Investigating Wave Propagation in the Lower Urinary Tract to Improve Urinary Bladder Emptying

Angelo Locatelli

Supervisor: PD. Dr. Francesco Clavica
Institution: ARTORG Center for Biomedical Engineering Research, Universität Bern
Examiners: PD. Dr. Francesco Clavica, Prof. Dr. Dominik Obrist

Introduction

Impedance pump is an innovative technique that can light up the future of urology providing a solution for what concern bladder underactivity or bladder outlet obstruction. It's principle consists in generating net flow by externally and asymmetrically compressing a compliant vessel. This creates travelling waves which are reflected at impedance mismatch points [1]. However, it's functioning is highly sensitive, among other parameters, to frequency and position of stimulation. The aim of this project is to assess how this two quantities influence pumping efficiency, direction and associated reproducibility of results.

Materials and Methods

A total of 327 experiments about wave speed and flow rate were performed on two urethral models: one Polydimethylsiloxane (PDMS) penis model (L=167mm, ø=8mm) and one penrose/silicone tube (L=167mm, ø=16mm). For wave speed evaluation, single pressure wave fronts were generated and recorded through pressure sensors. Once recorded, foot locations of the compression wave were extracted via MATLAB (Mathworks Inc., USA) in-house scripts. Delay of detection and distance between sensors were used to provide wave speed values. A linear motor (lm2070-80-01, FAULHABER, Germany) was used to compress the tested urethra models, according to several compression frequencies and positions along urethra models, Three repetitions of each test were conducted. Afterwards, video analyses based on variation of the water level in the reservoir were carried out to obtain the flow profile.

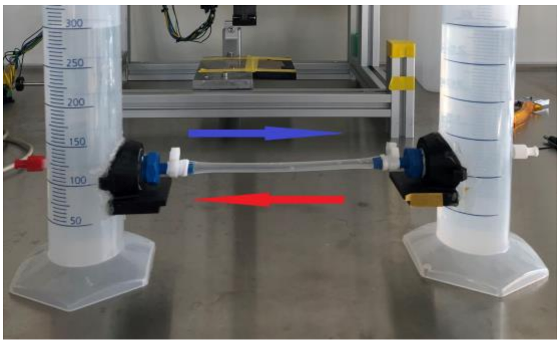

Fig. 1 System used with penrose tube to assess flow rate. It is composed from two plastic reservoirs, and connectors on which the silicone tube is mounted.

Outcomes included maximum flow rate, variation of water height and respective standard deviations given by mean of the three repetitions.

Results

Compared to the PDMS, the penrose model exhibited better results in term of reproducibility, symmetry and pumping performance, leading to a maximum flow rate of 3.2mL/ with a standard deviation of 0.1mL/s. At 69.5mm in Fig.2, no net flow was detected for all tested frequencies. The most efficient configurations turns out to be with actuator at 21Hz placed between 44-46mm or at 30Hz with an axial compression made at 12-17mm from the mismatch point. Reverse flow (red arrow in Fig.1) is observed and the most intense, equal to -0.6mL/s ± 0.02mL/s, stands at 40mm and 50mm with 17Hz. Wave speed for PDMS urethra was 3.59m/s ± 0.26m/s, while for the penrose was 2.24m/s ± 0.06m/s.

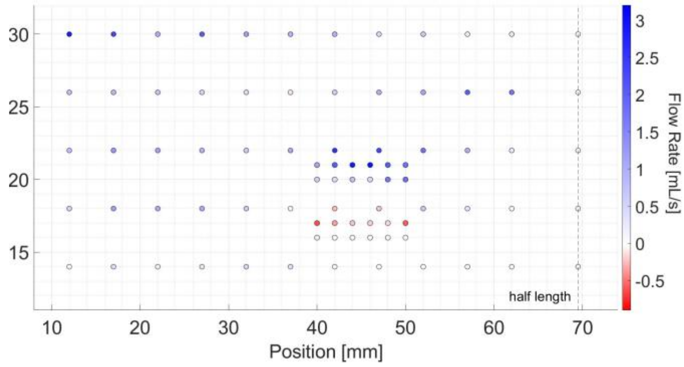

Fig. 2 Flow rate color-mapping for penrose model as function of frequency and position of compression. Each value is the result of the average of the three repetiotions.

Discussion

Results of the study are significant because they can be used to improve performances of a valveless pumping device, currently under investigation, to empty bladders.

References

D. Rinderknecht and G. Morteza. Development of a microimpedance pump for pulsatile flow transport-Part 1: Flow characteristics of the microimpedance pump. Part 2: A systematic study of steady and pulsatile transport in microscale cavities. PhD thesis, California Institute of Technology,Jan.

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