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Review paper

A Comprehensive Study of Micropumps Technologies

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This paper emphasis on physical principles and fabrication of micropump, state-of-the-art micropumping technologies developed over the past fifteen years, highlighting their advantages. Micropump features such as miniaturization potential, actuation voltage and ease and cost of fabrication are compared and application-determining performance characteristics discussed.

Keywords: Micropump, Microfluidic, Fabrication, MEMS, Fluid delivery

1. INTRODUCTION

When studying a class of devices it is always useful to have a good understanding of potential target applications. In the case of microfluidic systems, common target applications include chemical and biological analyses, biological and chemical sensing, drug delivery, molecular separation, amplification, sequencing and synthesis for environmental monitoring. A microfluidic system is one where fluid flows in miniature devices. It makes biological assays more effective through reduced reagent quantities and shorter reaction time. It is relatively inexpensive and can be integrated with other functional miniaturized components. It can contribute to the precision control systems of industries such as automotive, aerospace and machine tools. Most microfluidic systems have two or three-dimensional microchannels through which fluid samples are pumped (often concurrently and in various mechanisms), controlled and manipulated [1].

Most microfluidic systems need a self-contained active pump of a size comparable with the volume of fluid to be pumped. The key considerations for them include their reliability, power consumption, actuation voltage, ease and cost of fabrication, biocompatibility and a dosing accuracy comparable with that of a fuel pump [2]. A typical micropump is a MEMS device; it is the actuation source through which a fluid sample (drugs and therapeutic agents) is transferred with precision, accuracy and reliability from a reservoir to the target [3]. Typical applications include drug delivery and biomedical pharmaceutical, environmental monitoring and even homeland security applications such as Micro Total Analysis Systems (μTAS) or Lab-on-a-Chip (LoC) and Point of Care Testing Systems (POCT); reliability and robustness of the micropump are thus essential [4]. In such diagnostic systems, integration and miniaturization are achieved by combining on a single chip or package, MEMS micropumps, biosensors and a controlled drug delivery system. An appropriate and effective amount of drug can be precisely calculated by the controller and released at appropriate time by the micro actuator mechanism. The controlled drug release includes localized and precise site-specific drug delivery. It has many potential benefits besides reducing side effects (e.g. fluctuating levels of the circulating drug) and increasing therapeutic effectiveness [5].

Development of micropumps usable in single or two-phase cooling of microelectronic devices has been challenging because microelectronics cooling is highly demanding in its flow-rate requirement [6]. Recent developments in miniaturization of these systems have enabled their application to chemical and biological analyses. An obvious advantage of miniaturization besides reduced form factor of the systems when, e.g., applied to (μTAS), is the yield improvement to performance (e.g., faster completion of assays) and reduced cost (e.g., through fewer manual interventions, decreased quantities of samples and reagents and fabrication cost and use of disposable substrates [7]. Miniaturization also aids system transportation very advantageous to some applications [8].

Space exploration can also benefit from micropump technologies. Transportation of miniature roughing pumps for mass spectrometer systems, for example, meets the lightweight requirement of spacecraft [9].

This article comprehensively reviews and compares the state-of-the-art in micropumping of fluids through microchannels where it will be focusing on physical principles, engineering limitation sand fabrication. The comparison criteria are made on miniaturization potential, actuation voltage, ease and cost of fabrication, working principles and key design parameters. The micropumps' suitability to various applications is also discussed.

2. MICROPUMPS CLASSIFICATION

MEMS definition causes miniaturized pumping devices fabricated by micromachining technologies to be called micropumps [10]. The efforts in developing micropumping technology to system specification and requirements have produced various types of micropumps, which will be discussed in this section including their working principles. Micropumps can be classified into two categories, either mechanical or non-mechanical micropump. Further research and recent additions of electrowetting and bubble-driven micropumps have modified the initial categorization. Table 1 classifies micropumps as per this paper's review [11]. A mechanical micropump need a physical actuator or a pumping mechanism while a non-mechanical micropump has to transform non-mechanical energy into kinetic momentum to drive fluid in the microchannels. The next section divides micropump into sub-categories [12].

Table 1. Classification of Micropumps

Classification of Micropumps		
Mechanical displacement		Non-Mechanical
Actuation Method	*Micropumping Technique*	
Electrostatic	-Vibrating Diaphragm	Magneto-hydrodynamic
Piezoelectric	-Vibrating Diaphragm -Peristaltic -Flexural Plate Waves	Electro-hydrodynamic
Thermo-pneumatic	-Vibrating Diaphragm -Peristaltic	Electroosmotic -AC -DC
Shape Memory Alloy (SMA)	-Vibrating Diaphragm	Electrowetting
Bimetallic		Bubble type
Ion conductive polymer film		Electrochemical
Electromagnetic	-Vibrating Diaphragm	

2.1. Mechanical displacement Micropumps

These use the motion of a solid (such as a gear or diaphragm) or a fluid to generate the pressure difference needed to move fluid. The most common mechanical displacement micropumps are diaphragm pumps. Their actuation mechanisms are varied. They need a physical actuator for the pumping and they have moving parts. The actuator has to run itself with dead volume (V_0) in the chamber. Fluid flow is achieved by oscillatory or rotational pressure forces. The oscillations create pressure (ΔP), which is a function of the stroke volume (ΔV) inside the chamber and produced by the actuator. The compression ratio is defined as [13]:

$$\varepsilon = \frac{\Delta V}{V_0} \qquad (1)$$

Examples of mechanical micropumps include piezoelectric, electrostatic, thermo pneumatic, electromagnetic, bimetallic, Ion Conductive Polymer Films (ICPF)and Shape Memory Alloy (SMA) [14].

2.1.1 Electrostatic Micropumps

These use electrostatic forces in their actuation mechanism. The pressure difference induced by the membrane deflection in the pump chamber forces fluid in the reservoir to flow in the microchannels. The electrostatic force is defined as “the electrical force of attraction and repulsion induced by an electric field" where like charges repel, unlike charges attract. The electrostatic force applied on the electrostatic plates can be expressed by the equation:

$$F = \frac{dW}{dX} = \frac{1}{2}\left(\varepsilon A \frac{V^2}{X^2}\right) \qquad (2)$$

with F being the electrostatic force of attraction, W the stored energy, ε the dielectric constant.

A the surface area of the electrodes, X the electrode spacing and V the applied voltage.

Fabrication of these mechanisms on an electronic chip is generally considered easy, but the electrostatic actuator has only a small stroke and the deflection of the diaphragm can be easily controlled by the applied voltage. The main advantages of electrostatic micropumps are low power consumption and fast response [15].

2.1.2 Piezoelectric Micropumps

The conversion of mechanical energy to electronic signal (voltage) and vice versa is called piezoelectric effect. A stress applied to such materials will alter the separation between the positive and the negative charges, causing surface net polarization. This piezoelectric actuation has a strain induced by an applied electric field on a piezoelectric crystal. The piezoelectric effect relates to the coupling between mechanical deformation and electrical polarization.

$$\varepsilon = s^E \sigma + dE \qquad (3)$$

with s^E being the compliance tensor and σ the stress, d the tensor of the piezoelectric-charge constant and E the electric field.

Piezoelectric mechanism finds common use in the reciprocating micropumps of drug delivery and other biomedical applications. Main advantages of piezoelectric actuators include large actuation force, fast response and simple structure. Their fabrication, however, is complex as is processing of piezoelectric materials. Another disadvantage is the comparatively high actuation voltage and small stroke [16].

2.1.3 Thermo-pneumatic Micropumps

In these types of micropumps, a chamber full of air is periodically and alternately expanded and compressed by a pair of heater and cooler. The periodic change in the volume of the chamber gives the membrane a regular momentum so fluid can flow out. This type generates relatively large induced pressure and membrane displacement. Two disadvantages are the need for the driving power to be maintained at a constant and specific level and the slow response. The thermo-pneumatic actuation has a thermally induced volume change and/or phase change of fluids sealed in a cavity with at least one compliant wall. The pressure increase in liquids is expressed as:

$$\Delta P = E\left(\beta \Delta T - \frac{\Delta V}{V}\right) \tag{4}$$

with ΔP being the pressure change , E the bulk modulus of elasticity, β the thermal expansion coefficient, ΔT the temperature increaseand $\Delta V/V$ is the volume change percentage [17].

2.1.4 Shape Memory Alloy (SMA) Micropumps

SMAs are metals that show two unique properties such as pseudo elasticity and Shape Memory (SM). The diaphragm of SMA micropumps is made mostly of Titanium/Nickel alloy (TiNi), which is a highly suitable material for micropump actuators because of its high recoverable strain and actuation force capability for large pumping rates and high operating pressures [18]. Its high work output per unit volume makes it suitable in sizes for MEMS applications. They can change shape when subjected to a stimulus. The SM effect involves a phase transformation between two solid phases: the austenite phase (at high temperatures) and the marten-site phase (at low temperatures). When heated to austenite start temperature, the material starts forming a single-variant austenite. If not subjected to mechanical constraint, the material will return to a pre-deformed shape, retained upon cooling back to marten-site phase. If subjected to mechanical constraint, the material will exert a large force while assuming a pre-deformed shape [19].

Main advantages of micro SMA pumps are high force-to-volume ratio, ability to recover large transformation stress and strain upon heating and cooling processes, high damping capacity, chemical resistance and biocompatibility. Their disadvantages are the need for specific SMA materials, relatively high power consumption and uncontrollable deformation of the SMA owing to temperature sensitivity. A more practical, effective and complex-characteristic design for TiNi film devices is required through multiple DOF and compact structures [20].

2.1.5 Bimetallic Micropumps

Bimetallic actuation functions differently on different Coefficients of Thermal Expansion (CTE) of materials. The diaphragm of bimetallic micropump is made of two different metals having different CTEs. Bonding mechanism of dissimilar materials and their subjection to temperature changes induce thermal stresses because the coefficients of the metals differ, providing a means for actuation. The implementation extremely simple with large forces generated but the deflection of a diaphragm is only can be achieved by thermal alternation. The key advantage is that bimetallic micropumps require relatively low voltages than the other types of micropumps. Their main disadvantage is their unsuitability to high-frequency operation [21].

2.1.6 Ion Conductive Polymer Film (ICPF) Micropumps

ICPFs are polymer MEMS actuators that can be actuated in aqueous environments with large deflection. They need lower input power than the conventional MEMS actuators.

They are actuated by stress gradient from the ionic movement due to an electric field. Their key advantage is fast response.

ICPF is composed of polyelectrolyte film with both sides chemically plated with platinum. The two films have high electrical conductivity. One diaphragm-end is fixed. An ICPF diaphragm can be controlled by bending it upside or downside in certain value of voltages applied to the electrodes within a certain duration. The presence of an electric field causes the ions in both sides of the polymer molecule chain to move to the cathode. Simultaneously, each ion with a positive charge will take some water molecules and move towards the cathode. The ionic movement causes the cathode to expand and the anode to shrink [22]. The presence of an alternating voltage signal will bend the films alternately. The applications of ICPF to delicate micro robots and micro manipulators that perform surgical operations have been reported.

ICPF actuators have advantages such as low driving voltage, quick response and biocompatibility. They can work in aqueous environments. Their major drawback is low repeatability in batch fabrication [23].

2.1.7 Electromagnetic Micropumps

A typical magnetically actuated micropump has a chamber with inlet and outlet valves, a flexible membrane, a permanent magnet and a set of drive coils. Either the magnet or the set of coils may be attached to the membrane. The strength of the magnet can be varied by changing the electric current flow through the coils. Current driven through the coils produces a magnetic field that creates attraction or repulsion between the coils and the permanent magnet which provides the actuation force. The electromagnetic actuation is realized by a solenoid plunger. The force developed by the actuator depends on the applied current and on the number of turns. A miniaturized electromagnetic actuator comprises a soft magnetic mass suspended by a spring beam and an external solenoid coil [24].

Their main features are high power consumption and heat dissipation. Their disadvantage is the difficulty in miniaturization, owing to the size of the required solenoid coil. Addressing the disadvantage is the integration of the magnets, the cores, or the micro coils, for compactness and smallness [25].

2.2. Non-mechanical Micropumps

This type of pump has to transform specific non-mechanical energy (electroosmotic, electro-hydrodynamic, magneto-hydrodynamic, electrowetting, etc.) into kinetic momentum to drive fluid sample into the microchannels. Non-mechanical pumps can be further categorized as being electrical, chemical, or magnetic. Electro and magneto-kinetic micropumps convert electrical and magnetic energy into fluid motion. The pumping is continuous, so the resulting flow is generally constant/steady. Electrokinetic micropumps often use an electric field to pull ions within the pumping channel, in turn dragging along the bulk fluid by momentum transfer due to viscosity. Magneto kinetic micropumps typically use Lorentz force on the bulk fluid to drive microchannel flow. Dynamic pumps do not usually have valves; they obtain directionality from the direction of the applied force [26].

2.2.1 Magneto Hydro Dynamic (MHD) Micropumps

MHD micropump concept is new. Among the earliest developed was by Jang and Lee in 1999. They used the principle of Lorentz force acting on the moving charges in a fluid. MHD refers to the flow of electrically conducting fluid in electric and magnetic fields. A typical MHD micropump structure is rather simple; microchannels, two walls bounded by electric-field-generating electrodes and two walls bounded by opposite-polarity magnetic-field-generating permanent magnets. Lorentz force is the driving source; it is perpendicular to both the electric field and the magnetic field [27]. MHD micropump pressure and flow rate are:

$$\Delta P = J_y B_x L$$
$$Q = \left| J_y B_x \right| \frac{\pi r_0^4}{8\eta} \qquad (5)$$

with J_y being the current density, B_x the magnetic flux density, L the distance between electrodes, r_0 is one half of the hydraulic diameter of a microchannel and η the viscosity [28].

MHD micropumps can easily pump any conducting liquids of range 1 S/m or most of the aqueous solutions used in biological applications. They can pump high-conductivity fluids, so are suitable for medical/biological applications. Their main disadvantage is the bubbles generated by ionization; it affects the flow rate. Bubble generation is reducible by reversing the direction of the applied voltage. An ac driving mechanism will improve their performance [29].

2.2.2 Electro Hydrodynamic Micropumps (EHD)

EHD pumps induce flow through use of electrostatic forces on dielectric liquids. The mechanism allowing transduction of electrical to mechanical energy in an EHD micropump is electric field acting on induced charges in a fluid. The flow of fluid is manipulated by interaction of electric fields with the charges they induce in fluid. Fluid must have low conductivity and be dielectric. The electric body force density F_e resulting from an applied electric field with magnitude *E* is:

$$F_e = qE + P.\nabla E - \frac{1}{2} E^2 \nabla \varepsilon + \frac{1}{2} \nabla \left[E^2 \left(\frac{\partial \varepsilon}{\partial \rho} \right)_T \rho \right] \qquad (6)$$

with q being the charge density, ε fluid permittivity, ρ fluid density, T fluid temperature and P the polarization vector [30].

The DC charged injection of EHD micropumps uses Coulomb force exerted on the charges between two permeable electrodes in direct contact with fluid to be pumped. Ions are injected into fluid by electrochemical reactions from one or both electrodes. EHD micropumps types are induction, injection, polarization, or ion-drag, each referring to the method of introducing the charged particles into fluid [31].

2.2.3 Electro-Osmotic Micropumps (EO)

They use the surface charge artificially developed via electrodes or spontaneously developed when a liquid comes in contact with a channel wall. Oppositely charged ions generated in fluid shield the surface charge. They can be manipulated by DC or AC electric fields. Both have been much researched in recent years [32].

2.2.4. DC Electro-Osmotic (DCEO) Micropumps

Their construction uses fused silica or glass capillaries with electrodes that provide an electric field along the channel length. In silica-based channels, when an electrolytic solution comes in contact with the channel wall, the surface silanol groups spontaneously deprotonate, leaving a negatively charged boundary. Application of DC electric field increases the force on fluid near the capillary wall through increased charge density. The charges respond to the electric field. Motion of fluid to the microchannel interior is instigated by viscous forces. In the solution, the induced surface charge attracts positively charged ions and repels negatively charged ions. An electric double layer forms along the capillary wall. The channel centre has neutral charge density [33].

Two main challenges to EO micropumps are: microchannel-blocking bubbles and low stall pressure (typical of EO micropumps with an open channel). The large currents generated in the open channel may produce bubbles; electrolysis and reactions at the electrodes produce ions that can contaminate the sample and generate the bubbles. High pressure can be built up by using very small channels or by keeping the channel densely packed. Some of the disadvantages of EO micropumps often are alleviated by controlling the packing of particles in the channel [34].

2.2.5. AC Electro-osmotic (ACEO) Micropumps

ACEO flow is a viable micro-scale pumping mechanism for conductive or electrolytic solutions. Unlike the deprotonation on the channel surface in DCEO flow, electrodes positioned on the channel boundary provide the charge necessary to establish an electric double layer. Asymmetric electrodes induce an electric field and draw the diffused layer charges along the electrode surface. The advantage of ACEO pumps includes possible achievement of high velocities for very small voltages less than 10V. Also, as voltages increase within the range, flow can be reversed, making this pump type bi-directional. Reverse-flow velocities have been found to be larger than forward flow velocities [35] .

2.2.6. ElectroWetting (EW) Micropumps

EW involves wettability change due to applied electric potential. In electrowetting, fluid is transported by surface tension, an interfacial force dominating at micro scale. Voltage is applied to the dielectric layer, decreasing the interfacial energy of the solid and liquid surfaces, causing fluid flow. Heating of the liquid is not required. Response is faster and power consumption is lower, both as compared with thermo capillary micropumps. Liquid metals contacting electrolytic solutions develop charged interfaces that act like capacitors owing to electrochemical reactions [36].

Liquid metal is more wettable (i.e., has lower surface tension) in the region of high charge density than it is in the region of low charge density. The surface tension is lower to the right than to the left of the mercury. The surface tension gradient induces motion to the right of the liquid metal. The pressure build-up by the CEW actuation is given by:

$$\Delta\rho_{CEW} = 2q_0\Delta\phi\left[\frac{1}{D}+\frac{1}{W}\right] \quad (7)$$

with $\Delta\emptyset$ being the voltage difference between the two ends of the liquid metal, q_0 the initial charge per unit area in the electrical double layer in the absence of the applied potential and D and W respectively the depth and width of the channel [37] .

There is increased interest to incorporate EW-based fluid delivery techniques in lab-on-a-chip applications. The principle uses an array of electrodes individually controlled to move droplets in any planar directions on a surface such that they could be introduced to other droplets in mixing and/or chemical reactions [38].

2.2.7. Bubble-type Micropumps

The pumping effect in these is based on the periodic expansion and collapse of the bubbles generated in the microchannels. The volume of the bubbles are expanded and collapsed periodically by a controlled voltage input. The volume change in the chamber is achieved through a diffuser/nozzle mechanism that also determines the flow direction. The main advantage of this micropump type is the possible mixing of two or more kinds of doses during the expanding/collapsing cycles. The main disadvantage is the frequent heating required, limiting application [39].

2.2.8 Electrochemical Micropumps

Their actuation force uses bubbles electrochemically generated in the microchannels. Application of a DC current electrolyzes the water between two platinum electrodes in a saline solution, generating gases and consequently a pressure that in turn moves liquid solutions inside the chip. This pump is very effective in pumping milliliter solution volumes. It also consumes little power [40].

The most common feature of electrochemical micropumps is the generation of bubbles by electrolysis in which water decomposes into hydrogen gas (H2) and oxygen gas (O2). A key component is a bubble reservoir filled with a redox electrolyte solution. It uses the bubble force generated by electrochemical reaction during electrolysis. It needs electrodes to supply the electricity, also fluid channels, electrolysis chamber (for bubble generation) and inlet and outlet reservoirs [41].

Design and construction of this micropump type is relatively simple. It can also be easily integrated with other microfluidic systems. Its limit is possible collapse (into water) of the generated bubbles, upsetting drug-release reliability [42].

3. EVALUATING MICROPUMPING TECHNOLOGIES

As have been described here and elsewhere, various possible pumping mechanisms have been proposed to meet various needs of micro scale flows. As pump sizes reduce to micro domains, the effect of centrifugal and inertial forces are generally limited. In some pumping techniques, with the decreased length comes increased force per unit volume. The large surface-to-volume ratios amplify the effects of the viscous forces, often becoming the dominant force. This section compares various pumping technologies qualitatively and quantitatively, especially concerning their target applications.

The first step to selecting a micropump technique for a given application is to determine the pumping requirements and the usage environment. A few applications are listed below. The references included indicate the pump types having been considered for those applications. This shall in no way imply superiority of the pumping solutions suggested [43].

3.1. Chemical and biological analysis

One requirement of these is the ability to reduce sample size or reagent quantities. Also, they must accommodate a wide range of desired flow rates and pressure drops. Their devices should cost little and be disposable to avoid contamination. Electrokinetic pumping has been commonly used for its device simplicity. However, despite the electro-osmotic pumping used, the flat velocity profile reduces sample dispersion. Use of pneumatic and other diaphragm micropumps have not been common owing to device complexity and the larger footprint of the pumps. Pneumatic pumps have been used in cell culturing and surface plasmon resonance (SPR) detection. Electrochemical pumps have been used in influenza subtype identification and sequencing and DNA microarray processing [44].

3.2. Thermal management techniques

Piezoelectric pumps have been used as thermal management solutions especially for their low power consumption. Relatively large flow rates are required to accommodate the high heat fluxes given that liquid flow is the heat transport mechanism. Self-cooled printed circuit boards (PCB) have also been developed wherein the pump and microfluidics are integrated into the PCB [45].

3.3. Insulin dosage control delivery

This application is expected to grow explosively owing to increased interest in in-situ medical treatment of diabetic patients. Insulin delivery devices generally do not need high flow rates, still, they should be capable of providing precisely-metered small doses, their flow rate capable of operating independently of back pressure and they should be bio-compatible. Constant flow over a range of environmental back pressures thus factors in their design and valving [46].

3.4. DNA hybridization

A limiting factor in DNA research is the long hybridization time required for strong detection signals. When the hybridization is static, only very small percentages of the reagent are introduced to the probes on the chip. Mixing can significantly reduce binding times and increase the reagent quantities presented to the probes. The bi-directional flow capability of AC electro-osmosis micropumps would allow the DNA or biological materials in solution to be driven back and forth over the probes or the measurement regions [47].

4. GUIDELINES TO MECHANICAL DISPLACEMENT PUMPS

When considering reciprocating or displacement pumps that exert oscillatory or rotational pressure forces on the working fluid through moving boundaries, the following are important considerations [48].

4.1. Susceptibility to bubbles

Micro scale displacement pumps often have a pump chamber, actuator and valves. Susceptibility to bubbles can be significantly problematic to these pumps and/or their valving. Peristaltic pumps can be bubble-tolerant and self-priming and provide bi-directional flows [49] .

4.2. Bi-directional flow

Peristaltic pumps are capable of bidirectional flows because diaphragm actuation provides the valving; with nozzle/diffuser or fixed-geometry flap valves, the flow is uni-directional. Depending on the application, reversible flow may be important to device operation [50].

4.3. Scaling Reduction

Pump diaphragm diameter affects pump performance; it reduces the maximum pressure. Scaling reduction may not be best for when a vibrating diaphragm pump is selected with a requirement for large-back pressures and the limit of a highly constrained space [51].

4.4. Actuation mechanism

Selection of the driving mechanism significantly affects pump operating conditions [52].

5. GUIDELINES TO NON-MECHANICAL PUMPS

Research in this has developed greatly and produced new possibilities. Electro-kinetic and magneto-kinetic pumps provide direct energy transfer to pumping power and generate constant/steady flows (as compared with oscillatory pumping), owing to the continuous addition of energy. Their performance, however, is often limited by the properties of the selected fluid. Also, some pump types need specific surface characteristics for flow generation. The next section outlines important considerations to selecting or designing dynamic pumps [53].

5.1. Flow profiles

EO pumps are characterized by a diffuse layer largely unaltered by the driving electric field and have a blunt and plug-like flow-profile, which can reduce sample dispersion. Dynamic pumps generally are not pressure-driven, so the resulting flow does not always have parabolic velocity profiles [54].

5.2. Fluid properties

Electro-osmotic pumps need a specific pH to create an electric double layer when surface deprotonation is used. In electrokinetic devices, fluid itself is crucial to flow generation. Fluid selection is rather limited to those that can provide the properties necessary to actuation. Pumping with higher-conductivity fluids with electrokinetic methods has been investigated (Fuhr et al., 1992; Wu et al., 2007) [55].

5.3. No moving parts

The motion typically required in many dynamic pump types to generate fluid motion comes from both static and dynamic potentials rather than from dynamic mechanical motion, allowing for pump designs with simple structures and no moving parts [56].

5.4. Pressure

MHD pumps generally have low pressures whereas EO pumps have high pressures. The high pressure in EO pumps is usually achieved by densely packing the channel with particles. The packed particle beds, however, create a tortuous flow path, causing flow to mix in the pump section. This may be undesired, depending on the pump location in the flow loop or flow path. Also, not all dynamic pump types have well-characterized back pressures that limit their application [57].

6. FABRICATION COST

The first MEMS micropump was fabricated through conventional technique but not until 1984 was a micropump based on silicon micro fabrication technologies is introduced. Since then it has attracted interest in its ability to considerably shrink the size of a fully functional pump. In 1990, Smits published the results of his work [58]. The micropump that he designed was a peristaltic pump comprising three active valves actuated by piezoelectric discs. The device was developed primarily for the used in controlled insulin delivery systems. It was to maintain blood sugar levels without frequent needle injections. Van Lintel et al. presented the first micro diaphragm pump with passive check valves in 1988. He is the one who reported the first attempt at fabricating silicon micropump with piezoelectric actuation. It was a three-layer set-up with two glass sheets enclosing an anisotropically etched silicon wafer [59].

The reciprocating displacement type micropump comprised a pump chamber, a thin glass pump membrane actuated by piezoelectric disc and passive silicon check valves to direct flow. This was the first reported work on successful fabrication of micropumps through micromachining technologies. The first micropump without valve was pioneered by Stemme in 1993 used nozzle or diffuser as the flow rectifier [60].

The material cost of silicon and elaborate micro manufacturing processes, however, burden the use of such silicon micropumps because they need expensive silicon micromachining facilities and processes. The use of plastics proves a great potential in lowering fabrication cost. Plastics are disposable, inexpensive and its mass-production manufacturing makes it even more competitive. Several micropumps have been developed through mass-production manufacturing [61].

Richter et al. recently developed a plastic micropump comprising five parts: one plastic pump body, one metal diaphragm and three piezoelectric ceramics. It overcomes the drawback of passive micro check valve through it's a simple valve actuation design: piezoelectric-actuated diaphragm valves without a valve flap. The use of plastic simplified the structural design, fabrication, assembly and made the pump attractive to low-cost applications. Despite having been presented, micropumps have yet to have any design optimization [62].

For the pump's handling and dosing of chemical gases and liquids, polyether–ether–ketone (PEEK) of plastics builds the pump body as PEEK is very stable and highly resistant to various chemical gases and liquids. PEEK is expensive but only a small amount (a few grams) of it suffices for the micropump and so the costs can still be kept low. A lower-cost alternative needs a different material and more research into a suitable candidate [63]. Stainless steel (no. 1.4310 from Ergeste) is chosen for the metal diaphragm. Adhesives for bonding the metal diaphragm to the plastic body have been tested, resulting in the choice of special methanol-stable epoxy glue. The glue's stability to methanol solvents renders the pump suitable for fuel-cell applications. Also, to improve the interface sealing and prevent pump leakage, a layer of polydimethylsiloxane (PDMS) elastomer (PDMS Sylgard, Dow Chemical's) is deposited onto the metal diaphragm by a two-step spin coating process [64].

7. CONCLUSIONS

In vibrating diaphragm micropumps, the cross-sectional area of the pumping chamber determines the flow rate per unit area [65]. The area covered by the valves and devoted to electrical and fluidic connections are excluded as they are minimizable in an integrated system. The ones using electrostatic actuation also have low power requirements [66]. Those operating at system resonant frequency can achieve very high flow-rates per unit power input. In general, the pumping techniques using surface effects such as electroosmosis and flexural plate waves perform better in terms of flow rate per unit area [67].The pumps with the highest flow rate per unit area presented in literatures are nozzle-diffuser micropumps with piezoelectric actuation, injection-type EHD pumps, electro-osmotic micropumps and flexural plate wave pumps [68]. Most pump designs have yet to be rigorously optimized. With design improvements, rotary pumps and other actuation schemes such as SMAs can achieve high flow-rates. Electro-osmotic micropumps can be attractive as external pumps for their possible high back pressures, their simple structure and lack of moving parts. At so small dimensions, the surface forces may start to dominate volume forces, increasing the suitability of pumping mechanisms that are based on surface forces (e.g., electro-osmotic, flexural plate waves and electrowetting pumping). Among the micropumps presented in literatures, the ones having the highest-pressure heads are, in descending order, vibrating diaphragm micropumps with check valves, electro-osmotic pumps and micropumps without valves but with nozzle diffusers. With an input-power constraint, electrowetting micropumps are ideal because their power requirement per unit flow-rate is minimal [69].

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