Peristaltic Pumps

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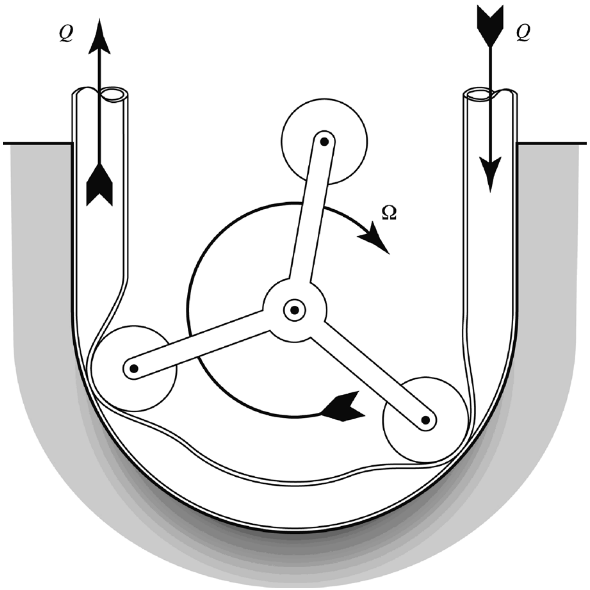
Synonyms

Flexible member pump; Membrane pumps; Diaphragm pumps; Valveless pumps

Definition

Peristaltic pumps are mechanical displacement pumps that induce flow in a fluid-filled, flexible-walled conduit through ▶ peristalsis – transport due to traveling contraction waves. While macroscale peristaltic pumps appear in a variety of configurations, micropumps based on this principle almost exclusively use the sequenced contraction and expansion of a small number of discrete actuators – typically three – placed along the fluid channel.



Peristaltic Pumps, Figure 1 A typical macroscale rotary peristaltic pump design. A set of rotating contact elements cyclically compresses a flexible tube. The elements may roll or slide along the tube; the number of contact elements may also vary. The fluid never leaves the tube, protecting the fluid from contamination by the pump elements, and the pump elements from corrosion or abrasion by aggressive fluids. The tube itself is easily replaced. Tube size and composition may be adjusted to accommodate different fluids, or to change the flow rate. Changing the rotation rate Ω changes the flow rate; changing the sign of Ω reverses the flow direction

Overview

Peristaltic pumps emulate the biological process of peristalsis, in which material is moved through the esophagus or other anatomical passage by the contraction of smooth muscle in rhythmic waves. Figure 1 is a schematic of a typical ▶ rotary peristaltic pump, as implemented at the macroscale, in which a set of revolving contact elements creates a traveling compression wave in a section of a flexible tube. Figure 2 is a schematic of a macroscale implementation of a ▶ linear peristaltic pump, in which a number of translating piston actuators cyclically compress a flexible tube. In either case, a moving boundary displaces fluid and induces a flow, placing peristaltic pumps in the class of positive-displacement pumps. Increasing the tube diameter or the pumping cycle frequency increases the flow rate. A major attraction of macroscale peristaltic pumps is cleanliness. The fluid is completely isolated from the pump components since it never leaves the tube. Furthermore, it is a simple matter to change the tubing to avoid cross-contamination between fluids, and the tubing material may be tailored to ensure compatibility with a particular application. The pumping action is relatively gentle, making peristaltic pumps suitable for reactive liquids or cell suspensions. Also, the pumps may be self-priming due to the low-pressure region created behind the moving constriction, and the flow direction can be easily reversed. At the macroscale, use of rotary peristaltic pumps is widespread, while linear peristaltic pumps are found mainly in niche applications, such as intravenous drug delivery.

Cleanliness considerations do not translate to the microscale, nor does the notion of interchangeable components. Therefore, use of peristaltic micropumps must be justified for other reasons. Self-priming and bi-directionality remain appealing features, but perhaps the main benefit is simplicity of design. Any mechanism providing volume change in a chamber or channel is a candidate actuator for a peristaltic pump. For example, a typical active microvalve involves a membrane that can be forced against a seal to close, or pulled away from the seal to open [1, 11]. Two or more active microvalves can be arranged in sequence and actuated with the appropriate phasing to create a linear peristaltic micropump. The principles of peristaltic pumping apply to liquid or gas, and micropumps for both have been demonstrated. The ease with which an actuator can be transformed into a pump accounts for the wide variety of linear peristaltic micro-

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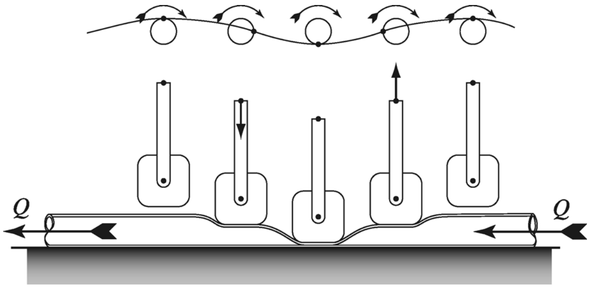


Peristaltic Pumps, Figure 2 A typical macroscale linear peristaltic pump design. A set of translating actuators cyclically compresses a flexible tube. The number of actuators may vary. Features of macroscale rotating peristaltic pumps also apply to microscale linear peristaltic pumps. Linear peristaltic pumps are used in applications such as drug delivery, when accurate flow rate control is needed

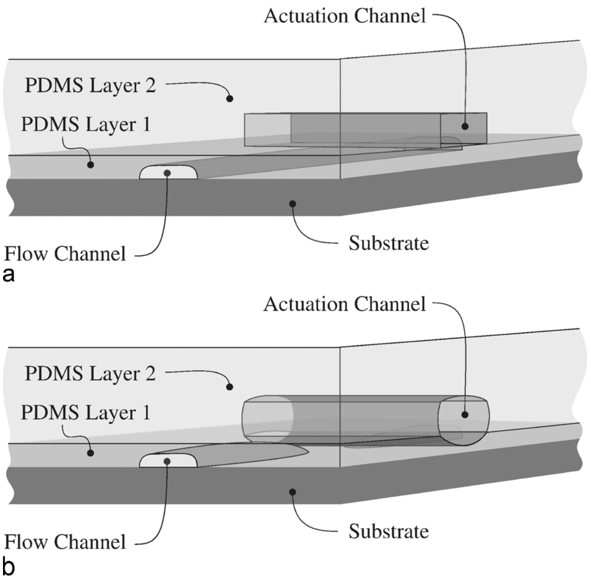
pumps that have been reported, including an early micromachined silicon and glass version by Smits (as reviewed in [2, 11], or see the original paper referenced therein).

In this article we refer to an actuator as *closed* when the associated chamber or channel volume is minimized, and as *open* when that volume is maximized. The volume in the closed position is called the ►dead volume, and the difference between the volume in the open position and the dead volume is the ►stroke volume. The ►stroke volume divided by the dead volume is the ►compression ratio. Compression ratio is particularly important when pumping gases, since the amount of gas ejected from the actuator is reduced by the compressibility of the fluid. In pure liquids this effect is much less, but compression ratio plays an important role in determining how severely bubbles impact a particular pump. In classifying micropumps, a distinction is sometimes made between ►integrated actuators – fabricated along with the other components of the microfluidic system – and ►external actuators – fabricated separately and attached later. It is common to see peristaltic micropumps of either kind.

While a linear pump can easily be built up from two or more basic actuation units, microscale rotary mechanisms face significant fabrication challenges. Accordingly, rotary peristaltic micropumps have seen very limited implementation. While several rotary mechanical micropumps are described in the literature (see the summaries in [1], [2] and [11], for example), to our knowledge there is only one example in the literature of a true rotary peristaltic micropump; Kim et al. (Sensors and Actuators A 128:43–51, 2006) use a silicone membrane for the deformable channel surface, and a rotating ferrofluid plug driven by an external magnet for actuation. Also notable is a miniature rotary peristaltic pump by Bar-Cohen and Chang (*Smart Structures and Materials 2000: Active Materials: Behaviors and Mechanics*, *Proceedings of SPIE 3992, 2000*) that employs a traveling flexural wave in a piezoelectric ring. Reflecting their dominance in the literature, the subsequent discussion will consider only linear peristaltic micropumps, which for brevity are henceforth referred to simply as peristaltic micropumps. Most reported microscale implementations approximate continuous peristaltic action using two or more (most commonly three) discrete collapsible chambers. A notable exception, and one which does not fit well into the analysis scheme presented here, is the colloidal peristaltic pump presented in [14]. There a large number (approximately twelve) of 3 μm colloidal silica particles are oscillated sinusoidally in a 6 μm deep polydimethylsiloxane (PDMS) channel using external actuation by optical trapping with a scanned 532 nm laser. Flow rate on the order of 1 nl/h are induced in an aqueous solution by the resulting traveling wave. Peristaltic actuation through movable suspended particles is, to our knowledge, unique to [14]. The remainder of this paper focuses on the much more common case of discrete collapsible chambers.

Peristaltic pumps have found wide application in microfluidic systems based on PDMS, due to their ease of fabrication and versatility in operation. PDMS is a popular material for microfluidics for several reasons. Compared to traditional semiconductor industry fabrication schemes, both the materials and processing tools are inexpensive. PDMS is biocompatible, resistant to many chemicals, and transparent to wavelengths of light useful in chemical and biochemical analysis. Figure 3 shows a typical actuator design, similar to that first reported in [3], with the fluid channels formed between a thin PDMS layer and a glass, silicon, or PDMS substrate, and actuator channels formed at right angles to the fluid channels in a thicker PDMS layer, bonded to the planar side of the fluid layer. We refer to the area of the fluid channel under an actuator channel as a pump chamber, even though it is geometrically identical to the rest of the channel. The actuator channels are inflated from an external pressure source to close the pump chamber; when the pressure is removed the natural elasticity of the PDMS restores the channel to its original shape. As suggested by the figure, due to the low modulus of elasticity of PDMS, the pump chambers may be designed to collapse completely when closed, allowing them to act either in concert for pumping, or in isolation as pure valves.

In a peristaltic pump, the volume of fluid displaced by the individual actuators is rectified, that is, directed in the desired direction of flow, solely by the phasing of multiple actuators. Other types of displacement pumps rectify flow differently. A passive check-valve pump uses two passive, one-way check valves incorporating moving flaps or other



Peristaltic Pumps, Figure 3 A typical pump chamber in a multilayer PDMS implementation (similar to those of [3]). The chamber is open by default, as shown in (**a**). Inflating the actuation channel with compressed air closes the chamber, as shown in (**b**). PDMS layer 1 defines the fluid channels (sometimes called the working channel), while PDMS layer 2 defines the actuation channels. In both cases casting or spinning PDMS on a negative mold creates the channel structures. The molds are defined photolithographically, with standard positive resists used for thin channels (on the order of a micrometer) and negative resists such as SU8 used for thicker channels (up to one millimeter). The patterned side of the fluid layer is bonded to a glass, silicon, or PDMS substrate, and the patterned side of the actuation layer is bonded in turn to the planar side of the fluid layer. For convenience, we refer to the area of the fluid channel under the actuation channel as the pump chamber, even though it is otherwise indistinguishable from the rest of the channel. Baking the fluid layer mold at high temperature after the pattern is developed causes the photoresist to *reflow*, yielding rounded corners in the fluid channels. These rounded corners reduce leakage when the chamber is closed

components. One such valve at the pump inlet permits fluid to enter the chamber as the pump chamber expands, but prevents it from leaving as the chamber contracts. Likewise, a passive check valve at the outlet prevents backflow on the expansion stroke, but permits fluid to exit the chamber on the contraction stroke. Valveless rectification pumps are similar, but use flow diodes instead of mechanical check valves. Flow diodes are diffuser/nozzle structures or so-called Tesla valves that have no moving parts, but exhibit a preferential flow direction due to their geometry. More detail on passive check-valve or valveless rectifier pumps may be found in [1], [2] or [11]. Since these require only a single actuator, versus two or more for peristaltic pumps, they are natural design alternatives. Reasons to prefer a peristaltic design include ease of integration onto the microchip, with no special fabrication steps or external connections. For example, [12] presents an electrostatically-actuated peristaltic micropump suitable for incorporation into a micro-total analytical system (μTAS) with no additional special considerations. Peristaltic pumps avoid the more complex microfabrication typically required for passive check valves, and the susceptibility of such valves to fouling by particles or bubbles. Valveless rectifiers are simple to fabricate, and not prone to fouling, but may have a larger footprint than peristaltic designs. Finally, peristaltic pumps may be used to generate forward or reverse flow by modifying the actuator phasing, while bidirectional versions of the other two options are difficult to realize. Some guides to peristaltic pump design advise adding series passive check valves to aid in rectification if the application allows [1], though this reduces to some extent the operational and fabrication benefits that peristaltic designs offer over check-valve designs.

In many peristaltic micropump implementations identical actuators provide both stroke displacement and flow valving. However, these functions may also be divided between actuators of different design. One analysis reported in the literature treats a large stroke displacement actuator with a low flow resistance, flanked by active valves with small stroke displacement [4]. While similar in concept to a passive check-valve diaphragm pump, the valving actuators in this case are actively opened and closed by external commands, and flow direction is determined by the phasing of these commands with regard to actuation of the central diaphragm. Hence, this design may properly be classified as a peristaltic pump.

Due to a combination of surface tension and viscosity effects, the pressure drop per unit length along a channel with characteristic length l (radius or hydraulic diameter) scales unfavorably (as $1/l^3$ or $1/l^2$, depending on the analysis). Thus mechanical pumps must generate extremely high pressures to maintain flow rates as channel diameters shrink. For this reason, it is widely accepted that mechanical pumping mechanisms, including peristaltic pumps, are not suitable for very small channels. Nguyen and Wereley survey the literature, and, using water as a working fluid, give 10 μl/min as the flow rate below which nonmechanical pumping principles are preferred [1]. However, that boundary is somewhat arbitrary, as demonstrated by flow rate of 0.14 μl/min reported in [3] for pneumatically-actuated peristaltic pumping of water in PDMS channels, the flow rate of 1.7 nl/min for electrostatically-actuated peristaltic pumping of ethanol in parylene channels [12], and the flow rate of 1 nl/h reported in [14] for optically-actuated peristaltic pumping by colloidal particles suspended in an aqueous solution.

Basic Methodology

A linear peristaltic micropump consists of two or more actuators arranged along a channel, and driven in a particular sequence. Important parameters that quantify pump performance are the maximum flow rate, denoted here by Q^*, the maximum pressure that can be generated between the pump inlet and outlet (called the ►pump head), denoted here by p^*, and the efficiency with which the pump converts external power into flow power, denoted here by η. Given a particular actuator, or set of available actuators, pump design consists of selecting the number of actuators, the actuation sequence, and the frequency of actuation. Important information about the individual actuators includes the stroke volume, V, the dead volume, V_0, the flow resistance in open and closed configurations, R_o and R_c, respectively, the maximum pressure the actuator can generate, p^*_{act}, called the ►closing pressure, the minimum time required for the actuator to either open or close completely, T_{min} (the minimum time to open and the minimum time to close are typically different; here T_{min} would be the larger of the two) and the energy or power required to operate an actuator. Almost any mechanical transducer can be used as an actuator in a peristaltic micropump, making generalizations about pump performance difficult. For example, the response time of a thermopneumatic actuator may be dominated by heat transfer out of the actuator, while that of a piezoelectric actuator is primarily determined by material stiffness and membrane geometry. As another example, a thermal actuator may require continual power to remain in the closed position, while – once closed – an electrostatic or pneumatic actuator may be held closed with little or no additional power consumption.

In this article we focus on those considerations that are common to all peristaltic pump designs. This rules out detailed physical analysis. However, some broad assumptions about actuator and system characteristics allow rough comparisons for pump design purposes. We begin by assuming ideal actuators, with infinite flow resistance in the closed position and zero dead volume. The first property allows easier accounting of the movement of fluid, the second allows issues related to fluid compressibility to be ignored.

The first step in this simplified analysis is estimating the volume of fluid transported in a complete pumping cycle. To do this, we apply the following rules:

1. When an actuator closes, the fluid within is completely expelled.
2. When an actuator opens, it completely fills with fluid.
3. If an actuator remains closed during a step in the sequence, no fluid may pass across it.
4. If, in a single step, some actuators open while others close, and there is a path for fluid to flow between them, then fluid is transferred from the latter to the former.
5. If an actuator opens or closes with no closed actuators between it and either the inlet or the outlet, the transfer of fluid is made equally from or to both sides. Once the fluid volume is computed, the flow rate flows from the actuation frequency. Maximum head is taken to be the maximum pressure exerted by an actuator (though this is not often known), and efficiency is estimated from the actuation sequence. Often, an estimate of flow rate is all that is desired. It must be noted that these estimates are very coarse, with for examples, errors of a factor of two reported, and attributed to unmodeled valve leakage [5].

Simplified Sequence Analysis

As an illustration of the simplified analysis, consider the three-step cycle shown in Fig. 4 for a pump consisting of three identical ideal actuators with stroke volume V. At the start of the cycle, no fluid has yet been transferred. In step 2, chamber $C1$ opens while chamber $C3$ closes. By rules 1 and 2, $C1$ must fill with volume V while $C2$ expels volume V. Since the intervening chamber $C2$ is closed, by rule 3 $C1$ must take volume V from the inlet, while $C3$ must expel volume V to the outlet. In step 3 $C1$ closes while $C2$ opens. By rule 4, since there is an open path between them, this transfers volume V from $C1$ to $C2$. Likewise the return to step 1, which completes the cycle, transfers volume V from $C2$ to $C3$. We conclude that, ideally, volume V is pumped per complete three-step cycle. At the final step the net fluid volume increase at the outlet must equal the net decrease at the input. This is a useful check that the analysis was done correctly.

Maximum Flow Rate

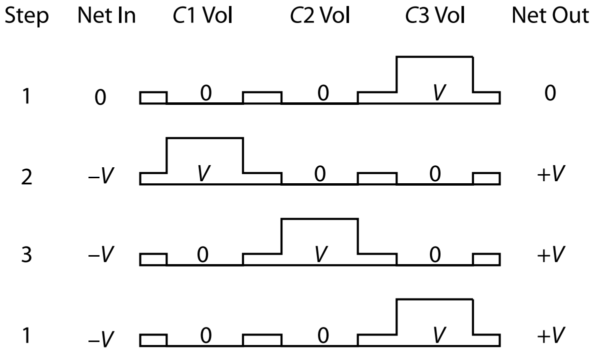
If each step requires time T, the pumping rate is $V/3T$. The maximum flow rate Q^* then corresponds to the minimum cycle time, T_{min}: $Q^* = V/3T_{min}$.

Maximum Head

In step 2, $C3$ is emptied to the outlet. If the outlet pressure is greater than p^*_{act}, the maximum pressure the actuator can provide, the chamber will be unable to close. Thus, the maximum head can be approximated as the closing pressure, $p^* = p^*_{act}$.

Efficiency

Fluid power is the product of flow rate by head. In an ideal ►positive displacement pump, flow rate is independent



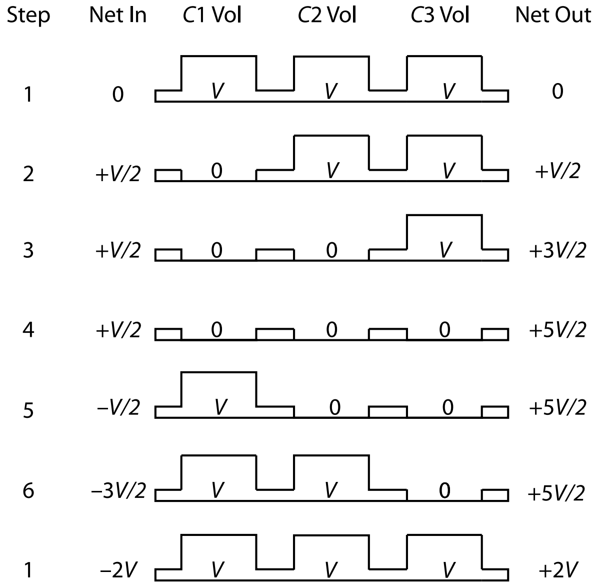
Peristaltic Pumps, Figure 4 A three-step, three-actuator pumping sequences, used in numerous pumps, including thermopneumatic pumps by Grosjean and Tai (see the reference in [1]) and Jeong, et al. [8]. The stroke volume is designated V. Chambers 1, 2, and 3 are abbreviated $C1$, $C2$, and $C3$. At every step two chambers are closed, reducing leakage. Note that Grosjean and Tai use a similar looking figure to define the pump sequence, but in their paper the raised level indicates actuation, which corresponds to *closing* the chamber

of head. Therefore, the maximum fluid power is approximately $P^* = p^* Q = p^*_{\text{act}} V/3T$. One actuator closes at each step, and two actuators must be held closed at each step. If energy E_{act} is required to close an actuator, and power P_{act} is required to hold it closed, we see that each cycle requires total energy $3E_{\text{act}} + 6P_{\text{act}}T$ (we assume no energy is required to open an actuator). The total power consumed per cycle is $(E_{\text{act}}/T) + 2P_{\text{act}}$, and the maximum efficiency is $\eta = p^*_{\text{act}} V/(3E_{\text{act}} + 6P_{\text{act}}T)$.

Experiments suggest that a linear relation between flow rate and pump head may be a more appropriate model [4]. In that case, the maximum fluid power is given by $P^* = p^* Q^*/4$.

Design Comparison

To illustrate the use of the simplified analysis in design, we compare the actuation sequence analyzed above to a six-step three-chamber sequence also reported in the literature [5], and shown in Fig. 5. The analysis is similar to the three-step sequence, except for steps 1 and 6, where an actuator closes with no closed actuators between it and either the inlet or the outlet. By rule 5, the expelled fluid is divided evenly between the inlet and the outlet. Two considerations make application of the simplified model more problematic in this case. The first is the all-actuators-open configuration in step 1. If there is significant head across the pump, this may produce an unmodeled backflow. The second is the assumption mentioned above, that the flow splits evenly in step 1 and 6. If the external system flow resistances as seen from the input and output are different, or if there is significant head, the split will *not* be even.



Peristaltic Pumps, Figure 5 A six-step, three-actuator pumping sequences proposed in the literature [5]. At step 1 all chambers are open, allowing backflow which will reduce the net flow rate

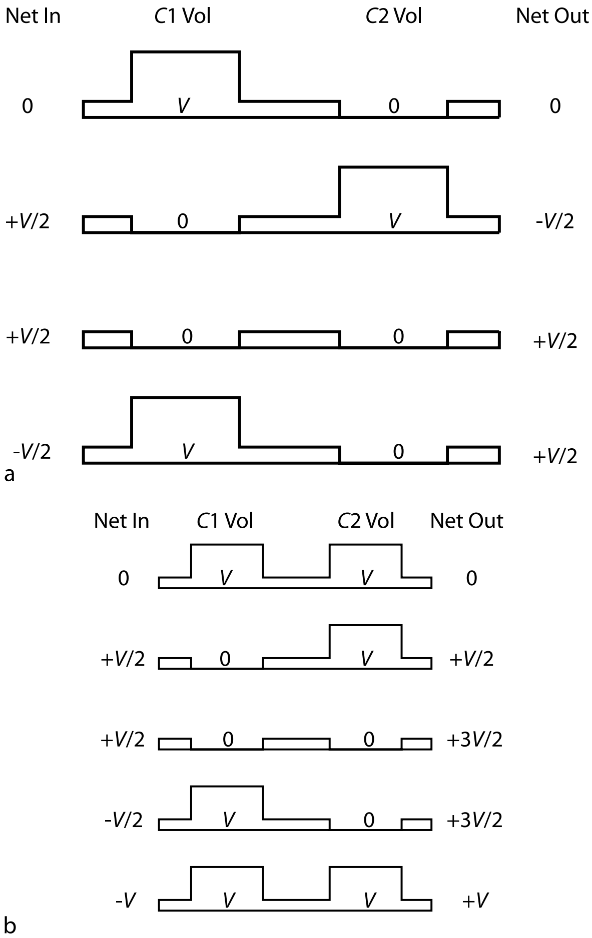
These issues aside, we find that a volume $2V$ is pumped in each six-step cycle.

If each step requires time T, the pumping rate is $2V/6T$, or $V/3T$. The maximum flow rate is $Q^* = V/3T_{\text{min}}$. This is the same as for the three-step sequence.

In steps 2, 3, and 4 chambers are emptied to the outlet. The maximum head can again be approximated as the closing pressure.

The maximum fluid power is approximately $P^* = p^* Q = p^*_{\text{act}} V/3T$. In each cycle, energy $3E_{\text{act}}$ is required for actuator closings, and energy $9P_{\text{act}}T$ is needed to hold actuators closed, for a total cycle power requirement $(E_{\text{act}}/2T) + (2P_{\text{act}}/3)$. The maximum efficiency is $\eta = 2p^*_{\text{act}} V/(3E_{\text{act}} + 9P_{\text{act}}T)$.

To the level of modeling accuracy applied, the two designs have the same maximum flow rate and maximum head. The six-step sequence is more power efficient, but there are unanswered questions about how well this design will perform, particularly if there is significant backpressure. If power usage is more important than maximum head, the six-step design may be preferable, but further investigation will be required. If efficiency is not important, the three-step sequence should be chosen for its higher resistance to backflow. An advantage of the six-step sequence is that only one chamber is actuated per step. This makes the sequence insensitive to asymmetries in opening versus closing actuators. For example, if the actuators close ten times faster than they open, fluid may not transfer



Peristaltic Pumps, Figure 6 Pumping sequences for two-actuator peristaltic pumps [6]. Sequence A has at least one actuator closed at all times. Sequence B has a higher maximum flow rate, but may have problems with backflow under high head conditions

smoothly between chambers in the sequence of Fig. 4, but instead may leak out of the chamber. Avoiding this risk by adding intermediate steps, as for example, in the five-step sequence used by Smits [2] TS3 will reduce the maximum flow rate, and make the sequence of Fig. 5 more attractive. Most reported peristaltic micropump designs use three actuators. A few use more, but typically not more than five. Since the actuators take up potentially valuable space, and require connections for power and switching, it is desirable in many applications to use the minimal possible number. Some authors have stated that three chambers are the fewest possible for peristaltic pumping. However, the sequences shown in Fig. 6 have been shown to provide peristaltic pumping using only two chambers [6].

Diagrams like Figs. 4–6 are easy to grasp visually, but a more compact description of pumping sequences is also useful. One method used in the literature (for example, in [7]) is to indicate an actuated (closed) chamber by a 1 and an un-actuated (open) chamber by a 0. Thus the sequence in Fig. 4 is written {110, 011, 101}, that of Fig. 5 becomes {000, 100, 110, 111, 011, 001}, and those of Fig. 6 A and B are {01, 10, 00} and {00, 10, 11, 01}, respectively. The five-step sequence used by Smits [2] TS4 is, following the convention of 1 for closed, 0 for open, {011, 001, 101, 100, 110}. (This sequence does involve an actuator closing simultaneously with another opening, but they are separated by a closed chamber.) Some care is required in using this notation, since some mechanisms close when actuated (e. g. thermopneumatic), while others open (e. g. piezoelectric).

Dynamic Modeling

Leaks, backpressure, and actuator dynamics all influence the performance of peristaltic pumps. Leaks and backpressure effects will alter the distribution of fluid as actuators open and close. The dynamics of the actuators determines the maximum actuation rate, which in turn limits the maximum flow rate. These effects can be incorporated into lumped-parameter models for analysis and simulation. We do not pursue this further in this article, but refer the reader to the literature. One general approach is presented in [4]. Also relevant for further study are [5], [12] and [13], which present dynamic models for pneumatic and electrostatic pumps, respectively. All of these works are applied to liquid pumps. For gas pumps, or robustness to bubbles, compressibility becomes a factor. Some considerations of micropumps for compressible fluids may be found in [1]. Finite-element analysis of individual chambers can also be used to obtain detailed predictions of pump dynamic performance.

Key Research Findings

Fabrication Materials and Actuator Types

Peristaltic pumps have been demonstrated in all materials common for microfluidic applications, using a variety of physical principles. Space constraints prevent a comprehensive listing, but [1], [2], and [11] provide good overviews. Summarizing here just the main fabrication materials, the operating principle, and the working fluid, these include for silicon and/or glass devices, piezoelectric, thermopneumatic, and electrostatic actuators pumping water. For PDMS devices only pneumatic pumping of water is reported. For plastic or mixed-type devices, electrostatic, thermopneumatic, and pneumatic actuators are reported, pumping water and air. A novel traveling wave pump implemented using optically-actuated col-

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Editor's or typesetter's annotations (will be removed before the final TEX run) loidal silica spheres is reported in [14]. Recently, Husband et al. (Microelectronic Engineering 73–74:858–863, 2004) report a piezoelectric silicon-glass device pumping water and air, Pan et al. (Proceedings of the 26th Annual International IEEE EMBS Conference, 2004) report a magnetically driven PDMS pump for water, and Jeong et al. report a thermopneumatic PDMS pump for water [8]. Electrostatic pumps implemented in parylene for integration with μTAS devices are reported in [12].

Reducing Device Complexity

Peristaltic pumps already offer simpler fabrication than passive check-valve pumps. Even the simplest actuators, however, require at least two patterned layers, as, for example, in the PDMS device of Fig. 3. However peristaltic pumping was recently demonstrated in a device with a single patterned layer of PDMS on an un-patterned glass or PDMS substrate [9]. The single patterned layer incorporates both pneumatic actuation lines and the water-filled working channels. Rather than press on the fluid channel from above, the actuation lines in this device squeeze the working channel from the sides. While these actuators provide enough stroke and increased resistance for pumping, cell sorting, and mixing, in their closed configuration they are still quite leaky, and do not completely seal the channel. However, the elimination of the second layer, along with the associated alignment and bonding tasks, should yield a significant reduction in device cost.

Reducing Number of Off-Chip Connections

Each actuator in a peristaltic pump is an active element, requiring switching and power. In applications such as Lab-on-a-Chip, the number of required off-chip connections can become very large. As mentioned above, one means for doing this is by using two actuators instead of three [6]. The proliferation of pumps and valves is a particular problem in pneumatically actuated pumps that require external valving. An innovative approach to ameliorating this problem is described in [10] for pneumatically actuated pumps fabricated in PDMS. Here, a single pneumatic actuation channel has a serpentine shape, crossing over the straight channel containing the working fluid at multiple intersections. The phasing of the actuation is provided by the natural delay associated with the propagating inflation zone in the serpentine channel. In this way the equivalent of a three-, five-, or seven-actuator peristaltic pump is realized using a single actuator line.

Multifunction Components

The individual actuators in a peristaltic pump can also serve as individual valves. Additional flexibility in integrating peristaltic pumps into large microfluidic systems can be gained from providing components with multiple functions. An example of this is the rotary mixer described in [7], which can serve as an effective dynamic micromixer, or as a peristaltic pump, depending on the actuation pattern of the individual valves.

Future Directions for Research

Progress in peristaltic pump design has paralleled the advancement of microfluidics as a field. As new materials and applications have been developed, peristaltic pump designs have soon followed. This trend seems likely to continue.

Peristaltic pumps are commonly used in PDMS-based systems. As those systems have attracted increasing amounts of research effort, due to their biocompatibility and demonstrated success in manipulation of cells, genetic material, and proteins, new innovations in PDMS-based pumps have followed. Progress has been made in reducing the need for external connections, but improvement would be welcome in the area of fully integrated pumps of reasonable cost and low complexity.

Cross References

► Electromagnetic Valves
► Electrostatic Valves
► Magnetic Pumps
► Membrane Actuation for Micropumps
► Microactuators
► Microfluidic Dispensing
► Microfluidic Mixing
► Microfluidic Pumping
► Piezoelectric Valves
► Piezo/PZT in Microfluidics
► Pneumatic Valves
► Soft Photolithography
► Thermomechanical Valves
► Thermopneumatic Valves
► Versatile On-Chip Liquid Handling Techniques

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