. Sakaguchi, S. Miyagawa, H. Nishi, Y. Yoshikawa, S. Fukushima, T. Daimon, Y. Sawa, Recovery of right heart function with temporary right ventricular assist using a centrifugal pump in patients with severe biventricular failure. J. Heart Lung Transplant. 31, 858–864 (2012).
3. A. Wisniewski, D. Medart, F. H. Wurm, B. Torner, Evaluation of clinically relevant operating conditions for left ventricular assist device investigations. Int. J. Artif. Organs 44, 92–100 (2020).
4. Y. Lu, L.-f. Zhu, Y. Luo, Development and current clinical application of ventricular assist devices in China. J Zhejiang Univ Sci B 18, 934–945 (2017).
5. C. O'Brien, J. Monteagudo, C. Schad, E. Cheung, W. Middlebrook, Centrifugal pumps and hemolysis in pediatric extracorporeal membrane oxygenation (ecmo) patients: An analysis of extracorporeal life support organization (elsi) registry data. J. Pediatr. Surg. 52, 975–978 (2017).
6. T. Yamagishi, F. Kunimoto, Y. Isu, H. Hinohara, Y. Morishita, Clinical results of extracorporeal membrane oxygenation (ecmo) support for acute respiratory failure: A comparison of a centrifugal pump ecoo with a roller pump ecoo. Surg. Today 34, 209–213 (2004).
7. A. P. S. Thiera, T. Noel, F. Kristiansen, H. M. Karlsen, A. E. Fiane, J. L. Svennevig, Evaluation of oxygenators and centrifugal pumps for long-term pediatric extracorporeal membrane oxygenation. Perfusion 22, 323–326 (2007).
8. M. J. Santiago, A. Sanchez, J. Lopez-Herce, R. Perez, J. del Castillo, J. Urbano, A. Carrillo, The use of continuous renal replacement therapy in series with extracorporeal membrane oxygenation. Kidney Int. 76, 1289–1292 (2009).
9. B. J. Boulton, P. Kilgo, R. A. Guyton, J. D. Puskas, O. M. Lattouf, C. A. Cooper, J. D. Vega, M. E. Halkos, V. H. Thourani, Impact of preoperative renal dysfunction and hemolysis in pediatric extracorporeal membrane oxygenation. J. Thorac. Dis. 10, 316–323 (2011).
10. Y. Yamamoto, D. Yamamoto, M. Takada, H. Naito, T. Arie, S. Akita, K. Takeshi, Excellent skin temperature sensor and stable gel-less sticky eeg sensor for a wearable flexible healthcare patch. Adv. Healthc. Mater. 6, 1700495 (2017).
11. D. W. Kim, S. Baik, H. Min, S. Chun, H. J. Lee, K. H. Kim, J. Y. Lee, C. Pang, Highly permeable elastomers with tunable mechanical properties for magnetically actuated devices. Nat. Commun. 8, 15329 (2017).
12. T.-i. Kim, Y. Huang, M. C. Montana, J. P. Golden, M. R. Bruchas, R. W. Gereau IV, J. A. Rogers, Wireless, soft electronics for rapid, multisensor measurements of hydration levels in healthy and diseased skin. Proc. Natl. Acad. Sci. U.S.A. 118, e0203998118 (2021).
13. L. Xu, S. R. Gutbrod, A. P. Bonifas, Y. Su, M. S. Sulkin, N. Lu, H.-j. Chuang, K.-j. Jiang, Z. Liu, M. Ying, C. Lu, R. C. Webb, J.-s. Kim, J. I. Laughner, H. Cheng, Y. Liu, A. Ameen, J.-w. Jeong, G.-t. Kim, Y. Huang, I. R. Efimov, J. A. Rogers, 3D multifunctional integumentary membranes for spatiotemporal cardiac measurements and stimulation across the entire epicardium. Nat. Commun. 5, 3329 (2014).
14. D. Lu, Y. Yan, R. Avila, I. Kandelis, I. Stepien, M.-h. See, W. Bao, Q. Yang, C. Li, R. C. Haney, E. A. Waters, M. R. MacEwan, Y. Huang, W. Z. Ray, J. A. Rogers, Bioreosorbable, wireless, passive sensors as temporary implants for monitoring regional body temperature. Adv. Healthc. Mater. 9, 2000942 (2020).
15. D.-h. Kim, N. Lu, R. Gaffarian, Y.-s. Kim, S. P. Lee, L. Xu, J. Wu, R.-h. Kim, J. Song, Z. Liu, J. Viventi, B. de Graaff, B. Elolampi, M. Mansour, M. J. Slepian, S. Hwang, J. D. Moss, S.-m. Woon, Y. Huang, B. Litt, J. A. Rogers, Materials for multifunctional balloon catheters with capabilities in cardiac electrophysiological mapping and ablation therapy. Nat. Mater. 10, 316–323 (2011).
16. Y. Yamamoto, D. Yamamoto, M. Takada, H. Naito, T. Arie, S. Akita, K. Takeshi, Excellent skin temperature sensor and stable gel-less sticky eeg sensor for a wearable flexible healthcare patch. Adv. Healthc. Mater. 6, 1700495 (2017).
17. D. W. Kim, S. Baik, H. Min, S. Chun, H. J. Lee, K. H. Kim, J. Y. Lee, C. Pang, Highly permeable elastomers with tunable mechanical properties for magnetically actuated devices. Nat. Commun. 8, 15329 (2017).
18. H. Wang, G. Pastorin, C. Lee, Toward self-powered wearable adhesive skin patch with bendable microneedle array for transdermal drug delivery. Adv. Sci. 3, 1500441 (2016).
19. Y. Kim, G. A. Parada, S. Liu, X. Zhao, Ferromagnetic soft continuum robots. Sci. Robot. 4, eaa3729 (2019).
20. T. Xu, J. Zhang, M. Salehizadeh, O. Onenazh, E. Ameen, Millimeter-scale flexible robots with programmable three-dimensional magnetization and motions. Sci. Robot. 4, eaa44944 (2019).
21. W. Hu, G. Z. Lum, M. Mastrangelo, M. Sitti, Small-scale soft-bodied robot with multimodal locomotion. Nature 554, 81–85 (2018).
22. A. D. Mickel, S. M. Won, K. N. Oh, J. Yoon, K. W. Meacham, Y. Xue, L. A. McIlevard, B. A. Copits, V. K. Samimeni, K. E. Crawford, D. H. Kim, P. Srivastava, B. H. Kim, S. Min, Y. Shiuan, Y. Yun, M. A. Payne, J. Zhang, H. Jang, Y. Li, H. H. Lai, Y. Huang, S. Park, W. R. Gereau, J. A. Rogers, A wireless closed-loop system for optogenetic peripheral neuromodulation. Nature 565, 361–365 (2019).
23. S. L. Park, D. S. Brenner, G. Shin, C. D. Morgan, B. A. Copits, H. U. Chung, M. Y. Pullen, K. N. Sah, D. Davidson, S. J. Oh, J. Yoon, K.-j. Jiang, V. K. Samimeni, M. Norman, J. G. Grajales-Reyes, S. K. Vogt, S. S. Sundaram, K. M. Wilson, J. S. Ha, R. Xu, T. Pan, T.-i. Kim, Y. Huang, M. C. Montana, J. P. Golden, M. R. Bruchas, R. W. Gereau IV, J. A. Rogers, Soft, stretchable, fully implantable miniaturized optoelectronic systems for wireless optogenetics. Nat. Biotechnol. 33, 1280–1286 (2015).
24. W. Lu, W. Bai, H. Zhang, C. Xu, A. M. Chiragelli, A. Vázquez-Guardado, Z. Xie, H. Shen, K. Nandiyala, H. Zhao, K. Lee, Y. Wu, D. Franklin, R. Avila, S. Xu, A. Rwei, M. Han, K. Kwon, Zhou et al., Sci. Adv. 2021; 7 : eabi7203 27 October 2021
45. X.-W. Lu, W. Liu, Z.-Q. Wu, X.-H. Xiong, Q. Liu, W.-J. Zhan, H. Chen, Substrate-independent, fully flexible electromagnetic vibration sensors with annular field confinement origami magnetic membranes. Adv. Funct. Mater. 30, 2001553 (2020).

46. K. Luo, Y. Wang, H. Liu, M. Dular, J. Chen, Z. Zhang, Effect of coating thickness on a solid-liquid two-phase flow centrifugal pump under water medium. J. Mech. Eng. 65, 251–261 (2019).

47. A. Ahmed, X. Wang, M. Yang, Biocompatible materials of pulsatile and rotary blood pumps: A brief review. Rev. Adv. Mater. Sci. 59, 322–339 (2020).

48. W. Drukker, in Replacement of Renal Function by Dialysis: A Textbook of Dialysis, J. F. Maher, Ed. (Springer Netherlands, 1989), pp. 20–86.

49. H. Schiffl, S. M. Lang, R. Fischer, Ultra pure dialysis fluid slows loss of residual renal function in new dialysis patients. Nephrol. Dial. Transplant. 17, 1814–1818 (2002).

50. S.-C. Chan, C.-R. Chen, C.-H. Liu, A bubble-activated micropump with high-frequency flow reversal. Sens. Actuators A Phys. 163, 501–509 (2010).

51. B. Liu, J. Sun, D. Li, J. Zhe, K. W. Oh, A high flow rate thermal bubble-driven micropump with induction heating. Microfluid. Nanofluid. 20, 155 (2016).

52. X. Zhou, M. Gao, L. Gui, A liquid-metal based spiral magnetohydrodynamic micropump. Micromachines 8, 365 (2017).

53. T. Pan, S. J. McDonald, E. M. Kai, B. Ziaie, A magnetically driven pdms micropump with ball check-valves. J. Micromech. Microeng. 15, 1021–1026 (2005).

54. H. K. Ma, R. H. Chen, N. S. Yu, Y. H. Hsu, A miniature circular pump with a piezoelectric bimorph and a disposable chamber for biomedical applications. Sens. Actuators A Phys. 251, 108–118 (2016).

55. C.-W. Dong, L.-G. Tran, W.-T. Park, A polymer membrane electrolysis micropump powered by a compact wireless power transmission system. J. Mech. Sci. Technol. 35, 697–706 (2021).

56. P. Kawan, S. Leahy, Y. Sawai, A thin pdms nozzle/diffuser micropump for biomedical applications. Sens. Actuators A Phys. 249, 149–154 (2020).

57. Y. Zhou, F. Aimrouch, An electromagnetically-activated all-pdms valveless micropump for drug delivery. Micromachines 2, 345–355 (2011).

58. H. Genzler, R. Sheybani, P.-Y. Lu, R. L. Mann, E. Meng, An implantable mems micropump system for drug delivery in small animals. Biomed. Microdevices 14, 483–496 (2012).

59. B. Pečar, D. Križaj, D. Vratčnik, D. Resnik, T. Dolžan, M. Malek, Piezoelectric peristaltic micropump with a single actuator. J. Micromech. Microeng. 24, 105010 (2014).

60. F. A. Mohd Ghazali, C. K. Mah, A. AbuZaiter, P. S. Chee, M. S. Mohamed Ali, Soft dielectric elastomer actuator micropump. Sens. Actuators A Phys. 263, 276–284 (2017).

61. V. Cacucioiu, J. Shinbakte, Y. Kuwajima, S. Maeda, D. Floreano, H. Shea, Stretchable pumps for soft machines. Nature 572, 516–519 (2019).

62. J. Darabi, W. Haixia, Development of an electrohydrodynamic injection micropump and its potential application in pumping fluids in cryogenic cooling systems. J. Micromech. Microeng. 14, 747–755 (2005).

63. A. Homsy, S. Koster, J. C. T. Eijkel, A. van den Berg, F. Lucklum, E. Verpoorte, N. F. de Rooij, A high current density dc magnetohydrodynamic (mhd) micropump. Lab Chip 5, 466–471 (2005).

64. P. Xuanming, L. Bo, X. Dongmei, J. Sufang, in Proceedings of the 4th IEEE Conference on Industrial Electronics and Applications, ICEIA 2009. (2009), pp. 1199-1202.

65. O. C. Jeong, S. W. Park, S. S. Yang, J. J. Pak, Fabrication of a peristaltic pdms micropump. Sens. Actuator A Phys. 123–124, 453–458 (2005).

66. P. S. Chee, M. N. Minjal, P. L. Leow, M. S. M. Ali, Wireless powered thermo-pneumatic micropump using frequency-controlled heater. Sens. Actuator A Phys. 233, 1–8 (2015).

67. F. Goldschmidtboing, A. Doll, M. Heinrichs, P. Woias, H. J. Schrag, U. T. Hopt, A generic analytical model for micro-diaphragm pumps with active valves. J. Micromech. Microeng. 15, 673–683 (2005).

68. T. T. Nguyen, M. Pham, N. S. Goo, Development of a peristaltic micropump for biomedical applications based on mini lipca. J. Bionic Eng. 5, 135–141 (2008).

69. R. Shabani, H. J. Cho, A micropump controlled by ewod: Wetting line energy and velocity effects. Lab Chip 11, 3401–3403 (2011).

70. M. Matar, A. T. Al-Hallouhi, A. Dietzel, S. Büttgenbach, Microfabricated centrifugal pump driven by an integrated synchronous micromotor. Microsyst. Technol. 23, 2475–2483 (2017).

71. M. Shen, L. Dovat, M. A. M. Gijs, Magnetic active-valve micropump actuated by a rotating magnetic assembly. Sensors Actuators B Chem. 154, 52–58 (2011).

72. M. Q. A. Rusli, P. S. Chee, R. Ansar, K. X. Lau, P. L. Leow, Electromagnetic actuation dual-chamber bidirectional flow micropump. Sens. Actuator A Phys. 282, 17–28 (2018).

73. T. Nishinaka, H. Schima, W. Roethy, A. Rajek, C. Noinj, E. Wolner, G. M. Wieselthaler, The duraharev vad, a magnetically levitated centrifugal pump: The university of vienna bridge-to-transplant experience. Circ. J. 70, 1421–1425 (2006).

74. D. J. Farrar, K. Bourque, C. P. Dague, C. J. Cotter, V. L. Poirier, Design features, developmental status, and experimental results with the heartmate iii centrifugal left ventricular assist system with a magnetically levitated rotor. ASAIO J. 53, 310–315 (2007).

75. Y. Sawa, Current status of third-generation implantable left ventricular assist devices in japan, durarev and heartware. Surg. Today 45, 672–681 (2015).

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