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Sensors and Actuators B 105 (2005) 28–38
SENSORS AND ACTUATORS B CHEMICAL
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Micropumps—past, progress and future prospects

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Available online 6 May 2004

Abstract

If a high and on-going research dynamics is taken as a rejuvenating factor microfluidics can still be regarded as a truly young discipline, although some microfluidic devices definitely can not be considered as “youngsters” any more. In a time of 30 years of ISFETOLOGY it may be worth to take a look at these devices in order to examine, whether they had—and still have—the potential to stimulate the imagination and creativity of researchers in a similar way as the Ion Sensitive Fieldeffect Transistor did since its invention in 1970. The area of micropumps is definitely one of those “long runners”. Starting in the mid 1970s a steadily growing and astonishing diversity of micropump principles, technical concepts and applications has emerged in this area. Until today MEMS science is delivering a constant flow of novel modelling approaches, microstructured materials, actuation principles, fabrication technologies and applications, that are readily taken and transferred into micropump research. Among the potential applications especially the combination of biochemical sensing and microfluidics has provided a substantial stimulus for micropump research and development in the past and will do so in the future.

Keywords: Micro diaphragm pump; Valveless micropump; EHD micropump; MHD micropump; Electrokinetic micropump; Piezoelectric actuator; Thermopneumatic actuator; Electrostatic actuator; Self-priming; Bubble-tolerance

1. Introduction

Most micropumps found today can roughly be divided into two groups:

The so-called “reciprocating micropumps” use the oscillatory or rotational movement of mechanical parts to displace fluid. Micropump development has started with “piston type” reciprocating micropumps like micro diaphragm pumps and peristaltic micropumps that do still form the main representatives of this class in the MEMS world. The common principle employs a pump chamber which is closed with a flexible diaphragm on (at least) one side (Fig. 1a). Oscillatory movement of the diaphragm generates a two-phase pump cycle with periodic volume changes and, hence, under- and overpressure transients Δp in the pump chamber. During the so-called “supply phase” the underpressure in the pump chamber will suck fluid through the inlet into the pump chamber. During the following “pump phase” overpressure in the pump chamber transfers liquid into the outlet. The valves at the inlet and outlet will block unwanted reverse flow in the respective pump phases. They therefore act as “fluidic rectifiers” that direct the bidirectional fluid movement generated inside the pump chamber into a desired unidirectional flow. The medium (i.e. liquid or gas) is obviously delivered in a series of discrete fluid volumina whose magnitude depends from the actuator stroke volume ΔV, i.e. its net volumetric displacement during one cycle.

It should be mentioned, that biological evolution has perfected this “technical” concept—even in micro dimensions—long before any technical realizations were even considerable. Some predaceous bugs, like Graphosoma lineatum, [1,2] use piston micropumps at the end of their picking stylet for the external digestion of their insect prey (Fig. 1b). These micropumps exhibit a muscle-operated piston and passive flap valves as inlet and outlet check valve. One micropump is used to dispense saliva loaded with digestion enzymes through the picking stylet into the victim’s body. A second micropump sucks the externally digested food back.

“Continuous flow micropumps” are based on a direct transformation of nonmechanical or mechanical energy into a continuous fluid movement. Devices with ultrasonic, magnetohydrodynamic (MHD), electrohydrodynamic (EHD), electroosmotic or electrochemical actuation mechanisms have been developed up to now. As the design principles are quite different depending on the respective physical or chemical principle this publication will only give a brief overview for this class of micropumps.

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doi:10.1016/j.snb.2004.02.033

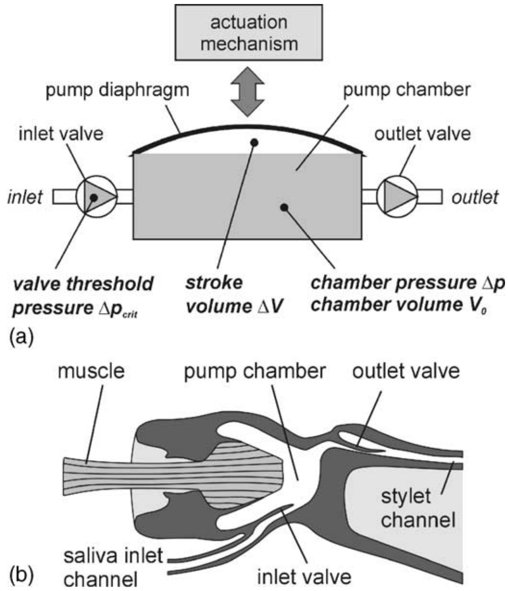

Fig. 1. (a) Principle set-up of a "piston type" micropump. (b) Schematic set-up of the saliva dispensing micropump of "Graphosoma lineatum".

2. Early research on reciprocating micropumps

Instead of using passive valves the first technical micropump designs were based on an actuation of both, the pump diaphragm and the valves. Spencer et al. [4] have presented an early example in 1978. They mention an even earlier publication from Thomas and Bessman [3] which dates from 1975. Spencer's approach is depicted in Fig. 2. A cylindrical micropump body was machined from stainless steel and covered with a piezoelectrically actuated stainless steel shim as a pump diaphragm.

The active flap valves were made from piezoelectric bimorphs with dimensions of 0.4 mm × 4 mm × 20 mm. These bimorphs were fully immersed in liquid and, thus, had to be coated to avoid electrical short circuits. With a pumping actuation voltage of 100 V a theoretical stroke volume

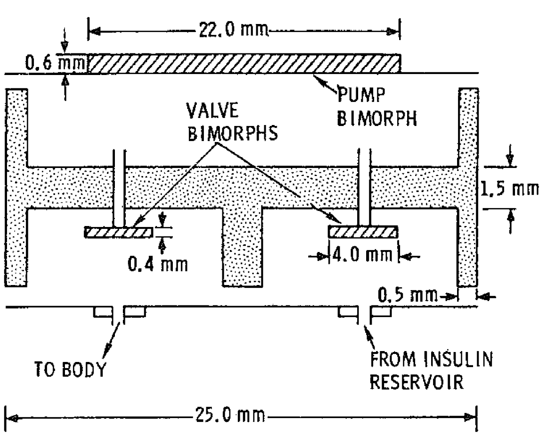

Fig. 2. Schematic cross-section of a micro diaphragm pump with active valves [4].

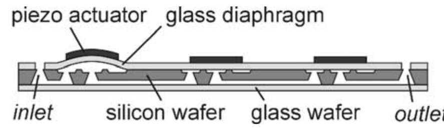

Fig. 3. Planar peristaltic micropump [5].

of 1.94 μl and a maximum output pressure of 100 mmHg were calculated. Measurement data confirm a maximum stroke volume of 1.5 μl with 90 V actuation and a maximum counter pressure of approximately 60 mmHg at 70 V actuation voltage [4]. Although used only in unidirectional mode here, this type of micropump would also allow bidirectional fluid transport by simply changing the valve actuation scheme.

A similar silicon-based device was first realized by Smits [5] with a planar peristaltic micropump. They published their results in 1990. The micropump is made from an anisotropically etched silicon wafer that carries valve seats on its topside and connecting microchannels at the bottom (see Fig. 3). The wafer is sealed with glass by anodic bonding from both sides. The top glass wafer was left unconnected to the three silicon valve seats and converted into three pump/valve diaphgrams by glueing piezo disks at these places. The problem of liquid wetting of the valve actuators is prevented here by the planar construction which separates the piezoactuators from the liquid flowpath.

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 for this device. 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].

The first micro diaphragm pump with passive check valves has also resulted from research at the University of Twente and was presented in 1988 by Van Lintel et al. [6]. It uses again a three-layer set-up with two glass sheets enclosing an anisotropically etched silicon wafer (Fig. 4). Typical dimensions were 12.5 mm for the pump diaphragm diameter and 7 mm for the diameter of the membrane valves. A stroke volume of 0.21 μl was achieved with an actuation voltage of 100 V. This corresponds to a maximum flow rate of 8 μl/min at 1 Hz operation frequency and a maximum counter pressure of 100 mbar.

While the early work of Spencer and Thomas is widely unknown today, the publications of Smits and Van Lintel mark

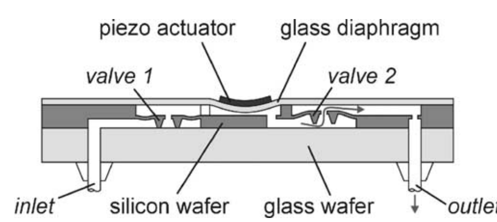

Fig. 4. Micro diaphragm pump with piezoelectric actuation [6].

the beginning of extensive micropump research in MEMS. An astonishing high number of diaphragm and peristaltic micropumps was developed in the following with varying valve types and geometries, actuation mechanisms and fabrication technologies. Several discriminating features and trends can be observed concerning the functional elements and basic properties of these devices, which shall be briefly summarized in the following.

3. Actuation principles

Up to now almost the whole range of microactuation techniques available have been used for the design of micropumps. Common principles include piezoelectric [11–15,17,40,42], thermopneumatic [7–10], electrostatic [18–20] and electromagnetic actuation [16,17], whereas some others, like shape memory [21] or magnetostrictive effects are rarely found.

As shown before, piezoelectric actuation was the first actuation principle used in micropumps. It is a very attractive concept, as it provides a comparatively high stroke volume, a high actuation force and a fast mechanical response. Moreover, commercial PZT material is readily available for a hybrid integration. The comparatively high actuation voltage and the mounting procedure of the PZT disk can be regarded as disadvantages. A systematic optimisation of the mounting process can significantly improve reliability and yield of this type of actuator [22,23]. Nevertheless, hybrid integration requires a very well-defined glueing which is critical for the actuator performance and not easily done. Therefore, screen printing [24,25] and thin-film deposition of PZT material have been studied as an alternative quasi-monolithic integration technique. Although the feasibility of these techniques could be demonstrated, the resulting strokes (e.g. 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 V_{PP} 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 V_{PP} 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].

As a second concept thermopneumatic actuation was first demonstrated by Van de Pol et al. [7] with a micropump similar to Van Lintels device. Instead of a PZT disk an air-filled chamber with an internal heater resistor was integrated on top of the pump diaphragm. The heater is realized either as a free-hanging structure [7–9] to achieve a better thermal efficiency or simply attached to the pump diaphragm [10]. This type of actuator represents a low-voltage alternative to the piezoelectric drive and does not require big efforts concerning the electronic driver [7]. Moreover, thermopneumatic actuators can be made very compact [10], but are nevertheless capable to generate strokes up to several 100 μm to achieve high pump rates [8]. Integration into standard silicon processing is easily achieved [9]. A crucial drawback of this actuation principle is a relatively long thermal time constant, especially during the cooling process. This limits the upper actuation frequency to approximately 50 Hz. The typical electrical power consumption is in the range of several watts which usually excludes portable operation from a battery. Also, at these power levels a heating of the transported medium can not be excluded.

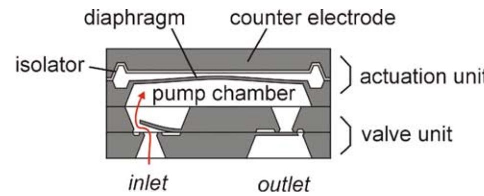

Fig. 5. Electrostatically actuated micro diaphragm pump [19].

Fig. 5 shows the first practically successful micropump with electrostatic actuation from Zengerle et al. [19], who also realized the first vertically-stacked modular micropump design in silicon. 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 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 relaxates to its rest position. With this rapid actuation principle bidirectional pumping was observable at high operation frequencies caused by a time delay between the diaphragm movement and the somewhat slower valve switching [19].

Electrostatic actuation offers operation frequencies up to several kHz, an extremely low power consumption and full MEMS compatibility [18–20]. A major disadvantage results from the inherently small actuator stroke, which is usually limited to practical values around 5 μm with corresponding actuation voltages around 200 V. Also degradation of the actuator performance is found sometimes in long-term high voltage operation. This is due to the build-up of surface charges at the insulator 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 used 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.

4. Valve design

In the MEMS devices demonstrated above mechanical check valves were used, either with membranes or with flaps. The effort to design and fabricate such valves should not be underestimated. A number of critical 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 can be a critical issue, especially in polymer-fabricated devices. There is also the risk of valve blocking by even small particles, which instantly degrade the pumping performance. This limits the application range of most valve-based micropumps to filtered media.

The so-called "valveless" micropump concept can avoid these problems. The device was introduced by Stemme and Stemme [27]. It uses diffuser/nozzle elements with flow-rectifying properties to mimic the function of a check valve (Fig. 6). According to [27] a maximum achievable forward–backward flow ratio of 2.23 can be calculated for this type of "valve" which is sufficient for a pumping effect. The prototype shown in Fig. 6 was fabricated from 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 approximately 100 mbar were found for water, depending on the diffuser/nozzle geometry in use. This micropump was also able to pump gases.

Based on this prototype various planar silicon micropumps were realized in the following. To reduce inlet and outlet pressure pulses and to increase the pump flow performance, a set-up of two valveless micropumps was realized

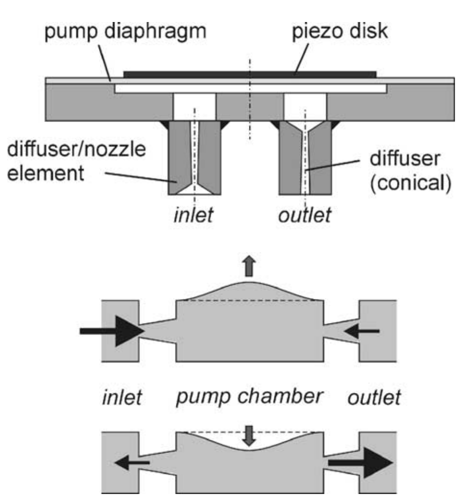

Fig. 6. Valve-less micro diaphragm pump with piezoelectric actuation [27]: schematic diagram (top) and pumping principle (bottom).

by Olsson et al. [28,29]. Another planar design was developed by Foster et al. [30] on the basis of a more complicated flow-rectifying valve structure proposed by Tesla [31]. Vertically-stacked devices are 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 technological effort. A theoretical treatment of the pump principle in conjunction with the corresponding diffuser/nozzle elements is available in a number of publications [30,32–34].

The advantage of valveless micropumps is a relatively simple construction in comparison to pump concepts with check valves or active valves. 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 lack of self-blocking. Any overpressure at the outlet will cause reverse flow that becomes predominant when the pump is switched off. 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 piezoelectric bimorph. Here the dynamic modulation of the gap width between boss and valve seat does result in a flow-rectifying behaviour. The device was able to transport liquids and gases. The pumping direction could be reversed by a variation of the actuation frequency. Reverse flow in the off state could be prevented to a certain extent by simply closing the valve. During operation however, reverse flow was present as in the other valveless designs.

5. Basic dosing properties

Among the large number of micropump designs and concepts the basic dosing properties turn out to be quite similar. Fig. 7 shows the typical behaviour of the pump rate as a function of the actuation frequency and also the typical influence of a varying outlet overpressure [11].

Usually the pump rate is linearly increasing with the operation frequency up to a certain corner value (here approximately 150 Hz). From this linear behaviour the stroke volume ΔV of the diaphragm actuator can be calculated (here approximately 122 nl). In the linear range the actuator is obviously 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 flow 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 as mentioned 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 increases.

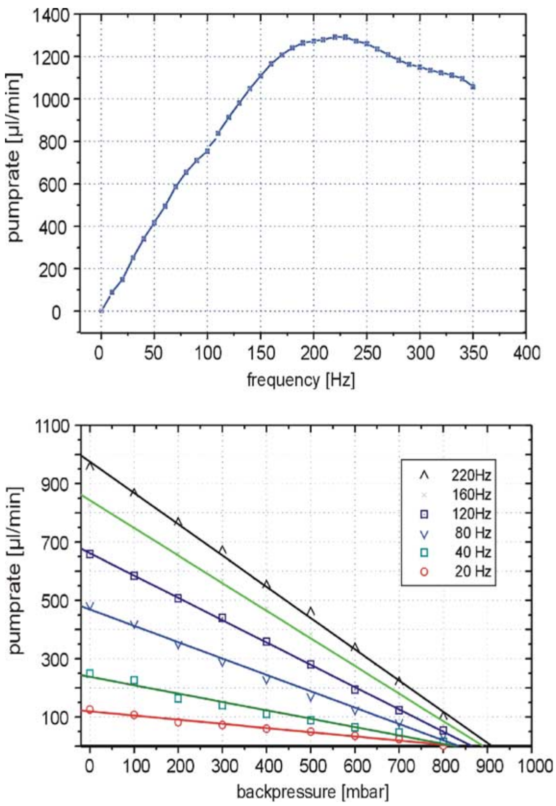

Fig. 7. Typical dosing behaviour of a micro diaphragm pump: pump rate as function of the operation frequency (top) and as function of the applied outlet backpressure (bottom), both with water as medium [11].

At a certain value (here 800–900 mbar), 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. A similar characteristic is found for underpressure at the inlet.

Especially the influence of a varying pressure can pose a limit to the dosing accuracy of a micropump. The application of high force actuation helps to 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. Additionally, metric dosing can be provided via design. Refs. [40,41] describe a precision micropump with a mechanical double limiter that sets the diaphragm movement to a pre-defined value and thus allows only a certain amount of liquid to be transported during each pump cycle. With this concept metric dosing could be demonstrated in a flow range of 0–2 ml/h and for outlet overpressures and inlet underpressures of 200 and −100 mbar, respectively.

None of the earlier micropump designs described above was self priming and bubble-tolerant which meant that even small gas bubbles in the pumpchamber could severely deteriorate the pumping performance of these devices. Many devices even stopped working as soon as a gas bubble arrived at the inlet valve [39]. To avoid this, complicated and unreliable manual priming procedures had to be performed in the first time. A more practical approach was made with a CO_2-purge of the dry device [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.

These problems 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 several physical reasons [35,36]. This discussion was set in 1996 with the first "self-filling" polymer-fabricated micropump by Döpper et al. [13]. A first comprehensive treatment of the subject was performed in 1998 by Richter et al. Using basic laws of thermodynamics and fluid compression, minimum values for the compression ratio ε, i.e. the ratio of the stroke volume ΔV and the pump chamber volume V_0 were calculated and proven with experimental data [39].

Typical ε 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 has to be achieved for a gas pump with a wetted inner structure which meets the case of bubble-tolerance and self-priming.

These design rules were consequently followed in the first—and up to now smallest—self-priming and bubble-tolerant silicon micropump that was realized in 1996 by Linnemann et al., Woias et al. and published in 1998 [11,12]. As seen in Fig. 8, the internal volume of the pump chamber was considerably reduced by means of a combined chemical and mechanical grinding of the middle wafer. Moreover, an optimised piezoelectric actuator was designed capable to perform comparatively high strokes at low chip

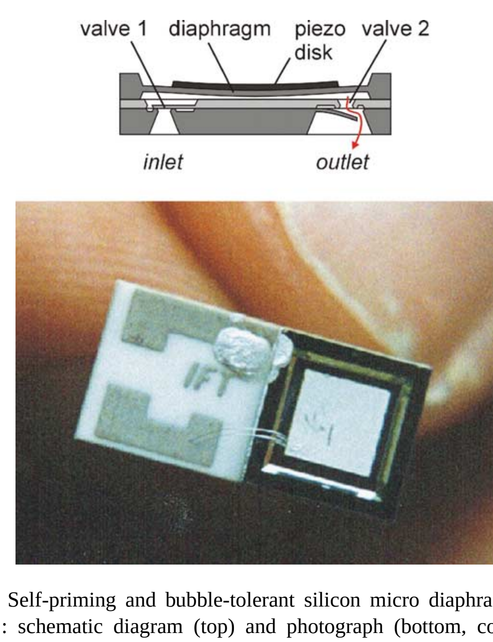

Fig. 8. Self-priming and bubble-tolerant silicon micro diaphragm pump [11,12]: schematic diagram (top) and photograph (bottom, courtesy of FhG-IZM, Munich, Germany).

dimensions. 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 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 × 7 mm × 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 resulted in a limited practical applicability.

6. Materials and fabrication technologies

After the early designs which were realized by conventional machining [1,2], micropump fabrication has soon become an almost exclusive domain for silicon micromachining. Micromachined silicon and glass have been used advantageous due to the high geometric precision available with this technology. Aside from that long-term tests have shown that wear and fatigue of mechanically moving parts, e.g. valve flaps, does not occur in silicon micropumps [12,22]. From this reason, 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]. However, the disadvantages of a rather high fabrication cost and a limited material choice have stimulated the search for alternatives. In the meantime polymer microfabrication, namely microinjection moulding [10,13,42], polymer hot embossing [43] and stereolithography [15,16] have been demonstrated as alternative technologies for micropump fabrication. However, the goal of a true "low-cost" micropump, although often promised, is not satisfied with several of these technologies which are still highly complex and therefore comparatively expensive microfabrication processes. Moreover, other material-related aspects, like limited lifetime can be a critical issue.

In this situation an interesting pragmatic concept was presented by Piet Bergveld's group with the work of Böhm et al. [14,17] who has also done successful research on the later described electrochemical pumping principle. Instead of polymer microfabrication they have used conventional polymer molding as a true low cost method to realize micro diaphragm pumps with piezoelectric and electromagnetic actuation. Other research is using well-established printed circuit board technology for the realization of micropumps and microfluidic devices [8]. These concepts may not provide the performance of thoroughly designed silicon devices, but will definitely deliver an acceptable result at very moderate fabrication costs. They seem to be an interesting choice for all moderate-performance and low-cost applications, provided that other requirements, like reproducibility and operational stability, can be satisfied.

7. Continuous flow micropumps—an overview

In contrary to reciprocating micropumps the so-called "continuous flow micropumps" provide a direct energy transfer and hence a steady flow if required. Various working principles, like electrohydrodynamic, electrokinetic, ultrasonic and RF, magnetohydrodynamic and electrochemical displacement (EC) micropumps are found in literature. A frequent common property of these micropumps is the simplicity of the microstructures involved, as no mechanically moving parts are required. The performance is, however, in many cases strongly influenced by a number of fluid properties which limits a certain principle to a small class of fluids. In most cases gas transport is not possible.

The EHD principle of fluid transport was proposed first around 1960 and picked up 30 years later by Bart et al. [44] and Richter et al. [45,46] for their electrohydrodynamic micropumps. They used 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 micropump of Richter et al. may be taken as a practical example (Fig. 9): silicon chips with identical grid electrodes were realized by anisotropic etching and sputter deposition of gold, bonded together in pairs and placed into a flow channel. High voltage was applied between the electrodes to inject ions into the liquid. Transport of the ions in the electric field between the electrodes was able to induce fluid transport through the grid electrodes.

With voltages up to 800 V a high net flow of 14 ml/min was achieved with ethanol. The maximum outlet pressure was 24 mbar at 700 V actuation. Bidirectional pumping was feasible by reverse poling of the electrodes [45]. In a later publication a similar structure was proposed for flow measurement on the basis of charge injection [46]. The simplic-

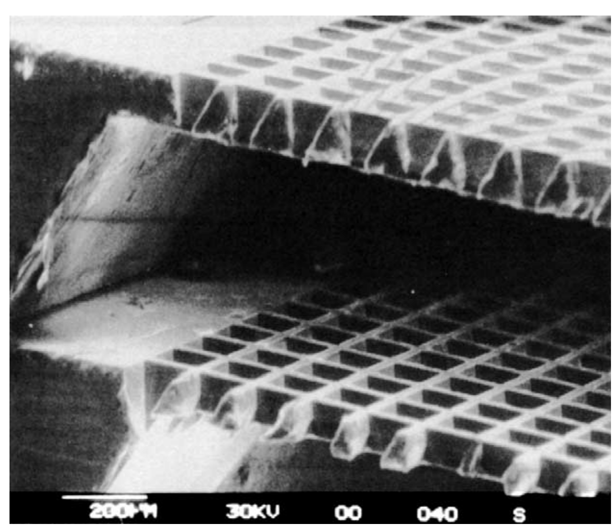

Fig. 9. SEM detail photograph of an EHD micropump [45].

ity of this concept is striking.

However, all EHD pumping principles rely critically on the electric properties, namely permittivity and conductivity, of the fluid to be transported. Typically the electrical conductivity must remain in between 10^{-14} 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 basis of capillary microelectrochromatography which has seen a tremendous progress in science as well as in commercialisation during the last years, mainly pushed by the "lab on a chip"-concept. 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 [48–50]. Among these an interesting publication was contributed by Paul et al. [51], who studied the electrokinetic transport of liquid in microporous media. By simply using a fused silica capillary filled with porous glass 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 atm 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 rather in micromixing or fluid positioning (e.g. the movement of bacteria-loaded liquid in microscopy) than in fluid transport.

Magnetohydrodynamic pumping uses the Lorentz effect. It is based on the injection of an electric current into two electrodes located at facing side walls of a microchannel. This charge injection generates a transversal ionic current in the microchannel which is simultaneously subjected to a magnetic field oriented in an angle of 90° to current direction and microchannel axis. The Lorentz force acting onto the ionic current in the solution will then generate a fluid flow in the microchannel direction. This pumping principle is also bidirectional by nature, as a flow reversal is easily achieved by a reversal of the electric current or the magnetic field vector. Typical micro MHD pumps can generate only small values for pump rate and achievable pressure, e.g. 63 μl/min and 1.8 mbar for the device described in [56] and do strongly depend on the ionic conductivity of the pumped medium. Electrolytic bubble generation at the injection electrodes can easily occur with dc operation. This problem has been solved with an MHD pump with ac current injection which additionally uses an electromagnet instead of a permanent magnet [57]. By driving both the magnet coil and the current injector from synchronized ac sources, the Lorentz vector and, hence, the fluid flow direction remain in the same direction. Choosing the operation frequency high enough will prevent the electrolytic formation of gas bubbles.

While representing a disturbing effect in MHD pumps, the electrolytic generation of gas bubbles can also be exploited as an excellent and simple actuation principle to realize high precision dosing systems. This 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. 10). By a closed-loop control of the gas generation process [60] this dosing system is capable to deliver fluid quantities as small as 100 nl with an accuracy of at least 5 nl.

Centrifugal pumping is a rather new concept of fluid transport. It uses the centrifugal force present in a rotating microchannel system (usually integrated into a disk) to transport fluid from the centre to the perimeter [61,62].

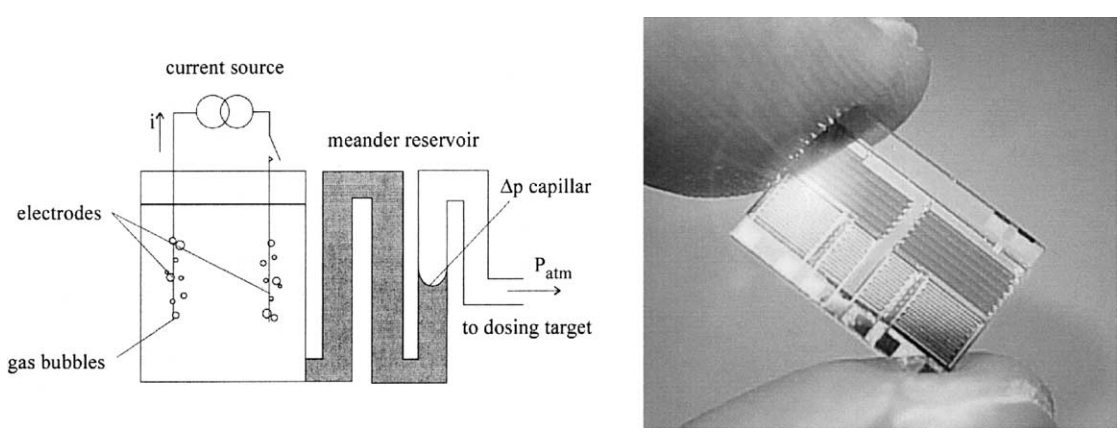

Fig. 10. Electrochemical displacement micropump: function principle (left) and photograph of a chip with two identical devices (right) [59].

The sequence of the fluid flow is controlled by capillary stop valves within the fluidic pathway that can be broken by a stepwise controlled increase of the rotation speed. One distinctive advantage of the concept is the presence of a single technological platform—usually a plastics disk with an integrated channel system—that nethertheless allows a multitude of fluid handling procedures to be integrated via design. Moreover, parallelization of flow-through chemical analysis is feasible without any additional effort by integrating a multitude of identical systems onto one disk. In this case a varying fluid density will influence the global flow characteristics, as it directly influences the centrifugal forces. Another predominant factor is the liquid–surface interaction between the fluid and the channel walls. This requires a thorough control of the surface properties within the fabrication process to achieve reproducible results.

8. Micropumps and μTAS

The concept of a “micro total analysis system” or μTAS proposed by Manz et al. [63] has stimulated widespread efforts to integrate chemical sensors and sample processing into highly miniaturized stand-alone analytical systems. Aside from the miniaturization aspect, the integration of flow-through analysis and chemical sensors provides a number of additional advantages. The combination of ISFETs and micropumps within μTAS research can be taken as an excellent example to discuss these issues.

Also in 1990 Gumbrecht et al. [64] demonstrated a fluidic concept which turned out to be quite popular in the following μTAS research with an ISFET-based pH detection system for clinical monitoring. Their approach was based on a combination of syringe pumps, microchannels and monolithic ISFET arrays instead of a single sensor. It uses the excellent matching of the chemical and electrical properties within a monolithic sensor array together with an appropriate liquid handling for the suppression of common mode noise disturbances. Moreover, a stable electrochemical reference potential is emulated which allows the application of a metal “reference electrode” instead of a fragile and maintenance-intensive glass reference electrode.

A more detailed evaluation of this and other concepts of flow-through analysis was given in [65]. The particular study cited here has used ISFET sensors with Si_3N_4 gateinsulator which are known to exhibit a comparatively high drift and hysteresis. Nevertheless it turned out that a pH detection accuracy of 0.002 was achievable in series measurements with a constant sample pH. When subjected to worst-case measurement conditions, i.e. alternate pH steps of ±1 units, the hysteresis effect of the ISFET sensor was still considerably reduced to approximately 0.01–0.02 pH degrees. This favourable behaviour was also demonstrated under more practical conditions with pH-measurements in an isolated perfused organ system, where pH deviations of 0.04 could easily be detected [66].

Aside from a high measurement precision Gumbrecht’s principle does not require a physical contact between the pumps and the sample solution. Therefore, his approach is well apted for a realization with micropumps which are frequently prone to contamination. This advantage was evaluated first by van der Schoot et al. [67] with a hybrid μTAS made from two micro diaphragm pumps and a K^+-sensitive ISFET as sensing element. Although [67] does not present explicit data on the detection accuracy it is obvious from the measurement curves that this system was able to measure potassium concentrations with high precision and reproducibility. In the following, a number of similar μTAS were realized in the mid 1990s, either based on micropumps [68,69] or on conventional pumps [70] and with potentiometric, amperometric and photometric detectors. This rise of scientific activity was accompanied by the foundation of a dedicated conference series on μTAS in 1994. The first two conference proceedings show a predominant interpretation of the μTAS concept as a combination of mechanical micropumps, microchannel systems and chemical sensors, thus mimicking traditional concepts of flow-through chemical analysis [71,72]. However, concerning the micropumps available at that time, one crucial drawback was originating from their lacking performance which did not include basic requirements like self-priming capability and bubble-tolerance. This and other factors, e.g. high system complexities, mutual interferences between sensors and actuators and problems associated with hybrid or monolithic integration, have delayed the further propagation towards fully integrated systems.

In the meantime the focus of μTAS research has shifted towards electrochromatography thus providing a higher impetus for the further development of elektrokinetic pumping and all associated fluidic devices. In the traditional flow-through analysis the concept of centrifugal fluid transport has arised as an alternative to mechanical micropumps [73,74]. However, within all these paradigm shifts in the μTAS area, the combination of liquid handling, fluid transport and chemical analysis has always provided a very effective and on-going stimulation of microfluidics and micropump research.

9. Conclusions and outlook

The brief—and definitely uncomplete—survey given in this paper reveals an astonishing multitude of micropump concepts, fabrication technologies, devices and applications. As outlined before, the pioneering work has started around 1975 and was shifted towards the MEMS area around 1990. From this time on the exploratory aspect definitely has been the first main driver. Research at that time shows the character of a scientific playground in its most positive sense and was necessarily performed to explore the whole spectrum of micropump concepts, actuation principles, valve designs and fabrication technologies conceivable. The primary goal was to realize a novel microfluidic device and to improve its performance. Aside from that, the mixture of microactuation, microfluidics, microfabrication and application-related aspects present in a micropump has generated an ideal "workbench" for a wide range of MEMS research until today.

The first explorative research phase has soon revealed the advantages and drawbacks of the various micropump concepts. A second phase has started around 1995 with the study of alternative fabrication technologies and genuine micropump-related problems. As a result the performance of micropumps has been constantly increased and features such as self-priming, bubble-tolerance, high flow and pressure capabilities or precision dosing have been addressed and solved.

Looking back at this research history one may criticize that the overall commercialisation process is still in its beginning. However, the development of the ISFET has taught us once more that MEMS devices as many other technical products take an astonishing long time from the first scientific publication to a commercial product [75]. It is therefore more efficient to count the positive aspects and to point out the remaining deficits. On the positive side, mature multidisciplinary knowledge is available today for a number of different micropump concepts. In some cases the optimisation of the fabrication processes has already reached a high level [23]. Also, the first micropumps have entered the marketplace in such promising areas as life sciences and biochemical analysis and others seem to be close to that point [40,41]. On the other side, cost- and fabrication-optimised designs and a good performance reproducibility are definitely the next goals to be achieved. Aside from that the perfection of appropriate design and modelling tools is still a need for both commercialisation and basic research. And yet, aside from all commercialisation issues, 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.

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Biography

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