Micropumps – summarizing the first two decades

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ABSTRACT

Among the large number of microfluidic components realized up to now, micropumps clearly represent the case of a "long runner" in science. A brief literature review reveals, that one of the first scientific papers on a micropump dates from 1978 [4], which is more than two decades ago. An increasing number of publications is found from that time on representing widespread research activities, and there seems to be no change of this trend. An astonishing diversity of micropump concepts and devices has emerged until today, reaching from peristaltic micropumps to a large number of micro diaphragm pumps to recent high-pressure devices without any moving parts. Electrohydrodynamic, electroosmotic, electrostatic, electromagnetic, magnetohydrodynamic, SMA, piezoelectric, thermopneumatic, hydraulic or pneumatic - almost every MEMS-based or mesoscopic actuation principle has been combined with micropumps. An outstanding diversity is also found in the fabrication technology – the span reaches from silicon-based devices over precision machining to injection moulding. This altogether makes it worth to summarize and also take a look into the future of micropumps – after the first two decades.

Keywords: Micro diaphragm pump, valveless micropump, EHD micropump, MHD micropump, electrokinetic micropump, piezoelectric actuator, thermopneumatic actuator, electrostatic actuator, self-priming, bubble-tolerance.

1. MICROPUMP OPERATION PRINCIPLES

Due to the multitude of fluidic designs, actuation principles and fabrication technologies, the classification of micropumps can easily run into a multidimensional approach. As this may generate confusion, a separation according to the type of flow is chosen in this publication. Following this approach, most micropumps found today can roughly be divided into two groups:

“Reciprocating micropumps” are using the oscillatory movement of mechanical parts to transfer mechanical energy into fluid movement. Fluid (i.e. gas or liquid) from these micropumps is delivered in a series of small discrete volumes, which make up a pulsating flow. Classical representatives of this group are “piston type” micropumps like micro diaphragm pumps and peristaltic micropumps. The majority of micropumps found today are following one of these two concepts, which allows a thorough evaluation and comparison of functional principles, basic properties and fabrication technologies. In comparison turbines and gear pumps, although widespread macromechanical pumps, are rarely found as micro components and will therefore not be described in this publication.

“Continuous flow micropumps” are based on a direct transfer of a nonmechanical or mechanical energy form into the movement of a fluid. A number of devices with different forms of primary energy has been developed, reaching from electrohydrodynamic to electroosmotic, magnetohydrodynamic and ultrasonic to electrochemical displacement micropumps. As the design principles of these micropumps are very different depending on the respective principle, this publication will only give a brief literature overview for this class of micropumps, while mainly focusing on the reciprocating concept.

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2. THE BASIC OPERATION PRINCIPLE OF RECIPROCATING MICROPUMPS

The development of micropumps started with the reciprocating or “mechanical” concept and can be traced back to the mid 1970s. These early developments as well as most of the following share the same diaphragm or “piston type” principle, which is depicted in Fig. 1. A pump chamber is realized, which is closed with a flexible diaphragm on one side (or more). By means of a suitable actuation principle, upward and downward movement of the diaphragm is achieved to generate volume changes and, hence, under- and overpressure transients $\Delta p$ in the pump chamber. The working principle can be described by a cyclic process, which is divided into a supply mode (the pump chamber volume increases) and a pump mode (the pump chamber volume decreases).

During the supply mode underpressure is generated in the pump chamber, which causes fluid to be sucked into the pump chamber through the inlet, as soon as $\Delta p$ becomes higher than the inlet valves threshold pressure $\Delta p_{crit}$. During the pump mode overpressure occurs in the pump chamber, which transfers liquid from the pump chamber into the outlet, as soon as $\Delta p$ becomes higher than the outlet valves threshold pressure $\Delta p_{crit}$. In this stage the inlet valve is blocking an unwanted reverse flow, as the outlet valve does during the supply phase.

The well-known switched capacitor voltage converter forms a close electrical equivalent to this pumping principle (see Fig. 1): here the capacitor C takes the task of the pump chamber. The electrical voltage across the capacitor is equivalent to the pump chamber pressure, whereas the electrical current takes the role of the fluid flow. Actuation is performed with an AC voltage source, which modulates the footpoint potential of C. It is easily concluded from both schematics, that backward flow from the outlet into the inlet is prohibited by the fluid directioning properties of the valves, whereas forward flow can be enforced by a rise of the inlet pressure over the outlet pressure.

It should be mentioned here, that the micro diaphragm pump concept has proven to be successful long before any of the technical realizations described here were initiated. Some predaceous bugs, like “Graphosoma lineatum”, [1,2] use piston micropumps for the external digestion of their insect prey (Fig. 2). One micropump is used to dispense saliva loaded with digestion enzymes through the bugs picking stylet into the victim’s body. A second micropump sucks the externally digested food back.
3. MECHANICAL MICROPUMPS – THE BEGINNING

The “biological design” of the “Graphosoma lineatum” micropumps uses a muscle-operated piston and passive flap valves as inlet and outlet valve. The first technical micropump designs used a slightly different approach, as they were based on an actuation of both the pump diaphragm and the valves. The device of Spencer et al. may also be taken as one of the early technical examples of these micropumps, which date back to the mid seventies [3][4]. His approach was a piezoelectrically actuated micro diaphragm pump with active flap valves designed for insulin dosing as a potential application (see Fig. 3). The cylindrical micropump body was machined from stainless steel. A 0.1 mm thick stainless steel shim was used as a pump diaphragm with the piezoactuator glued on top. The active flap valves were constructed in a normally-closed configuration using piezoelectric bimorphs with dimensions of 0.4 mm x 4 mm x 20 mm. The valve bimorphs were fully immersed into the dispensed liquid and, thus, had to be coated to provide electrical insulation. With an actuation voltage of 100 V at the pump diaphragm, a theoretical stroke volume of 1.94 µl and a maximum output pressure of 100 mm Hg were calculated. Measurement data confirm a maximum stroke volume of 1.5 µl with 90 V actuation and a maximum counter pressure of appr. 60 mm Hg at 70 V actuation voltage [4]. Although used only in unidirectional mode here, this type of micropump would also allow a bidirectional fluid transport by simply changing the valve actuation scheme.

Pump designs with fully immersed actuators, as described above, clearly present a weak point, as they might easily cause personal hazard due to the electrical potentials applied and also are prone to electrical failure. Smits et al. addressed this problem later on and realized a peristaltic micropump with active valves. They published their results in 1990 [5]. The micropump is made from an anisotropically etched silicon wafer, that bears valve seats on its topside and connecting microchannels at the bottom. The wafer is sealed on both sides with glass by anodic bonding. The top glass wafer was left unconnected to the silicon at the valve seats. By glueing three piezo disks at these places three pump diaphragms with valve function are realized. Peristaltic pumping is achieved by sequential activation of the piezo actuators.

For this device a maximum pump rate of 100 µl/min with an operation frequency of 15 Hz and a maximum outlet pressure of 60 mBar at zero flow are reported. Ref. [5] gives no geometrical data at all. A calculation of the realized flow rate indicates, however, that the pump diaphragms must exhibit a similar size as described in [4]. It is clearly visible that the problem of liquid wetting of the actuators is prevented here by a clear separation of both. Moreover, bidirectional pumping is easily achieved by changing the actuation scheme of the valves.

Fig. 3: Schematic cross section of a micro diaphragm pump with active valves [4]

Fig. 4: Peristaltic micropump [5]: schematic representation of the valve actuation scheme for pumping from inlet to outlet (top to bottom)
The first design of a micropump with passive check valves was published in 1988 by Van Lintel et al. at the University of Twente [6]. It uses again a three-layer setup with two glass sheets enclosing an anisotropically etched silicon wafer. Typical dimensions were 12.5 mm for the pump diaphragm diameter and 7 mm for the diameter of the membrane valves. With an actuation voltage of 100 V a stroke volume of 0.21 µl, a maximum flow rate of 8 µl/min at 1 Hz operation frequency and a maximum counter pressure of 100 mBar were observed.

The work of Smits and Van Lintel marks the beginning of a multitude of micro diaphragm pump developments that lasts until today. Most of the devices realized are based on the diaphragm type or the peristaltic type described above, but with varying valve geometries, actuation mechanisms and fabrication technologies. A number of trends can be observed concerning the various functional elements and basic properties, which shall be briefly summarized in the following:

4. ACTUATION PRINCIPLES

The actuators in use up to now cover almost the whole range of microactuation techniques available. The common principles are thermopneumatic [7,8,9,10], piezoelectric [11,12,13,14,15,17,40,42], electromagnetic [16,17] and electrostatic actuation [18,19,20], others, like shape memory alloys [21] or magnetostrictive materials are rarely found.

Piezoelectric actuation as shown in Fig. 4 and Fig. 5 in principle provides the advantages of a comparatively high stroke volume, a high actuation force and a fast mechanical response, and is, therefore, a very attractive and frequently used actuation principle for micropumps. Disadvantages are the comparatively high actuation voltage and the mounting procedure of the PZT disk, which requires a very well defined gluing process. It has been shown in recent research, that process optimisation of the mounting process can significantly improve reliability and yield for this type of actuator [22,23]. As an alternative, screen printing [24,25] and also thin-film deposition of PZT material have been studied. Although the feasibility of these techniques could be demonstrated, the resulting strokes (typ. 1 µm at 100 V in [24]) are small in comparison to glued PZT bulk material (e.g. 15 µm at 100 V in [11]). Optimisation of the geometrical design was done at several places to achieve higher strokes at lower voltages [11,26]. Typical actuation voltages of such optimised design are in the range of 100 V (e.g. 130 VpP for the micropump in [12]), which is a significant improvement in comparison to other micropumps that sometimes use commercial piezo buzzers without any optimisation (e.g. 400 VpP for the micropump in [13]). This lower actuation voltage is also helpful for the design of highly miniaturized electronic drivers, which allow low-power operation from a battery [12].

Thermopneumatic actuation was also investigated early as a low voltage alternative to piezoelectric actuation [7,8,9,10]. Fig. 6 shows the first design by Van de Pol et al. that follows Van Lintels set-up [7]. Like this most designs use an air-filled chamber on top of the pump diaphragm, that embodies a heater resistor, either free-hanging [7,8,9] or attached to the pump diaphragm [10]. Upon heating the air enclosed inside the chamber expands and causes a downward deflection of the pump diaphragm, which disappears after active or passive cooling. This type of actuator can be operated below 10 V without big efforts concerning the electronic driver [7]. Moreover, thermopneumatic actuators can be made very compact [10] and can also be tailored for strokes up to several 100 µm to achieve high pump rates [8].
Integration into standard silicon processing is easily achieved without the need of special mounting procedures [9]. A crucial drawback of this actuation principle is on one hand the relatively long thermal time constant of the air-filled actuation chamber, especially during the cooling process that limits the actuation frequency to appr. 50 Hz. On the other hand, the typical electrical power consumption is in the range of several Watts, which usually excludes portable operation from a battery. At these power levels a heating of the transported medium cannot be excluded.

Electrostatic actuators offer an extremely fast mechanical response, that allows operation frequencies up to several kHz, a low power consumption and full MEMS compatibility for an easy integration, and have therefore been chosen for several micropump designs [18,19,20]. Fig. 7 shows the first practically successful device of Zengerle et al., who also realized one of the first vertically-stacked chip designs in silicon [19]. The actuator is made from two silicon chips that embody the flexible pump diaphragm and a rigid counter electrode in a capacitor-like configuration. Applying a sufficiently high voltage to the capacitor electrodes causes electrostatic attraction of the pump diaphragm, which in extreme gets fully attached to the counter electrode. After discharge of the capacitor the pump diaphragm returns to its rest position by mechanical relaxation. Due to the high operation speed of the actuator, bidirectional pumping was observed with this micropump at high operation frequencies caused by a time delay occurring between valve switching and diaphragm operation [19]. However, not only for this design, but in general, a major disadvantage results from the inherently small actuator stroke, which is usually limited to values around 5 µm with actuation voltages around 200 V. A higher stroke while maintaining the diaphragm stiffness would require a significantly higher actuation voltage with the danger of dielectric breakdown inside the capacitor. On the other hand lower diaphragm stiffness, hence, a higher stroke at the same actuation voltage, would reduce the operation pressure, which is directly correlated with the mechanical relaxation forces. Also degradation of the actuator performance was frequently found in long-term operation. This can be attributed to the build-up of surface charges at the isolator inside the capacitor, which reduce the internal electrical field strength and, therefore, the stroke. Bipolar operation is a practical solution to overcome this problem at the prize of a more complex electronic driver.

Electromagnetic actuation is found sometimes [16,17]. Although not well compatible with MEMS integration, this actuation concept can easily be adapted in a modular way and offers the benefit of a separate optimisation of micropump and actuation unit. The two references cited here use a permanent magnet attached to the pump diaphragm that is moved by an external coil. The overall electrical and mechanical properties are comparable to thermopneumatic actuators with the advantage of a slightly faster mechanical response.

5. VALVE DESIGN

The “classic” design of a micro diaphragm pump shown in Figs. 5 to 7 implies the use of mechanical check valves, which can be realized as membranes or flaps (Fig. 8). The effort to design and fabricate such valves is considerably high, as a number of properties, like backward flow, pressure drop and switching speed have to be kept under tight control to achieve a working micropump. Moreover, wear and fatigue is a critical issue, especially in polymer-fabricated devices. There is also the risk of valve blocking by even small particles, which instantly would degrade the pumping performance and therefore limits the application range of most valve-based micropumps to filtered media.
The so-called “valve-less” micropump concept was tailored to avoid these problems. The device was first introduced by Erik and Göran Stemme in 1993 [27]. It uses diffuser/nozzle elements with direction-dependent flow characteristics to mimic, although not perfect, the function of a check valve. A maximum ratio of 2.23 between forward and backward flow can be calculated for this type of “valve” [27], which is sufficient for a pumping effect.

The prototype shown in Fig. 9 was fabricated in a cylindrical brass body with an outer diameter of 29 mm and tested with two different diffuser/nozzle geometries. The theoretically calculated forward-to-backward flow ratios of 1.48 and 1.67 agreed well with the experimental data. A remarkably high zero-pressure flow of 11 ml/min and a maximum outlet pressure of app. 100 mBar were found for water, depending on the diffuser/nozzle geometry in use. This micropump was also a gas pump.

Based on this prototype various silicon versions were realized in a fully planar design. Fig. 10 shows a realized chip with two micropumps that work in an antiparallel configuration to reduce inlet and outlet pressure pulses and to increase the pump flow performance [28,29]. A similar planar design was developed by Foster et al. [30] on the basis of a flow-rectifying structure originally proposed by Nicola Tesla [31].

Vertically stacked designs were originating from the work of Gerlach et al. [32] and, later on, Koch et al. [25]. They use the conical sidewalls of anisotropically etched silicon cavities to build a diffuser/nozzle element with no additional effort. A theoretical treatment of the pump principle in conjunction with the corresponding diffuser/nozzle elements is given in a number of publications [30,32,33,34].

The common advantage of valveless micropumps is a relatively simple construction in comparison to pump concepts with check valves or even active valves. Moreover, the pumping of particle-loaded media or sensitive material is easier to achieve due to the open flow structures. These benefits are, however, accompanied by the disadvantage that no selfblocking effect exists. As a result any overpressure build-up at the outlet will cause a more or less significant reverse flow that becomes predominant as soon as the pump is switched off. A different approach to a “valve-less micropump” with improved blocking capability was found by Stehr et al. [35,36], who discovered and evaluated the pumping effect of a bossed silicon diaphragm valve that was periodically actuated by a mesoscopic piezoelectric bimorph. It turned out that the dynamic modulation of the gap between boss and valve seat exhibits a flow-rectifying behaviour usable for fluid transport. Another version used an additional elastic buffer built into the pump chamber to generate a time delay between diaphragm movement and pressure build-up and, hence, a net flow. The device was able to transport liquids and gases. In both cases the pumping direction could be reversed by a variation of the actuation frequency. Due to the inherent valve function of the device, reverse flow in the off state could be prevented to a certain extent. During operation however, reverse flow is present like in the other valveless designs.
6. BASIC DOSING PROPERTIES

Among the large number of micropump designs and concepts the basic dosing properties turn out to be quite similar. This may be demonstrated with the device of Ref. [11]. Fig. 11 shows the pump rate as a function of the actuation frequency at zero differential pressure over the pump and the typical influence of a varying outlet overpressure at various pre-set operation frequencies.

![Fig. 11: Typical dosing behaviour of a micro diaphragm pump: pump rate as function of the operation frequency (left) and as function of the applied outlet backpressure (right), both with water as medium [11]](image)

The pump rate is linearly increasing with the operation frequency up to a certain corner value (here appr. 150 Hz). From this linear behaviour the stroke volume $\Delta V$ of the diaphragm actuator can be calculated, which is appr. 122 nl in this case. In the linear range the actuator is capable to deliver the full stroke and the duration of a pumping cycle is still long enough to allow a full relaxation of the valve movement and all pressure and flow transients. Above the corner frequency saturation is observed due to an increasingly insufficient relaxation and other, secondary effects, which depend on the individual micropump design. The falling branch of the pump rate curve indicates an increased energy loss (e.g. by squeeze film damping) and an increasing time lag between diaphragm and valve movement, which eventually can lead to reverse pumping [19].

The backpressure curve shows a linear degradation of the pre-set flow rate as soon as the outlet overpressure (or “counter pressure”) increases. At a certain corner pressure, the pressure peaks generated inside the pump chamber are no longer high enough to open the outlet valve, which means, that the micropump stops operation. For this particular design the maximum operation pressure at the outlet is 800…900 mBar. A similar characteristic, however with a different corner pressure, is found for underpressure at the inlet.

Especially the influence of a varying backpressure can pose a limit to the dosing accuracy of a micropump. One way to overcome this problem is the application of high force actuation principles that extend the maximum backpressure values well beyond the operational values expected in a certain application. In these cases the pressure-dependent flow curve may become sufficiently flat to ensure an accurate dosing at varying outlet overpressure.

Refs. [40,41] describe a mechanical double limiter concept to ensure a precise metric dosing in a limited frequency range. By this concept the diaphragm movement is set to a predefined value that will allow only a certain amount of liquid to be transported during each pump cycle. Successful metric dosing could be demonstrated in a flow range of 0…2 ml/hr and for outlet overpressures and inlet underpressures of 200 mBar and -100 mBar, respectively.
7. SELF-PRIMING AND BUBBLE-TOLERANCE

As there exist sometimes somewhat different definitions of “self-priming” and “bubble-tolerance”, a clarification from the perspective of reliability might be helpful first:

**Self-priming** describes the ability of a micropump to fill itself completely with liquid without any additional measures. This should be possible from the (totally) dry state as well as the “wet state”, i.e. after a previous emptying of the pump chamber and the connecting tubes. In the wet state residual liquid is usually trapped inside the system, which can hinder re-priming.

**Bubble tolerance** describes the ability of a micropump to transport a gas bubble completely through its inner structure without any residual gas left inside. The size of the gas bubble should not take any influence.

During the first phase of micropump development self-priming and bubble-tolerance were soon identified as a key criterion for a reliable operation. None of the early micropump designs described up to now was self-priming in the sense defined above. The manual priming procedures necessary turned out to be complicated and unreliable, as small gas bubbles remained often trapped inside the devices. A practical approach to solve this problem was made with a CO$_2$-purge to remove all air before the priming was started [37]. Residual CO$_2$ inside the pump was easily dissolved in the following (aqueous) priming solution, which resulted in a complete filling. The problem of bubbles travelling towards the micropump in the inlet tubing, however, remained unsolved. After a thorough and time-consuming priming most early micropumps simply stopped working as soon as a gas bubble arrived at the inlet valve [39].

The problems of self-priming and bubble-tolerance were discussed quite early [6,7,38], sometimes even with the pessimistic argument, that a micro diaphragm pump with passive valves would not be able to be self-priming at all due to physical reasons [35,36]. Nevertheless, the first successful approach was made in 1996 by Döpper et al. with a “self-filling” polymer-fabricated micropump [13]. The first - and up to now the smallest - working prototype of a self-priming and bubble-tolerant silicon micropump was realized in parallel by Linnemann, Woias et al. in 1996, who finally published in 1998 [11,12]. It took also until 1998 for the first comprehensive theoretical publication on the subject by Richter et al., who deduced by theory and experiment the physical requirements and boundary conditions, that have to be fulfilled to make a micro diaphragm pump self-priming and bubble-tolerant [39].

Using basic laws of thermodynamics and fluid compression, minimum values for the compression ratio $\varepsilon$, i.e. the ratio of the stroke volume $\Delta V$ and the pump chamber volume $V_0$ were calculated in [39] and proven with experimental data. Typical $\varepsilon$ values for a liquid pump were found to be in the order of $10^{-6}$, whereas a gas pump already requires a significantly higher compression ratio in the order of $10^{-2}$. An even higher compression ratio is required for a gas pump with a wetted inner structure, which meets the case of bubble-tolerance.

From these results the following design rules can be deduced for a truly self-priming and bubble-tolerant micropump:

- The threshold pressure $\Delta p_{crit}$ of the valves should be made as small as possible to keep the required pressure transients $\Delta p$ small,
- the stroke volume $\Delta V$ of the pump diaphragm should be made as large as possible by choosing an appropriate actuation mechanism, and
- the inner volume $V_0$ of the pump chamber should be made as small as possible: Both measures help to increase the compression ratio $\varepsilon = \Delta V / V_0$ as much as possible.

![Fig. 12: Self-priming and bubble-tolerant silicon micro diaphragm pump [10,11]: schematic diagram (top) and photograph (bottom, courtesy of FhG-IMS, Munich, Germany)]
These rules were followed in the design shown in Fig. 12, which is used as an example here. By choosing and optimising piezoelectric actuation a diaphragm stroke of 15 µm and a stroke volume of 85 nl were achieved with a diaphragm size of only 5 mm x 5 mm [11]. The internal volume of the pump chamber was reduced to 800 nl by means of a combined chemical and mechanical grinding of the middle wafer. The first design had a compression ratio of 1:9.4, which proved to be sufficiently high for self-priming and bubble tolerance [39]. Later versions were gradually further optimised [12], finally up to a compression ratio of 1:2, which is extremely high in relation to the small chip size of only 7 mm x 7 mm x 1 mm.

Other successful self-priming and bubble-tolerant designs, that follow the same rules, are described in Refs. [8,13,14,17,41,42]. Some approaches, e.g. the device described in [35,36] turned out to be only self-priming from the dry state, but not fully bubble-tolerant, which results in a limited practical applicability.

8. MATERIALS AND FABRICATION TECHNOLOGIES

After the early designs, which were realized by conventional machining [1,2], micropump fabrication became an exclusive domain for silicon micromachining at the beginning of the 1990s. Until today, the main advantage of this technology is definitely the high precision and reproducibility achievable with the materials in use, mainly silicon and glass. Moreover, the results of long-term tests allow the conclusion, that wear and fatigue of mechanically moving parts do not seem to pose any problem with this technology [12,22]. Nevertheless, the disadvantages of a rather high fabrication cost and a limited material choice have enforced a search for alternatives quite soon. Polymer microfabrication, namely microinjection moulding [10,13,42], polymer hot embossing [43] and stereolithography [15,16] were demonstrated as suitable technologies for micropump fabrication since around 1995. The material basis has definitely been broadened by this approach. However, the goal of a true “low-cost” micropump, although often promised, is not satisfied up to now with these technologies, which, although being more cost effective, are still highly complex and therefore comparatively expensive microfabrication processes. Moreover, other material-related aspects, like limited lifetime can be a critical issue. An interesting pragmatic alternative can be seen in recent “low tech” micropumps made by conventional moulding [14,17] or well-established printed circuit board technology [8]. This concept may not provide a micropump with the ultimate performance of a silicon micromachined device, but still an acceptable result at moderate fabrication costs. It therefore seems to be the future choice for all low performance and low cost applications, provided, that other requirements, like reproducibility and operational stability, can be satisfied. True high performance applications, e.g. in drug delivery, can still be regarded as a clear domain of silicon micromachining as demonstrated by recent industrial efforts in this direction [40,41]. Moreover, one should not forget, that the cost of a microcomponent is not exclusively defined by the fabrication technology itself, but mainly by an “intelligent” and fabrication-oriented design. It can be clearly seen in the large number of publications cited here, that this aspect has often been left out of consideration in favour of a scientific result, which is fully justified from the standpoint of science. However, a more fabrication- and cost-oriented design philosophy is a premier need for the near future to seed the basis for a broader commercial application.

9. CONTINUOUS FLOW MICROPUMPS – AN OVERVIEW

Continuous flow micropumps are based on a direct energy transfer, that results in steady flow through the respective devices. Various working principles, like electrohydrodynamic, electrokinetic, ultrasonic and RF, magnetohydrodynamic and electrochemical displacement micropumps are found in literature. A common property of continuous flow micropumps is the simplicity of the microstructures involved, which is frequently limited to microchannel structures or electrode arrangements. The performance is, however, in many cases strongly influenced by a number of fluid properties, which frequently limits a certain principle to a small class of fluids. Gas transport is, in most cases, excluded at all.

First proposals on EHD pumps originated around 1960. Thirty years later this idea was picked up by Bart et al. and Richter et al. for their electrohydrodynamic (EHD) micropumps [44, 45,46]. The basic principles used in their designs were the EHD induction effect [44], i.e. the generation and movement of induced charges at a fluid-fluid or fluid-solid boundary layer, and the EHD injection effect, which is based on the electrochemical formation and movement of charged ions [45,46]. The practical realization of an EHD micropump following the latter principle is shown in Fig. 13 as an example. For the micropump silicon chips with identical silicon grid electrodes and with gold layers on the opposing surfaces were realized by anisotropic etching and sputter coating, anodically bonded together in pairs and placed into a flow channel. The grid distance was defined by the wafer thickness. A high voltage was applied between
the electrodes to inject ions into the liquid. Transport of the ions in the electric field between the opposed electrodes was inducing a net fluid transport through the grid electrodes.

With voltages up to 800 V a remarkable net flow of 14 ml/min was achieved with ethanol. A reverse poling of the electrodes could generate bidirectional pumping. The maximum outlet pressure was 24 mBar at 700 V actuation [45]. In a later publication a similar structure was proposed for flow measurement on the basis of charge injection [46]. The simplicity of this concept is - still - charming. However, all EHD pumping principles rely critically on the electric properties, namely permittivity and conductivity, of the fluid to be transported. Typically the conductivity must remain in between $10^{-14}$ S and $10^{-9}$ S, which severely limits the application to non-conducting and non-ionic fluids (e.g. solvents). Consequently, only a few studies have followed the work described above, e.g. the transport of non-conducting solvents in microarray-based drug discovery [47].

Electrokinetic micropumping is the major basis of capillary microelectrochromatography, which has seen a tremendous progress in science during the last years, mainly pushed by the “lab on a chip”-concept. Electrokinetic pumping is also one of the first successful candidates for a broad commercial application of a micropump principle. (Fig. 14). A comprehensive treatment of this subject would by far exceed the scope of this publication and can be found in a more direct way in related textbooks or papers [48,49,50]. Among these a recent and interesting publication was contributed by Paul et al., who studied the electrokinetic transport of liquid in microporous media [51]. By simply using a fused silica capillary filled with porous beads and with a platinum electrode at either end, they could demonstrate flow rates in the range of 0.1 µl/min at pressures up to 200 atmospheres, which sets an absolute record in high-pressure micropumping.

RF and ultrasonic pumping exploit the dragging force of a progressive mechanical wave [52,53] or the “quartz wind” of a vertically oscillating surface [54], which are excited at a liquid-solid interface, e.g. the sidewall or end wall of a microchannel. Lamb-wave devices [52] as well as piezoelectric ultrasonic or RF transducers [53,54] have been applied for power generation with frequencies reaching from the ultrasonic range up to several 10 MHz. The common property of these principles is a very small pump rate in the range of only several µl/min and an extremely small outlet pressure (e.g. 0.13 Pa in [54]). This is also confirmed by recent simulations [55]. The premier application can therefore be seen in micromixing or fluid positioning (e.g. the movement of bacteria-loaded liquid in microscopy) instead of a genuine fluid transport.

Magnetohydrodynamic (MHD) pumping is based on the Lorentz effect and, hence, a very old pumping principle. It is based on the injection of a transverse electric current into an electrolyte solution in a microchannel, which is simultaneously under the influence of a transverse magnetic field, oriented in an angle of 90° to the current direction. The result is a longitudinal fluid flow along the microchannel due to the Lorentz force acting onto the ionic current in the solution. Like with EHD pumps bidirectional pumping is easily achieved by a reversal of the electric current or the magnetic field vector. Flow rates and outlet pressures depend strongly on the fluid geometry in use, however, microtechnical solutions typically show only small values for both, e.g. 63 µl/min and 1.8 mBar for the device described in [56]. Moreover, electrolytic bubble generation at the current-injecting electrodes poses a severe problem for this micropump principle. This particular problem has been addressed with an AC electrohydrodynamic pump, which uses an electromagnet instead of a permanent magnet [57]. By driving both the magnet coil and the current injector from a synchronous AC source, the Lorentz vector and, hence, the fluid flow direction stay always in the same direction. Choosing the operation frequency high enough prevents the electrolytic formation of gas bubbles.
Electrochemical displacement micropumps, on the contrary, use the electrolytic generation of gas bubbles as a primary pumping effect. The concept has found its first applications in drug delivery [58] and is well illustrated in the publications of Böhm et al. [59,60]. He uses a reservoir filled with an electrolyte and with two immersed electrodes to generate gas bubbles by current injection. The corresponding volume increase generates a continuous or step-by-step displacement in an adjacent meander, which carries the fluid to be dispensed.

Fig. 15: Electrochemical displacement micro-pump: function principle (left) and photograph of a prototype carrying two identical devices (right) [58]

The function principle and a monolithic integration of this concept are shown in Fig. 15. By a closed-loop control of the gas generation process [60] this dosing system is capable to deliver discrete fluid quantities as small as 100 nl with an accuracy of at least 5 nl.

10. CONCLUSIONS AND OUTLOOK

What is it that has made micropumps so fascinating during the last two decades? In the first phase of focused research starting around 1990 the pioneering aspect definitely has been the main driver. Research at that time shows, like in every pioneering phase, the character of a scientific playground area, where a whole spectrum of micropump concepts, actuation principles, valve designs and fabrication technologies were out to be explored. The primary goal, of course, was to realize a novel microfluidic device and to improve its performance. In addition, however, the mixture of microactuation, microfluidics, microfabrication and application-related aspects present in a micropump has generated an ideal “condensation point” for MEMS research. Therefore, aside from its genuine task, micropump research has also promoted the development of microactuators, various microfluidic devices and has served as a “workbench” for microfluidics research in general, and is doing so up to now.

During the first, explorative phase of research the advantages and drawbacks of the various micropump concepts have eventually become clear. A second phase can be seen starting around 1995 with the study of alternative fabrication technologies and genuine micropump-related problems. From this time on the performance of micropumps has been constantly increased and features such as self-priming, bubble-tolerance, high flow and pressure or precision dosing have been addressed more precisely and solved with various improvements in the design principles. As a result, today a whole spectrum of micropumps is available in various development stages, sometimes even on a more or less commercial basis [23].

The overall commercialisation process, however, is still in the beginning: first products have entered the marketplace not long ago (see Fig. 14) and others seem to be close to that point [40,41]. This may seem insufficient after more than a decade of concentrated research, however, one should keep in mind, that other MEMS components have needed similar time from the first scientific publication [61] to a commercial product [62]. Keeping commercialisation in mind, a cost- and fabrication-optimised design and a good performance reproducibility are definitely research topics for the next future. The work on appropriate design and modelling tools is another urgent need. And yet, although there has been a tremendous progress, there is still enough to learn about the fundamental - in part very special - processes occurring in micropumps and other microfluidic components and systems, like influences of media properties, long-term effects or liquid-surface interactions. This altogether will give enough work to do - for the next decades.
REFERENCES


