Peristaltic Pump with Continuous Flow and Programmable Flow Pulsation

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***Abstract*—In future, sensors with flexible electronics will be deployed more often in biomedical applications for the acquisition of biological signals in the human body. To ensure biocompatibility, these sensors are often encapsulated by polymers. For research and testing of new encapsulation materials and their biocompatibility evaluation, new test procedures will be necessary, which reflect the flow conditions of the physiological environment. This paper presents a new peristaltic pump which stimulates the tube by a circulating eccentric oscillation. The pump characteristics differ significantly from conventional roller pumps and combine continuous flow and programmable flow pulsation. In continuous mode the non-occlusive pump can reduce the flow pulsation by about 85% in comparison to a conventional roller pump. Programmable flow pulsation is achieved by controlling the oscillation amplitude. This control characteristic enables the generation of defined volume flow pulses. Pulse shapes of a second generation vessel in the human arterial network can be generated in experimentally, for instance. These pulse shapes are required for in vitro investigations of biocompatible encapsulations under physiological flow conditions.**

***Keywords*—peristaltic pump, in vitro circulation, biosensor, biocompatibility testing**

I. Introduction

For biomedical applications, sensors with flexible electronics will be required more frequently to detect biological signals in the human body [1]. Currently, these sensors are often encapsulated by polymers. This encapsulation ensures the biocompatibility of the sensor unit and prevents negative reactions between the human organism and the implanted unit. However, the polymers currently used for encapsulation are not capable of hermetically sealing the unit from the organism due to residual ion permeability [2]. New test procedures will be needed in the search for new encapsulation materials and their biocompatibility evaluation. These test procedures will need to reflect the flow conditions of the physiological surroundings as accurately as possible.

Artificial body fluids in in vitro circulation loops will be used in these investigations. Currently, roller pumps are often used to pump liquids through the test circuit [3]. These peristaltic pumps work according to the positive displacement principle and displace the volume peristaltically by completely occluding the pump tube. Due to the complete tube occlusion a strong volume flow pulsation occurs. This pulsation and the associated pressure peaks can negatively impact sensitive processes. For example, the pressure peaks can damage thin membranes during biocompatibility in vitro studies. Furthermore, roller pumps are not capable of generating specific pulse shapes needed to simulate physiological flow conditions. Centrifugal pumps are also used in these test circuits. These hydrodynamic pumps can provide a very constant volume flow for the test circuits, but are also incapable of generating the required defined pulse shapes.

The contribution of this paper is the introduction of a novel pumping principle based on a peristaltic pump, which can be used to generate a continuous volume flow without pulsation and programmable pulse shapes. The biocompatibility of encapsulations under physiological flow conditions can be tested with programmable pulse shapes. This is done, for example, with a typical pulse shape of a second generation vessel in the human arterial network (according to [4]).

II. Pumping Principle

In contrast to conventional roller pumps, a circulating eccentric oscillation stimulates the pump tube, which is not occluded completely during this process [5-7]. For this purpose, the tube is placed in the gap between the oscillating coupler and the housing of the pump section (see Fig. 1).

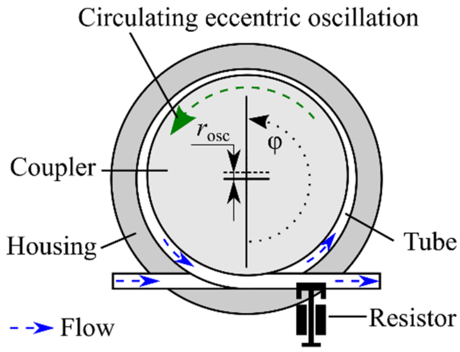

Fig. 1. Top view of the rotationally symmetrical pump section. The coupler performs a circulating eccentric oscillation with frequency f, oscillation amplitude r_{osc} and angle φ of oscillation.

A dynamic throttling device (resistor) [7] for decoupling the volume flow is fitted at the pump outlet. The pumping principle is based on a superposition of the displaced volume, the propagation of the pulse wave along the tube winding and the throttling effect of the resistor. The main ideas behind our new pumping principle are described in the following three paragraphs. Fig. 2 illustrates the pump section projected onto developed surfaces for better understanding.

A. Volume pulse

The coupler oscillates with frequency f and amplitude r_{osc} during the pumping operation. In Fig. 1 and Fig. 2, the angle φ describes the current position of the coupler during its eccentric motion. This motion deforms the tube periodically,

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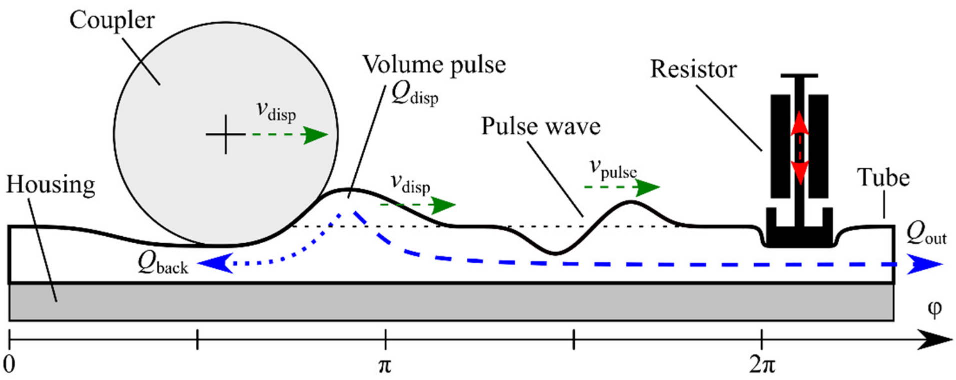

Fig. 2. Unrolled surface view of the pump section. The pumping principle of the non-occlusive peristaltic pump is based on a combination of volume displacement Q_{disp}, backflow Q_{back} and pulse wave propagation. The resistor decouples the volume flow at the pump outlet.

which displaces a part of the volume inside the tube. This generates a volume flow pulse Q_{disp}, which moves synchronously with the circulating oscillation with velocity v_{disp} (see Fig. 2). In consequence of the partial occlusion of the tube, a backflow Q_{back} occurs inside the tube through the residual gap. Hence, the flow at the pump outlet Q_{out} is formed by the superposition of Q_{disp} and Q_{back}.

B. *Pulse wave*

Because of the rotationally symmetrical pump section design (see Fig. 1), a pulse wave propagates within the pump tube in addition to the volume pulse. The pulse wave propagates at velocity v_{pulse} (see Fig. 2). This pace corresponds to the specific propagation velocity at which pressure waves preferentially propagate within the tube. Therefore, the pulse wave velocity is a kind of natural frequency of the tubing system.

C. *Resistor*

A dynamic throttling device [7] is placed at the pump outlet to generate the required unidirectional volume flow. For this purpose, a dynamic flow resistor is installed at the end of the tube winding. It acts passively on the tube and changes the flow resistance depending on the pressure in the tube (resistor motion see red arrow in Fig. 2). The resistor and the tube winding form an oscillator circuit which is excited by the pulse wave. The pulse wave is ahead of the volume pulse for $v_{pulse} > v_{disp}$. Therefore, the pulse wave deflects the resistor before the volume pulse reaches the pump outlet. The resistor deflection widens the cross-sectional area of the tube and the lagging volume pulse can pass through the resistor with low losses. The resistor thus smooths the flow Q_{out} and reduces the flow pulsation of the pump significantly.

III. System Components and Methods

A. *Pump actuator*

The pump is operated by an oscillating armature motor. Fig. 3 illustrates the pump actuator in longitudinal section (according to [5, 7, 8]). The oscillating armature motor consists of an electromagnetic transducer with radially arranged coils and an oscillating armature in the middle. A phase-shifted coil current excites the oscillating armature. This provides the required eccentric oscillation. Finally, the bending rod transmits the oscillation to the coupler and stimulates the pumping principle as described.

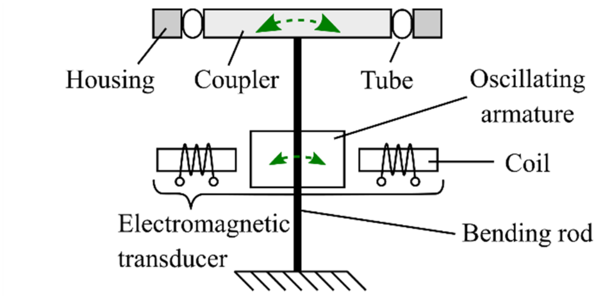

Fig. 3. Schematic illustration of the pump actuator in longitudinal section view (according to [5, 7, 8]). An oscillating armature motor generates the circulating eccentric oscillation (green arrows). The bending rod transmits the motion of the oscillating armature to the coupler.

B. *Experimental pump*

Fig. 4 shows a detailed view of the pump section with the components described above. A silicone tube (type: ECC-SIK, from Raumedic) [9] with an inner diameter $d_{tube,i}$ = 3/16" (≈ 4.8 mm) and a wall thickness w_{tube} = 1/16" (≈ 1.6 mm) is used in the experimental pump setup. The coupler has a diameter of 40.8 mm and the gap between the coupler and the housing is 6.2 mm wide. The pump is operated at an

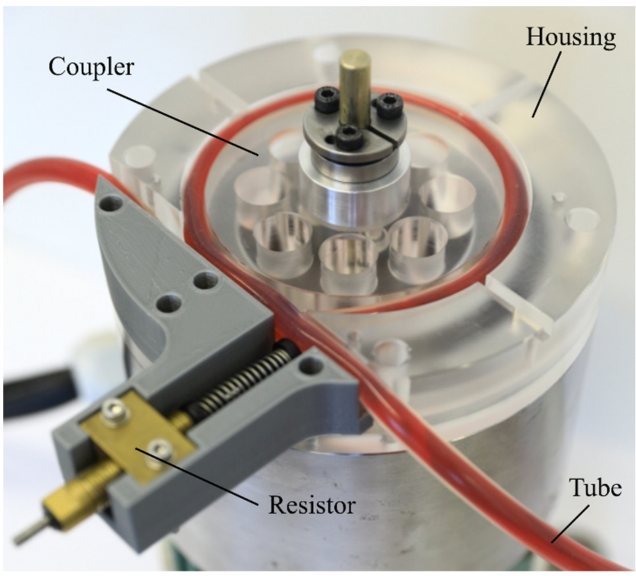

Fig. 4. Detailed view of the pump section of the experimental pump. A silicone tube is used in the experiments.

oscillation frequency $f = 100$ Hz in the following experiments. The pump oscillation amplitude r_{osc} is controlled with different signals to implement the different operating modes. In continuous volume flow mode, the pump operates with a constant oscillation amplitude (r_{osc} = const.). In order to generate the programmable pulses, the pulse shape of the desired flow pulse is modulated onto the signal ($r_{osc} \neq$ const.).

C. *Test circuit*

Fig. 5 shows the schematic illustration of the test circuit used for the experiments. It represents a typical test bench for researching and testing encapsulation materials and for their biocompatibility assessment. The test circuit consists of the pump, a flow sensor, a pressure sensor, an adjustable throttle valve and a reservoir for the process liquid. In future, a test chamber can be added to the circuit for material and biocompatibility testing. The hydraulic load of the test chamber is simulated in the experiments by the adjustable throttle valve, which replaces the test chamber in the circuit.

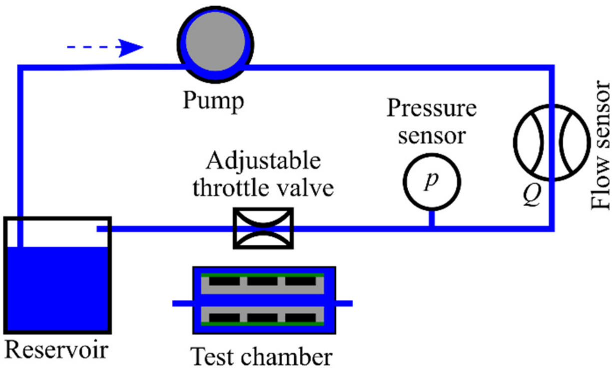

Fig. 5. Schematic illustration of the test circuit used for the experiments. The test chamber can be added for material characterization tests.

A dynamic non-invasive flow sensor (type: CO.55/060 V2.0, from Sonotec) [10] is used to acquire the transient flow characteristic of the volume flow rate Q generated by the pump. During the experiments the back pressure p is set by the adjustable throttle valve. For the test circuit, a silicone tube (type: ECC-SIK, from Raumedic) [9] with a length of 2 m is used and water (H_20) is selected as process liquid.

IV. Results

A. *Continuous Flow*

The continuous flow characteristic of the non-occlusive pump is compared with that of a conventional roller pump (type: MCP Process and Pro-380, from Ismatec) in the first experiment. For this purpose, the transient volume flow rate Q of the two different pumps is measured under identical conditions within the test circuit. The non-occlusive pump operates in continuous volume flow mode (r_{osc} = const.). The roller pump works with the same silicone tube (type: ECC-SIK, from Raumedic) [9] to avoid skewing the results at a speed of 82.3 rpm.

As an example, an average flow rate $Q \approx 300$ ml/min is selected for this experiment. The back pressure is set to zero ($p = 0$) to compare the pump characteristics without hydraulic load. The flow sensor captures the flow rate Q with a sampling interval of 50 ms for the conventional roller pump and 40 ms for the non-occlusive pump. Fig. 6 shows the measured volume flow of a conventional roller pump (red graph) and the non-occlusive pump in continuous flow mode (blue graph). According to [11] the pulsation δ_Q of the volume flow rate Q is defined by the maximum volume flow Q_{max}, the minimum volume flow Q_{min} and the average volume flow Q_m:

$$\delta_Q = 100\% \cdot (Q_{max} - Q_{min}) / Q_m \tag{1}$$

When looking at the measured volume flows in Fig. 6, a clear difference between the two pumps can be seen in the volume flow pulsation. Table I shows the measured values Q_{max}, Q_{min}, Q_m and the pulsation δ_Q for both pumps.

TABLE I. Measured values and the calculated pulsation

	Q_{max}	Q_{min}	Q_m	δ_Q
Non-occlusive pump	346 ml/min	261 ml/min	303 ml/min	28.1%
Conventional roller pump	676 ml/min	54 ml/min	337 ml/min	184.6%

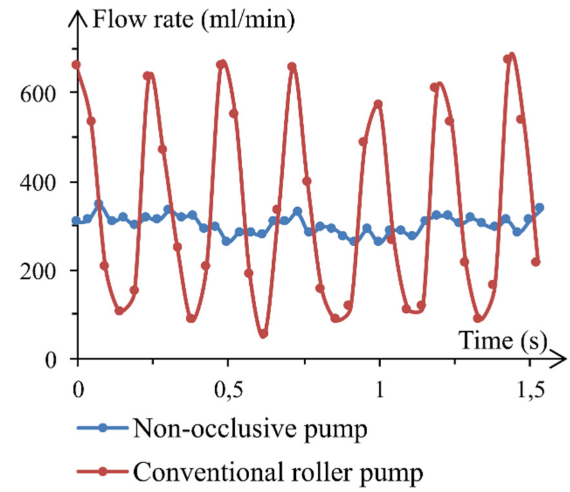

Fig. 6. Measured volume flow of the non-occlusive pump (continuous mode) in comparison to a conventional roller pump.

In contrast to the conventional peristaltic pump, the non-occlusive pump generates a significantly more uniform volume flow in continuous mode. The pulsation δ_Q (according to (1)) can be reduced by about 85% when using the non-occlusive pump in continuous mode instead of a conventional roller pump. This effect is based on the back flow within the tube and the resistor operation of the non-occlusive pump. Both these features smooth the volume flow pulsation. The strong pulsation of the conventional roller pump is due to the positive displacement during pumping. A defined volume is ejected at the pump outlet every pump revolution due to the complete tube occlusion. The discontinuous volume flow is caused by the rollers, which occlude the tube completely and interrupt the flow. The non-occlusive peristaltic pump reduces the volume pulsation and the associated pressure peaks significantly. Damage to membranes is thus avoided during membrane biocompatibility research and testing.

B. *Pulsatile Flow*

The non-occlusive pump can generate specific volume flow pulse shapes by modulating the oscillation amplitude r_{osc} during pump operation. As an example, three different pulse shapes are generated in the second experiment. This is

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achieved by modulating the oscillation amplitude r_{osc} with the shape of the desired flow pulse. The back pressure is set to 15 kPa at an average flow rate Q = 150 ml/min to simulate the hydraulic load of the test chamber. Fig. 7 shows two exemplary pulse shapes. For the green graph, the oscillation amplitude r_{osc} is modulated with a sinusoidal control signal. A sawtooth signal is used for the blue pulse shape.

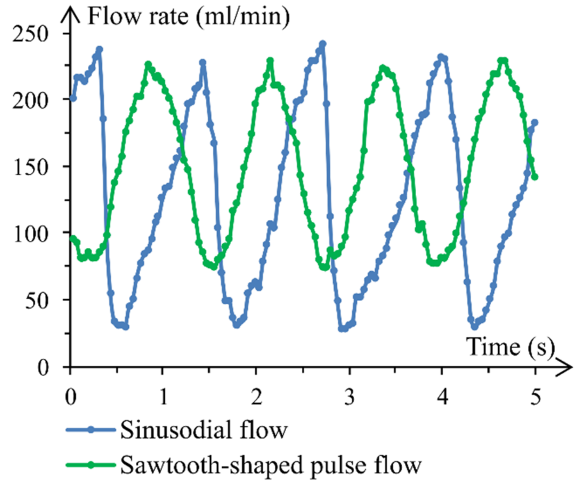

Fig. 7. Examples of generated volume flow pulse shapes. Multiple pulse forms can be realized with the non-occlusive pump by modulating the oscillation amplitude r_{osc}.

Generating special-purpose volume flow pulse shapes is of particular interest for in vitro investigations. By modulating the oscillation amplitude r_{osc} during pump operation defined volume pulses can be generated with the non-occlusive pump. As an example, Fig. 8 illustrates a generated flow, which is typical for second generation vessels of the human arterial network (according to [4]). This pulse shape can be used in material testing, for example, to simulate physiological flow conditions with artificial body fluids.

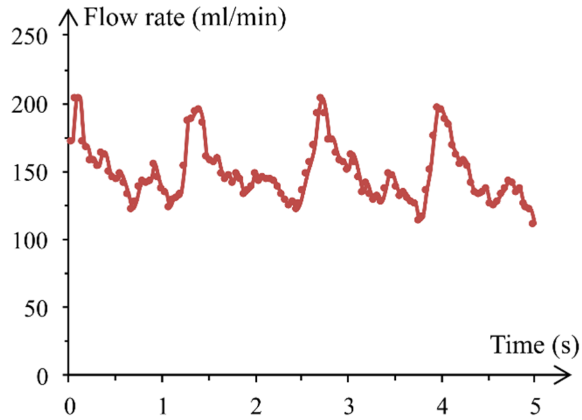

Fig. 8. Generated pulse shape with reference to the pulse shape of a second generation vessel in the human arterial network (according to [4]).

The experiments performed in this study show that the non-occlusive pump can generate multiple flow pulses. However, the slope of the generated pulse shape is limited by the inertia of the flowing medium and the non-occlusive pumping principle itself. This limitation is especially visible at the falling edges of the sawtooth-shaped pulse (blue graph in Fig. 7) and the rising edges of the generated second generation vessel flow (Fig. 8).

V. Conclusion

Biosensors are set to become more commonplace for biomedical applications in the human body. Biocompatibility investigations are carried out on new encapsulation materials as part of research and development programs. Test circuits with roller pumps are typically used in these investigations. However, conventional roller pumps can only generate a pulsating volume flow and this type of flow can impact sensitive processes such as in vitro studies on the biocompatibility of thin membranes. Furthermore, these types of pumps cannot generate custom pulse shapes that imitate physiological flow conditions in the human body.

To overcome these challenges, we have introduced a new pumping principle which operates without complete tube occlusion by stimulating the tube with a circulating eccentric oscillation. In our experiments volume flow transients are characterized. The non-occlusive pump reduces the volume flow pulsation δ_Q by about 85% over a conventional roller pump under identical test conditions. Furthermore, the non-occlusive pump can generate programmable pulse shapes, for example, a typical pulse shape of second generation vessels in the human arterial network.

Summarized, the non-occlusive pumping approach allows to generate both a continuous volume flow without pulsation and a programmable flow pulsation. In addition, it combines the advantages of the conventional roller pump, such as low operating costs and no cross-contamination, with a much simpler design. This makes this non-occlusive pump a very attractive tool for materials research and testing under in vitro conditions.

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