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A FIRST INHERENTLY PULSATION FREE PERISTALTIC PUMP.

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ABSTRACT

As all currently available peristaltic pumps produce a pulsed flow this article describes the principle and the development of a pulsation-free peristaltic pump. Pulsation can be avoided if at least four pumping chambers are arranged in a line and when actuated by a specially developed flow schedule. Our findings show that it is possible to build a pulsation-free linear peristaltic pump. A prototype shows a remaining pulsation of only 11% of the mean flow rate. But this pulsation can be further reduced when paying attention a trade-off between manufacturing expenses and attainable flow-consistency.

Index Terms - pulsation-free, peristaltic pump, tube pump, flow rate

1. INTRODUCTION

A variety of fluidic applications in chemical engineering, biotechnology and medical technology requires a constant and pulsation free flow, for example for time stable mixing processes or chemical reactions in microfluidic systems. Commonly used pumps for this purpose are so called peristaltic pumps. Even though they offer a lot of advantages, all currently available peristaltic pumps have the one major drawback. They are pulsatile. Hence they generate a flow which fluctuates cyclically in the magnitude of multiple percent of the average flow rate.
First we show in section 2 the main advantages of peristaltic pumps, which make them desirable for a variety of application. In section 3 we describe the two general designs of peristaltic pumps and it effect on the tubing fatigue. Subsequently the theoretical background of a simple designed pulsation free pump is given in section 4, followed by the experimental setup and the achieved results in section 5 and 6.

2. PERISTALTIC PUMPS

Peristaltic pumps are a type of positive displacement pumps. Positive displacement pumps in general use the volume variation of a pumping element, referred to as the pumping chamber, to propel the fluid. An increasing volume generates suction and a decreasing volume generates ejection. This working principle provides a nearly backpressure independent flow rate and maintains high efficiencies throughout the viscosity range [6].
In accordance to the peristalsis known from the propagation of contraction waves along muscular tubes in biological systems, i.e. for the transport of food through the digestive tract, the peristaltic pumps generate flow by periodical occlusion of a flexible tube to push a fluid in one direction. Therefore peristaltic pumps are also referred to as tube pumps.

Since the only pumping element in these pumps is a flexible tube there are a lot of practical benefits:

- Seal-less and leak-free: As only the tube comes in contact with the fluid, the fluid is completely isolated from the mechanical parts of the pump and vice versa. That minimizes the risk of cross contamination and exposure to corrosive or toxic substances. A worn or polluted tube can easily be replaced.

- Valve-less: To prevent internal backflow, only the occlusion of the tube is necessary. Since there is no risk of blocking, the pump can handle viscous or abrasive liquids, such as slurries.

- Self-priming and dry running: Fluids with entrained gasses and even only gasses can be pumped.

- Reversible: Due to the peristaltic principle the pump can be used in both directions. The pumping direction can be changed by reversing the drive direction.

Peristaltic pumps are also known to be relatively gentle, which makes them suitable for handling of shear sensitive products such as reactive liquids and cell suspensions [1, 4]. However, pumps used to circulate blood in bypass surgery are reported to cause blood damage and haemolysis, mostly caused by the full occlusion when the tube is completely compressed [2]. It has been shown that the blood damage can be reduced by slightly under-occlusive pump calibration [2]. However, the disadvantage of an under-occlusion is a small internal backflow in the pump, causing a backpressure dependence of the flow rate.

3. PUMP DESIGN

There are two functional designs for peristaltic pumps. The pumps can be build either in a rotary or in a linearly manner.
The common design is the rotary peristaltic pump that uses revolving contact elements to periodically compress the tube. These pumps are also referred to as roller pumps, since primarily rollers are used as contact elements. In most cases the tubing is placed in a raceway implemented in the pumps housing and aligned around a rotor with multiple rollers that press against the tube. To get the pump working without backflow, there are at least two opposing rollers necessary. Using more rollers will result in lesser pulsation [4] but will increase the wear of the tubing as well, since there are more occlusions in the same time [1].
In linear peristaltic pumps the tubing is placed flat on a platen and sequentially compressed by a peristaltic mechanism. The peristaltic mechanism consists of a set of at least three translational actuators and is usually driven by a camshaft. Linear peristaltic pumps are found mainly in niche applications for medical purpose were accurate flow rate control is needed [1, 3].
The advantage of the linear peristaltic pump was clarified by Peek et al. as they studied the time to tubing rupture and the failure pattern of tubes in a rotary peristaltic pump [7]. They identified three different failure patterns. Longitudinal creases are the result of multiple repeated folding of the tubing across its axis during the occlusion. Transverse tears and scallop-shaped defects in contrast indicate a high sheer stress on the tubing. In an additional experiment they showed that pure compression of the tube without sheering causes significantly less wear and did not initiate longitudinal creases [7]. That means that in rotary peristaltic pumps the movement of the occlusion zone is responsible for the wear of the

tubing. Therefore recalibration or changing of the tube is more often required than in linear peristaltic pumps. Hence the linear peristaltic pump provides better long term stability of the flow rate and better overall endurance of the tube.

4. THEORIE OF PULSATION-FREE PUMPING

Radial and linear peristaltic pumps produce a pulsatile flow. Nevertheless, according to a sequence plan described by Feller & Schimmelpfennig [5], it should be possible to realize a pulsation-free flow for every type of positive displacement pump with five linearly arranged pumping chambers.

In a linear peristaltic pump at least three chambers are needed to produce a directed and backflow-free flow at the outlet. One chamber has to act as a valve and fully occlude a tubing section, while the other two chambers either charge or discharge the fluid. The flow is generated by creating a traveling wave according to the peristaltic principle. Thus several steps are required to generate a pumping cycle.

There are two general ways to describe a pumping cycle. The first method is to indicate the chamber status at the beginning or at the end of each step. A closed chamber can be indicated by a 1 and an open chamber can be indicated by a 0. In this case the sequence of a three-camber pump can be written as 101, 110, 011.

However this is only a static description of the pump action. Therefore we prefer the second method that indicates the change of the chamber volume during the steps. A closing chamber that ejects fluid is then indicated by a 1 and an opening chamber that charges fluid is indicated by a -1. The 1 represents the normalized chamber volume, assuming that all chambers have the same size. Furthermore, a closed chamber is indicated by # and an opened chamber that does not contribute to the flow but does not prevent flow as well is indicated by a 0. The sequence of a three-chamber pump can be written as #1-1, -1#1, 1-1#. As # indicates the closed chamber, the sum of all values right of # represents the output volume of the step. It can be seen, that only during one step fluid is ejected at the outlet, while no flow is produced during the other steps. That means the efficiency of the pump is 1/3.

The efficiency can be improved to 2/3 if the volume of central chamber is twice that of the other cambers. In this case the sequence is #2-1, -1#1, 1-2#.

To further improve the efficiency to 1, there are at least 4 chambers required and the second chamber has to have three times the volume of the other chambers. Thus the sequence is written #3-1-1, -1#10, 1-3#1 and in every step one normalized volume is ejected at the outlet. Assuming that the three steps are isochronous and that the volume changes are linearly in time, there will be a continuous and pulsation-free flow at the outlet.

In their sequence plan shown in Figure 1 Feller & Schimmelpfennig use an additional chamber at the inlet. Thus a symmetric contribution of the chamber volume in the form of 11311 (chamber 1 till 5) can be achieved and the flow direction can easily be reversed while staying pulsation-free. The pump will then work pulsation-free at the outlet and the inlet as well. [5]

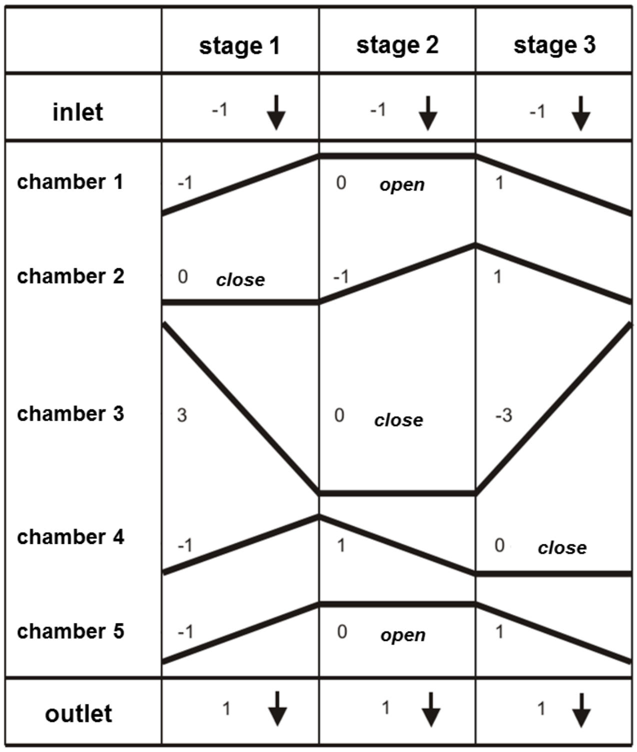

Figure 1: Sequence plan as described by Feller and Schimmelpfennig [5]. One pumping sequence consists of three isochronous stages. The lines indicate the volume change in the pumping cambers over time. The numbers (-3, -1, 0, 1, 3) indicate the normalized incremental volume, charged or ejected by the pumping chambers. A negative number represents inflow in a pumping chamber while charging fluid. Hence a positive indicates outflow by ejection fluid from a pumping chamber.

5. MATERIAL & METHODS

The sequence plan in Figure 1 describes the time characteristic of the flow in the pumping chambers that is required for a continuous pulsation-free flow. The prerequisite is that each pumping chamber produces a time constant flow rate. However, while compressing a tube, the inner cross section does not change linearly with the compression. This means, a constant compression rate produces a non-constant volume change and therefore a non-constant flow rate. In order to get a constant flow rate, the applied compression rate has to be modulated over time in an appropriate manner.

The characteristic relation between tubing compression and displaced volume was measured by stepwise compression of a tubing segment and measuring the displaced mass. The volume was calculated from the displaced fluid mass according to the equation

$$V = \frac{m}{\rho}$$

where V is the fluid volume, m the fluid mass and ρ the density of the fluid at 20°C.

For stepwise compression, the linear drive of a syringe pump (neMESYS, cetoni GmbH, Germany) was used. The displaced mass was measured by an analytical balance (AM50, Mettler-Toledo GmbH, Germany) with 0.1 mg resolution. A segment of the tubing was placed on a custom made table attached to the linear drive housing and a plunger of the same shape as in the pump was attached to the linear drive. The tubing segment was filled with fluid and locked at one side via a stopcock. The other side of the tubing was connected to the analytical balance via a line system. The displaced fluid was collected in a small beaker on the balance.

Prior to the measurement the beaker was filled with a small amount of water and the line system was adjusted to end underneath the waterline. To avoid evaporation losses over time a thin layer of oil (Ultragrade 19, Edwards Ltd., England) was placed on top of the waterline.
For movement-control of the linear drive a custom made LabView-program was used. For data acquisition from the analytical balance the software BalanceLink (Mettler-Toledo GmbH, Germany) was used.
At the beginning the piston was placed directly above the tubing segment, without touching it. Subsequently the measurement was started. At the beginning of each step the piston was moved 50µm toward the tubing segment at a rate of 50 µm/s and then stopped. After 10 s of "settling time" the weight was measured. Subsequently the next 50 µm step was performed in the same manner as described before. The test ended after the tubing was fully occluded and no further change of displaced mass was indicated by the analytical balance.
The above described experiment provides the basic characteristic between tubing compression and displaced volume. Based on this data the generation of a time-constant flow rate was evaluated.
Furthermore a prototype of a linear peristaltic pump with five pumping chambers was build Figure 2. As described in section 4 the third chamber has the triple size of the other chambers. In this prototype the peristaltic mechanism can be ether driven by a single cam shaft or by five custom made linear drives. The cam shaft is meant to be used in the final version as a low-cost and robust drive. For experimental purposes each pumping chamber is actuated by a separate linear drive, in order to test different types of tubing and different optimization strategies. The general design of the linear drives is shown in Figure 3. Briefly, they consist of a lead screw, driven by a servo motor with positioning control.
For continuous flow rate measuring a coriolis flow meter (mini Cori Flow 13, Bronkhorst Cori-Tech BV, Netherlands) was connected to the outlet of the prototype.
A specifically developed LabView-program was used to control the five linear drives and to visualize the measured flow rate. To control the compression rate of the drives the program uses 90 sampling points that can be adjusted in 1 µm steps by the user in the graphical user interface (GUI).

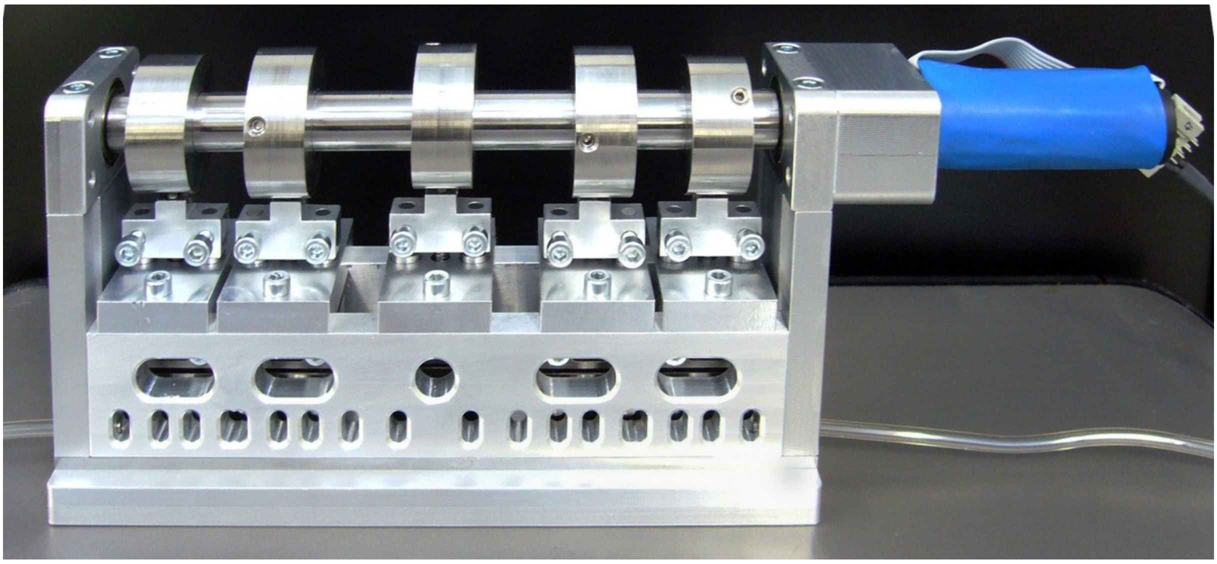

Figure 2: Prototype of the linear peristaltic pump. The cam shaft is used as a simple and robust drive. For optimization purposes the camshaft can be replaced by five linear drives.

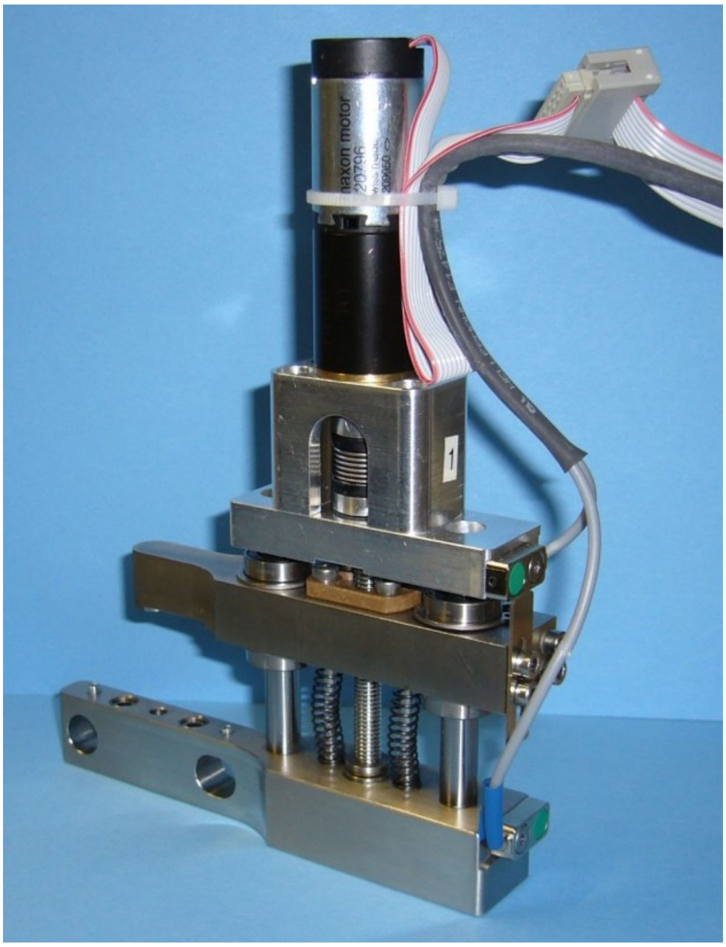

Figure 3: One of the linear drives that are used to actuate the pump. With the linear drives each chamber is controlled separately.

6. RESULTS

6.1 Tube characterization

First the single pumping chambers were characterized to derive the compression rate required for a constant flow. The compression-mass-characteristic of a custom made silicone tubing with 3 mm inner diameter is shown in Figure 4. As it can be seen, the characteristic is not linear. By inverting this characteristic to describe the compression rate as a function of the displaced mass, the flow rate can be linearized to a constant flow over time. Figure 5 shows the compression per displaced mass. The function for linearization is derived from a fitted second degree polynomial function.

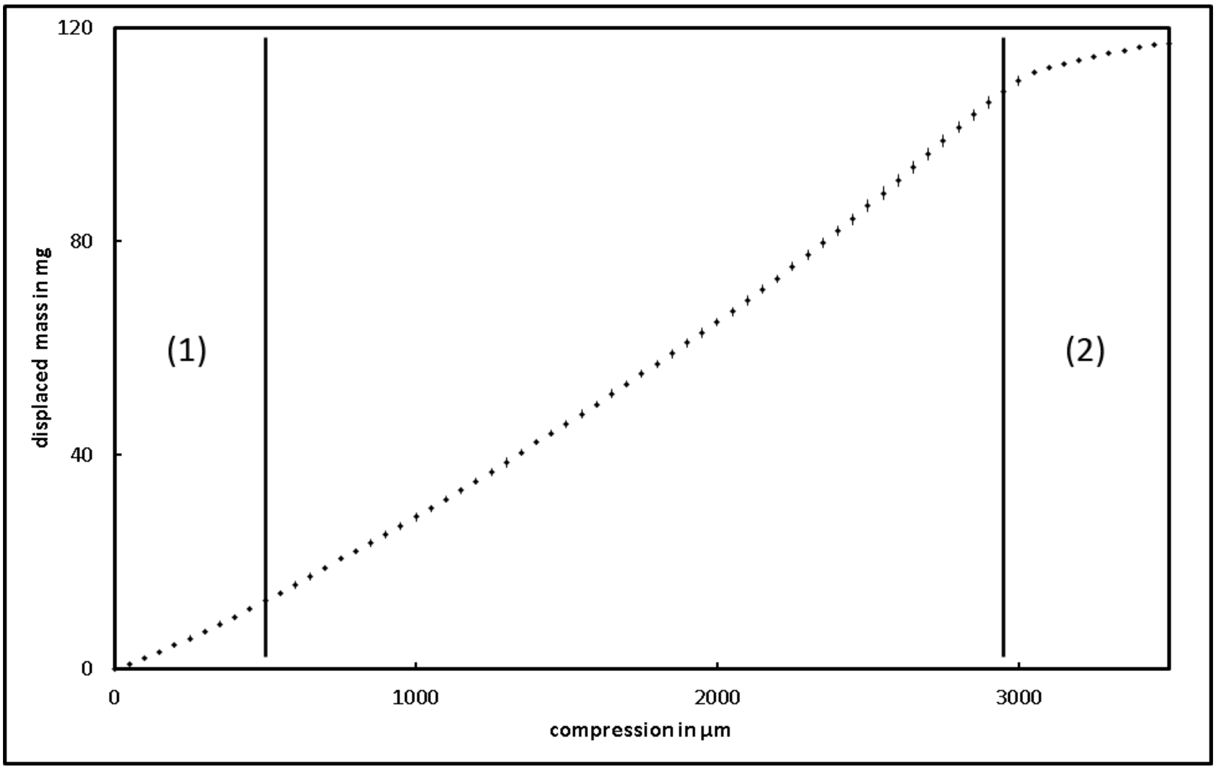

Figure 4: The displaced mass - compression characteristic is derived by stepwise compression of a tube while measuring the displaced mass with an analytical balance. Starting from an uncompressed tube at zero the tube is compressed until fully occluded. At (2) the tube is already occluded but due to the deformation upon further compression there occurs further mass displacement. For the linearization the region (1) and (2) are cut off. Region (1) is cut off because at the pump the tubing is never completely relieved to always ensure a defined condition and keep the tubing under tension when the pumping chamber is open.

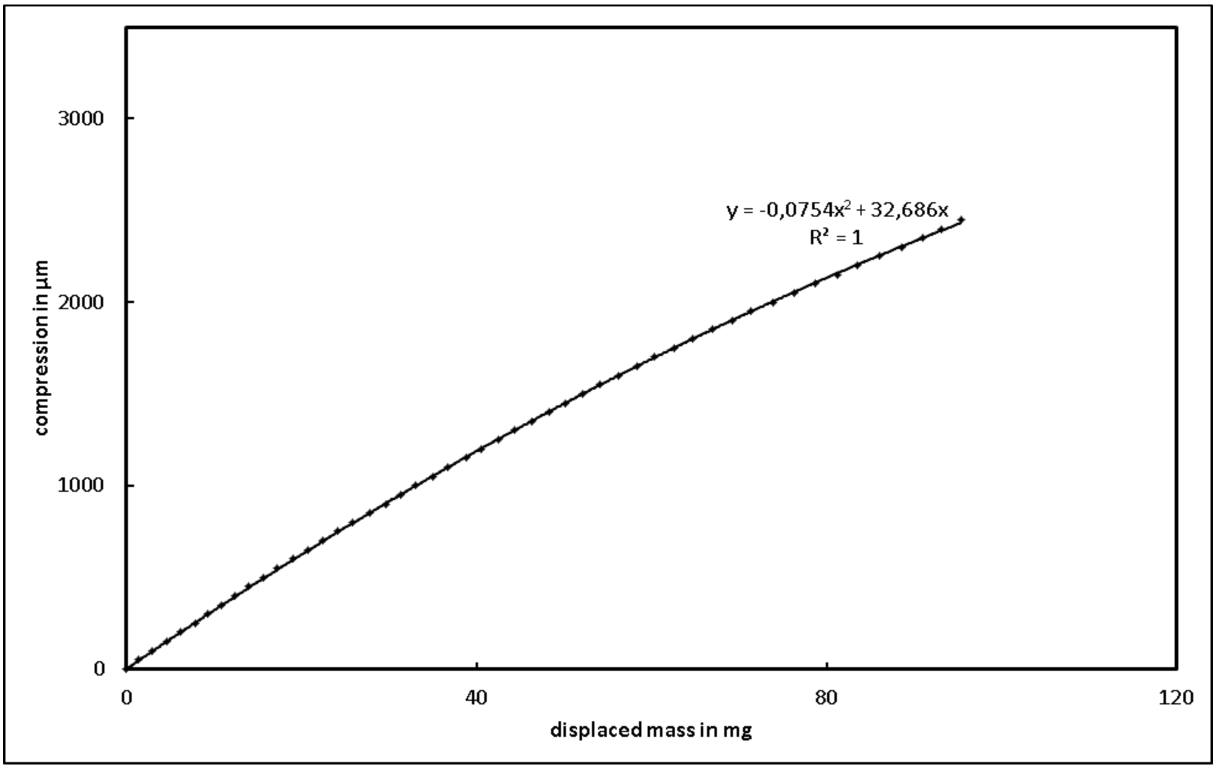

Figure 5: The compression - displaced mass characteristic is used to derive the function that describes the compression depending on the displaced mass. As function a second degree polynomial equation is fitted to the data by using the method of least squares.

Figure 6 shows the measured flow rate if applying the linearization function to the tubing via the previously described linear drive of a syringe pump. The pumping sequence is demonstrated where first fluid is charged in the chamber with a constant inflow as indicated by the lower plateau. Subsequently, in a second step the fluid is ejected with a constant outflow as indicated by the upper plateau. In the third step the pumping chamber stays closed and no volume is displaced.

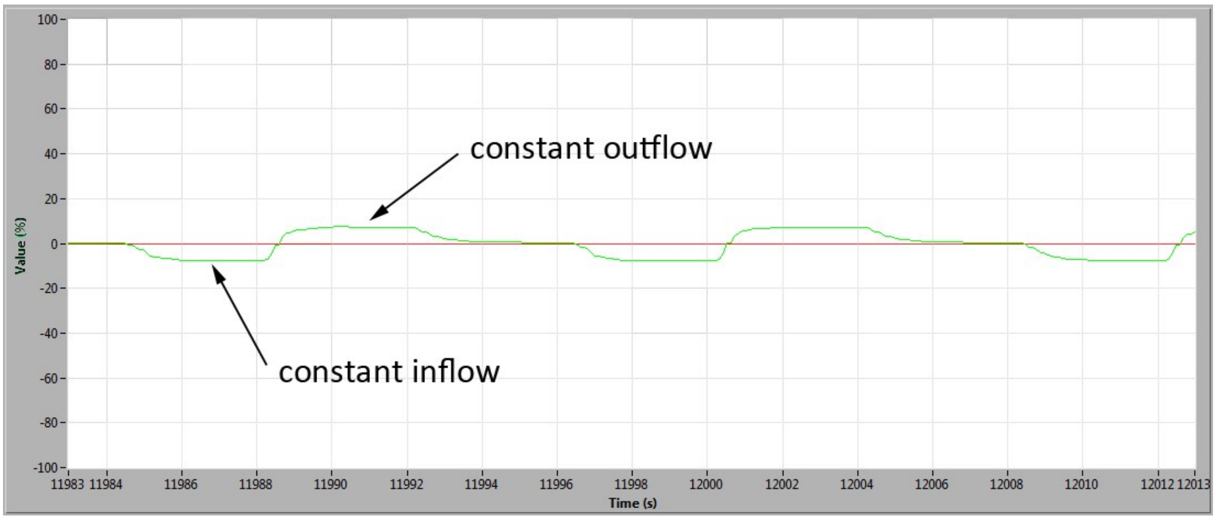

Figure 6: The flow rate over time is shown in percent of the measuring range of 1000 g/h. The lower plateaus show a constant inflow while the tubing is relieved and the upper plateaus show a constant outflow while the tubing is compressed. In the third stage, the flow rate follows the zero line since the tubing stays fully occluded for one third of the pumping cycle. The compression rate required for a constant flow was calculated from the linearization function and applied to the tubing by a linear drive.

6.2 Compression rate optimization

The linearization function derived from 6.1 was used as a starting point for the optimization of the compression characteristics of the pumping prototype. Further adjustment of the compression characteristic had to be done in an empirical process as the five pumping chambers have to work perfectly together. To control each pumping chamber separately the five linear drives were mounted on the prototype. With the linear drives it was possible to alter the compression characteristic of each pumping chamber with a step size of 1 µm.
Figure 7 shows the flow rate of an entire pumping cycle superimposed with the 90 sampling points of the compression rate control. It can be seen that there is still a fluctuation of 7% in the flow rate.

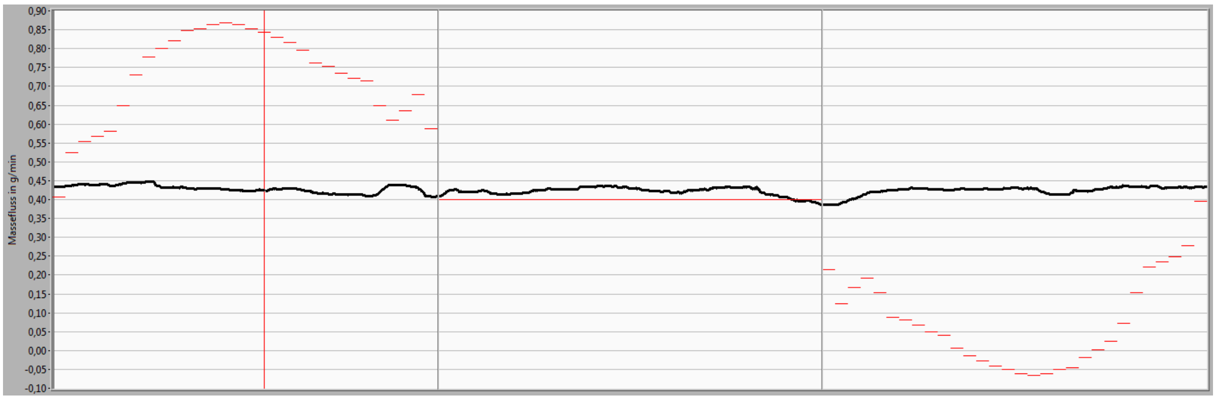

Figure 7: The pump was actuated by five linear drives. The black line shows the flow characteristic of an entire pumping cycle at 0.033 Hz. In this graph the flow is given as mass-flow in g/min. Since water was used as pumped fluid it is the same as ml/min. The flow fluctuates by ± 3 g/min around the mean value of 42 g/min corresponding to a pulsation of 7%. The red dashes indicate the sampling points of the compression rate over the three stages of the pumping cycle.

The optimized compression rates were used to design a cam shaft that applies the same movement pattern to the plungers as the linear drives. Figure 8 show the flow profile of the pump with the camshaft drive. The fluctuation of the flow rate is in the magnitude of 11%.

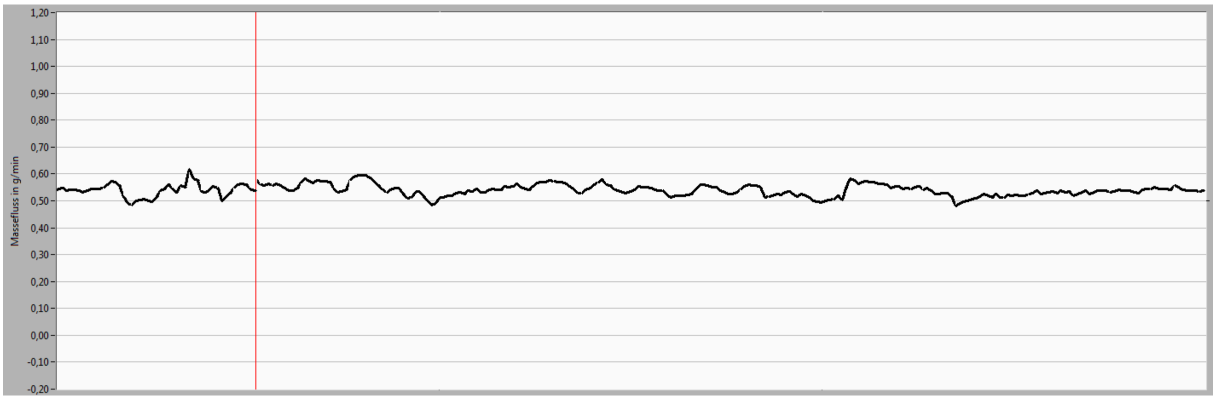

Figure 8: The flow rate is shown at a pumping frequency of 0.05Hz. The flow fluctuates by ± 6 g/min around the mean value of 55 g/min corresponding to a pulsation of 11%. The pump was actuated by a camshaft.

We have tested the prototype within a frequency range between 0.01 and 2 Hz (Figure 9). The frequency describes the rotational speed of the camshaft in revolutions per second. Our prototype covers a flow rate range from 100 µl/min to 17 ml/min. At frequencies less than 1 Hz the flow rate correlates almost linearly with the rotational speed. At higher frequencies the flow rate increase flattens.

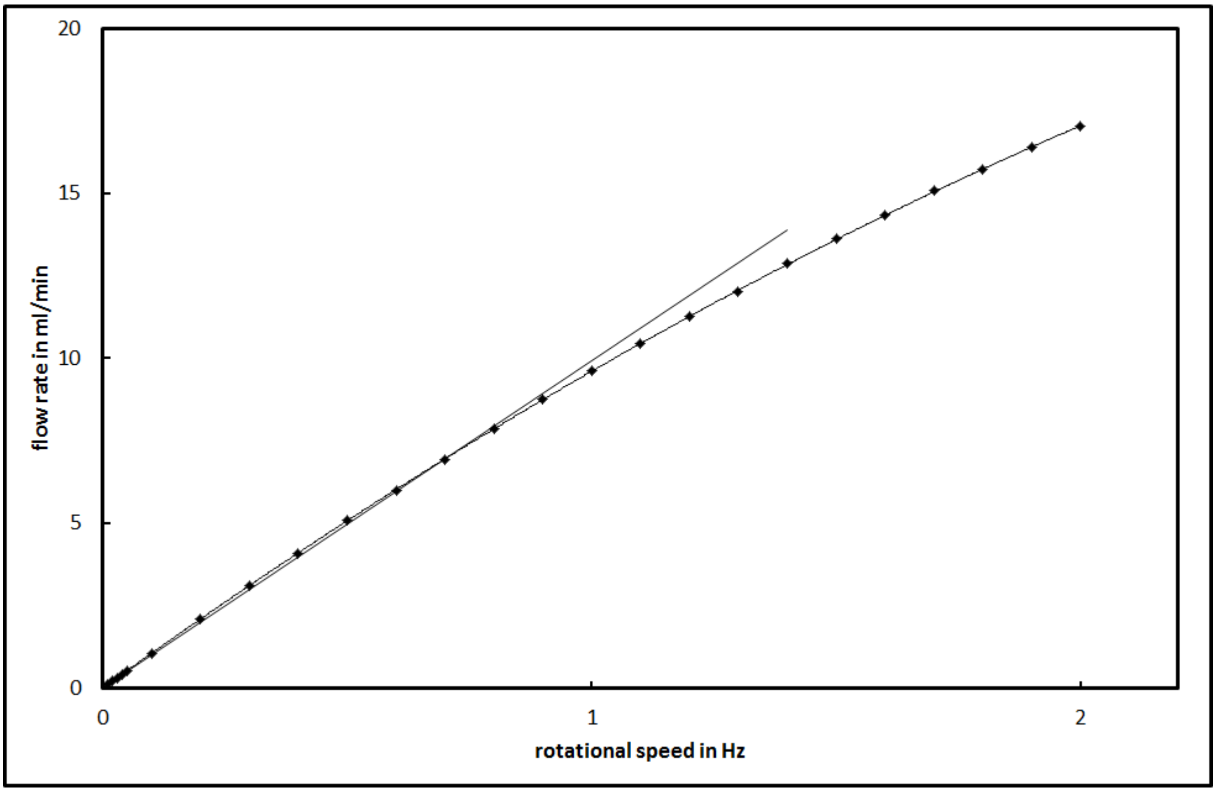

Figure 9: The flow rate is shown over the pumping frequency. Between zero and 1 Hz there is an almost linear correlation. At higher frequencies the slope flattens indicating slight disturbance by backpressure.

7. DISCUSSION

Our findings show that it is possible to build an almost pulsation-free linear peristaltic pump. However, our prototype is not perfectly pulsation-free for several reasons.
First the optimization of the compression rates did not result in a perfect pulsation-free flow due to the applied control strategy in the LabView program. Since the controllers of the linear drives can only apply constant velocities but no complex velocity profiles, the compression rate profile is generated by a sequence of commands. Hence the required speed between two sampling points is interpolated linearly and at each sampling point a new set speed is send to the controller. The resulting error would be negligible. However, the actual error is caused by latencies of the non-real-time operating system Windows. Thus, the time interval between the set speed commands fluctuates randomly. As result the linear drives are not fully synchronized on the one hand and the applied compression rate profile of each linear drive fluctuates slightly in a statistical manner as well. Therefore the generated flow does have a slight fluctuation that makes it difficult to perfectly adjust the compression rate of each pumping chamber. The only way to avoid this would be to use a real-time operating system with a fixed latency period.
Furthermore, if the prototype is operated with the camshaft it shows a greater fluctuation in the flow rate as compared to the linear drives. As the optimized compression rates of the linear drives are used to design the cams, there is already a small error as described above. In addition, there are manufacturing inaccuracies as well. The manufacturing accuracy of the state of the art machining centers is in the magnitude of 5 µm. That means the compression characteristics that are mapped on the cams can deviate up to ± 5µm from the nominal shape. This contributes to the measured fluctuation as well.

The pulsation resulting from the inconstant latency periods during the compression rate optimization can be improved by either applying latency independent optimization strategies, which still has to be developed, or by using the above described method on a real-time system. However, manufacturing inaccuracies will always be an issue that prevents a perfectly pulsation-free flow. There are possibilities to manufacture mechanical parts like the camshaft with even greater accuracy. But, that will disproportionately increase the production costs. Thus, there must be always a trade-off between the pumps retail price and the achievable flow-consistency.
Finally, the major limitation of our pump has to be mentioned. As roller pumps are known to work with tubing of different diameter, our pump will work with different tubing too. But the nearly pulsation-free flow is only adjustable for one type of tubing at the time. That means each type of tubing requires an own specifically optimized camshaft to get the pump to work pulsation-free. Nevertheless, our prototype proofs that it is possible to build an inherently pulsation-free tube pump without the use of a pulsation damper or complex online control.

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