Evaluation of a Novel Portable Micro-pump and Infusion System for Drug Delivery

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***Abstract*— In this paper the design, fabrication and experimental results of a novel portable fixed-displacement micro-pump for controlled dosing and timing is described. The new pump is developed especially for high efficiency, high accuracy, ease of use and very low cost for single use drug delivery systems which can overcome many of the deficiencies of current portable pumps. Primary tests have been conducted and the results have demonstrated that the pump has the ability to deliver high performance and accuracy with less than +/- 1% error over the whole operating flow rate range of 0.05-120 (mL/h). The pump is designed to be used with a motor drive, which has been configured to be the size of a typical pen, improving the patient's mobility and wellbeing. The new micro-pump can be used for a variety of applications including chemotherapy, insulin delivery, pain management and antibiotic therapy. A complete therapy system is enabled by providing physicians with devices that programme the Pen-drive for patient specific therapies.**

I. INTRODUCTION

Infusion pumps were first introduced approximately 30 years ago. Infusion pumps are medical devices that are used to deliver therapeutic drugs and fluids such as plasma, dextrose and saline solutions, into a patient's body in a controlled manner and at a precisely controlled rate. These pumps may be used during treatment such as diabetes, chemotherapy and also in pain management. Disposable infusion pumps provide patients with advantages such as light weight, small size, simplicity of use, independence from an external power supply etc. [1]. The use of infusion pumps has been dramatically increasing, and these pumps have found their way into various areas of health care to deliver maintenance fluids as well as blood transfusions and also total parenteral nutrition (TPN).

Micro-pumps are able to transport and dispense therapeutic agents in small quantities at flow rates of micro/milliliters per minute. Laser and Santiago categorise micro-pumps in a review article into two main categories: displacement and dynamic [2]. Displacement pumps exert pressure forces on the fluid through moving boundaries while dynamic pumps constantly inject energy directly into the fluid to increase either its momentum or pressure. The majority of micro-pumps are of displacement type in which the diaphragm acts as a moving surface in a periodic manner. The manufacture of the most common micro-pump is achieved by using a miniature diaphragm in the pump; an elastomeric component to transfer fluid easily from the inlet to the outlet. As the diaphragm moves the volume of the chamber increases and decreases, producing a force which draws fluid through the inlet to the outlet [3]. This device can control and deliver fluid flows in the micro/milliliter levels. Advantages of diaphragm micro-pumps include an oil free operation, low leakage and high reliability [4]. Disadvantages of diaphragm micro-pumps are their sensitivity to proximal and distal pressure and the need for valves, both to control fluid entering the diaphragm cavity and to prevent free-flow of fluid through the pump.

An ideal infusion pump is reliable, simple to use, portable, miniature, inexpensive and light weight; this makes it easy to use and carry around for patients in both hospital and homecare settings. Current drug delivery devices do not provide all of these capabilities in one device. Larger infusion systems may limit mobility and daily activities and also are very difficult to conceal.

The new micro-pump has been developed in an effort to meet the needs of the ambulatory drug delivery market and address the associated technical hurdles. The new design offers a very low cost, miniature, single use pump with improved accuracy and facilitates the design of simple, easy to use systems necessary for modern drug treatments. The new micro-pump has being designed to meet all known requirements and replaces both peristaltic and syringe driver products.

II. MATERIALS AND METHODS

A. The Proposed Micro-pump and Associated System

The micro-pump is a rotary positive displacement pump, with only one moving part, designed for high accuracy and low cost. The micro-pump consists of a flexible diaphragm integrally moulded to a housing carrying inlet and outlet ports (Fig. 1). The diaphragm spring regulates the maximum output pressure. Thus, if a needle were to become occluded the pump will reach a threshold pressure eliminating the need for a distal pressure sensor. The occluded max pressure recorded was 12±2 psi. The diaphragm spring also acts as a valve to prevent the free-flow of fluid through the pump (e.g. as a consequence of gravity or pressure on the reservoir). The check pressure measured was 11±1 psi.

Precise metering is of great importance in drug delivery [5], and the novel design of the micro-pump allows it to deliver very small amounts of medication over time. This can be useful especially when low targeted doses have to be delivered to patients. High accuracy and repeatability is provided by the injection moulding processes, and, since the pump component is designed to be disposed after use, accuracy is assured without expensive maintenance or complex calibration regimes.

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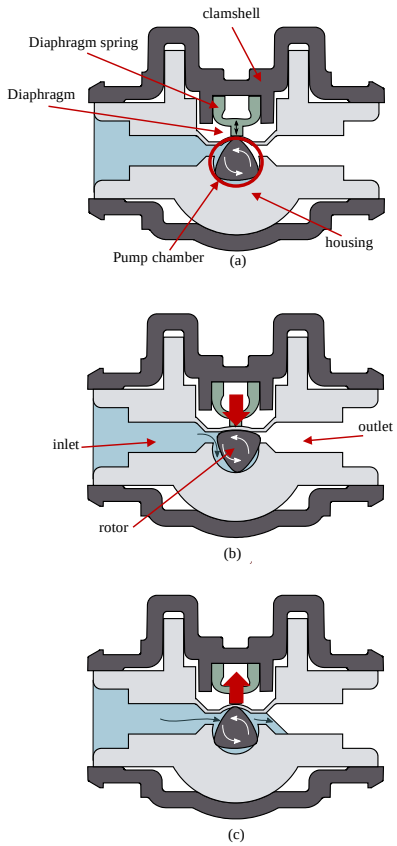

Figure 1. Cross section of the structure and the operation of a typical displacement micropump: (a) pump is not running and the diaphragm prevents fluid passing from the inlet to the outlet, (b) motor starts running drawing fluid in through the inlet and anti-clockwise around the pump, (c) the diaphragm expels the fluid through the outlet.

The pump is driven by a very simple motor drive, thus simplifying the design of the pump controller. As the pump is fixed displacement the flow rate is a function of motor shaft speed and dose volume is a function of time. A primary advantage of the new design is low torque, meaning that it can be driven with a compact motor with very low power consumption, which then results in the miniaturisation of the whole system. The delivery system, which looks like a typical writing pen, can be placed in a pocket (Fig. 2). The possibility to perform daily activities and the freedom of movement that is afforded with the new design will have favourable effect on the quality of life of patients receiving the medication for short or long periods of time. As patients do not need to carry a heavy bag or system around it's very easy to conceal enabling patients to enjoy life to the fullest. The miniature size also allows for less storage space in pharmacies and hospitals and the low cost reduces investment in inventory. The pump typically forms part of the disposable pack containing the drug on the distal side and a giving set comprising un-calibrated tubing with cannula or needle on the proximal side. A re-usable, programmable Pen-drive carries the drug delivery regime and drives the pump. An advantage of this system is the physician provides the Pen-drive to the patient that is specific to their treatment. The patient can use the Pen-drive to dispense many disposable packs. At the end of treatment, the Pen-drive is of such low cost it can be considered disposable. This means that there is no requirement for any equipment maintenance, calibration and inventory management. There is no black market value for the Pen-drive as it may be programmed to shut down after the therapy has been completed.

The pump is designed especially to be patient-friendly, it can be used without unintended patient intervention, which enables patients to use the medications safely, comfortably and conveniently at home. Also since the system is battery powered there is no need for a power cord. Flow rate can be modified by the doctor by re-programming the EPROM using the base system. This is accomplished simply by placing the Pen on the base system. It is envisaged that the base system may simply be an attachment to a smart-phone. Moreover, the doctor can read the activity of the Pen-drive to improve subsequent therapies. The micro-pump has been designed to be used in a wide variety of applications for constant-rate or programmed delivery of drug infusion.

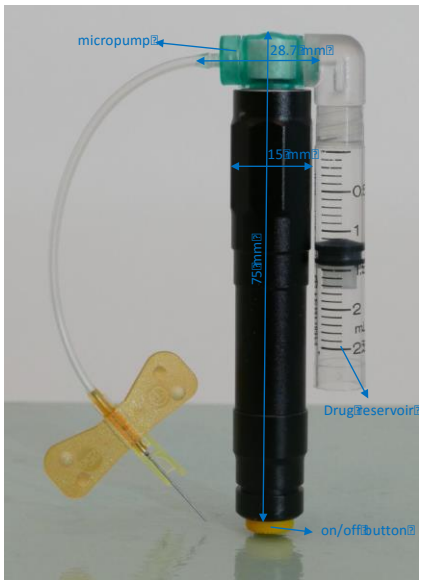

Figure 2. The disposable reservoir is attached to the pump and needle.

Inside the Pen there is a EPROM microchip, which can be programmed by the physician using the base system, which may be linked to the patient's records. The drug reservoir may be a vial attached to the pump as seen in Fig 2. The Pen is powered by a standard alkaline AAA battery which the patient can replace from time to time.

The size (29.8 × 14.6 × 14.0 mm) and weight of the new micro-pump and the simplicity of design (Fig.3), leads to a reduction in cost for the delivery system. Various medically approved materials may be used for the fluid contacting components of the pump. The number of cavities (boluses) on the rotor should be taken into consideration when dispensing an exact amount of fluid. A typical rotor has two boluses although higher resolution can be achieved by using more than two boluses. The test results that follow are for a three-bolus rotor.

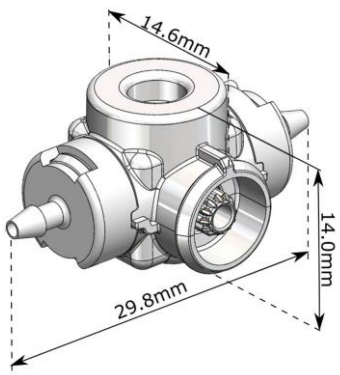

Figure 3. A three dimension picture of the micro-pump.

The pump chamber is 3 × 0.84 µl. The maximum measured volumetric flow rate, Q_{max}, is 9 mL/h at 1 rps (revolution per second) and 122 mL/h at 14 rps. The standard deviation (±SD) of the pump's flow rate over different speeds is SD≤ ±0.01. The pump can deliver continuous infusion rates ranging from 0.05 to 120 mL/h.

B. *Method of Testing*

To confirm pump performance will exceed the primary requirement specification several tests have been performed on the micro-pump including; flow rate at different rotor speeds, volumetric efficiency with water against partial occlusion in the outlet, volumetric efficiency with water at negative inlet pressures, volumetric efficiency over dynamic range with different viscosities (2 cP -1000 cP) and torque over dynamic range with water and solutions with different viscosities. The test unit consists of a micro-pump connecting to a high resolution stepper motor and torque rig. Calibration was first performed with water with zero delta pressure across the pump to quantify the flow rate and considered to be the Baseline. The motor was programmed to run at 1 rps for 100 revolutions when dispensing water. The test was then repeated at various speeds starting from 0.1 rps to 14 rps, the average dose at each speed was measured.

Six sets of testing were performed with various viscosities of glycerol solutions starting from 2 cP to 1000 cP to obtain the volumetric efficiency versus rotor speed and viscosity. The average dose was measured at each speed (0.1 rps -14 rps). The values for torque were also measured. Volumetric efficiency is defined as the volume of fluid dispensed with a given set of factors (rotor speed, fluid viscosity, pressure delta across the pump, longevity etc.) divided by the Baseline and expressed as a percentage.

The longevity test was performed to determine if the volumetric efficiency changes over extended operation, also to determine if there is any wear to the housing or rotor. The test was performed for a minimum of 2.5 hours. A LabVIEW (National Instruments Inc. Austin, TX, USA) virtual instrument (VI) was implemented to display the weight and torque on the notebook computer. The weight and torque values were recorded using a 16-bit data acquisition card (USB-6210, National Instruments, TX, USA) at a sampling frequency of 1 Hz.

Trumpet curves were calculated using Infuscale to show the accuracy performance of the micro-pump at set intervals in the first hour and during the final hour of the infusion.

III. Results

The pump was driven at various rotor speeds and the average dose at each speed was measured. The relationship of flow vs. rotor speed with water can be seen in Fig. 4.

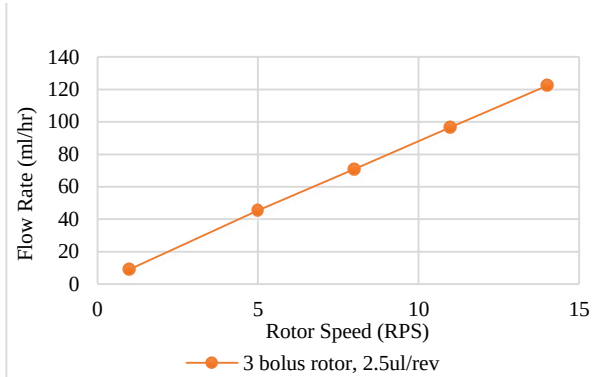

Figure 4. The relationship of flow vs. rotor speed with water.

Fig. 5 illustrates the volumetric efficiency and effect on torque with water against partial occlusion (back pressure).

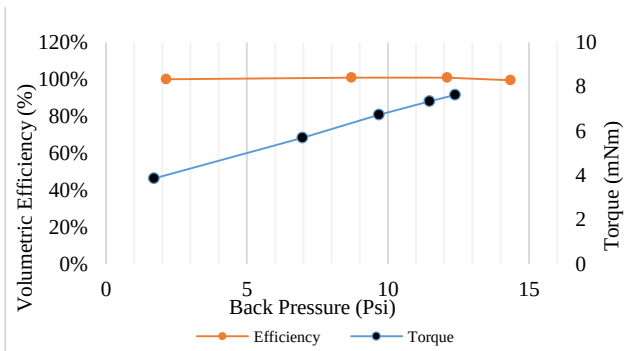

Figure 5. The volumetric efficiency and torque with water against partial occlusion.

Fig. 6 shows the volumetric efficiency with water at negative inlet pressures.

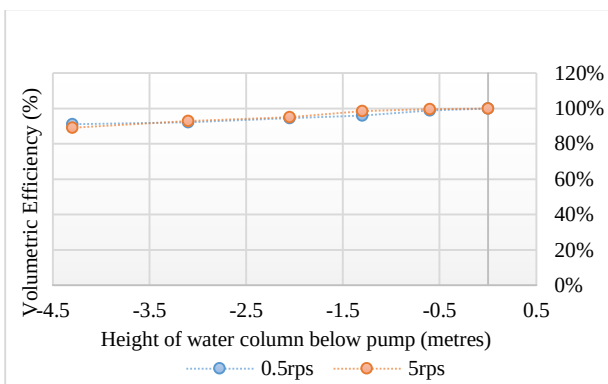

Figure 6. The volumetric efficiency with water at negative inlet pressures.

This test was performed to determine the sensitivity of a partial occlusion in the inlet (e.g. the height of a fluid

reservoir relative to the pump). Fig. 7 shows the results for volumetric efficiency over dynamic range with different viscosities starting from 2 cP to 1000 cP. The results for torque over dynamic range with water can be seen in Fig. 8. The results for the trumpet curves can be seen in Fig. 9 and 10.

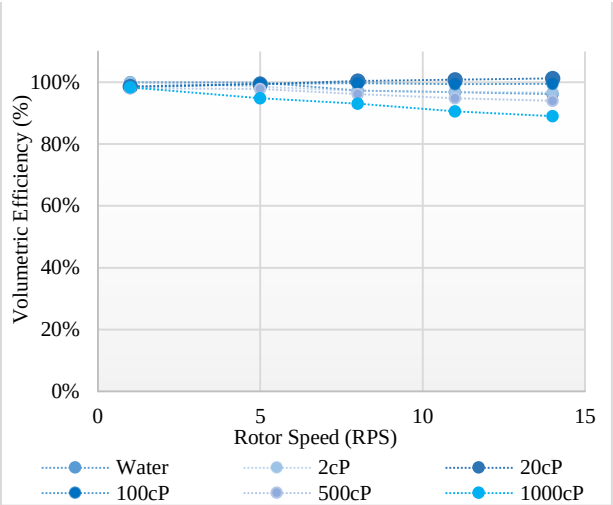

Figure 7. Volumetric efficiency over dynamic range with different viscosities.

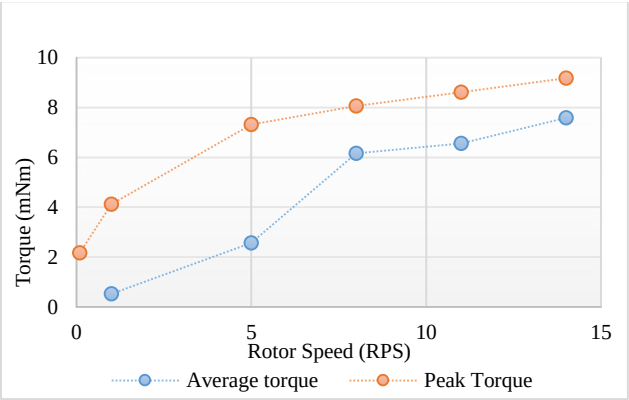

Figure 8. Torque over dynamic range with Water.

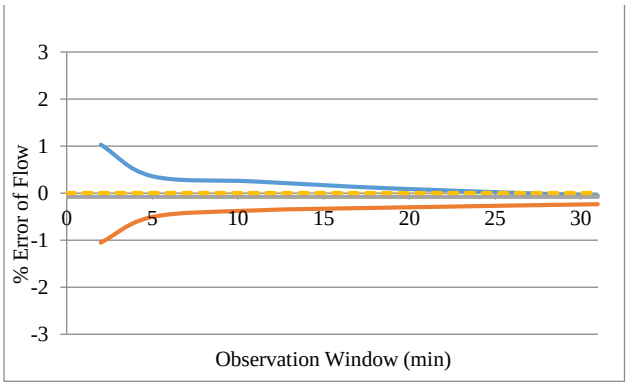

Figure 9. Trumpet curve during first hour of test period after 15m start-up period.

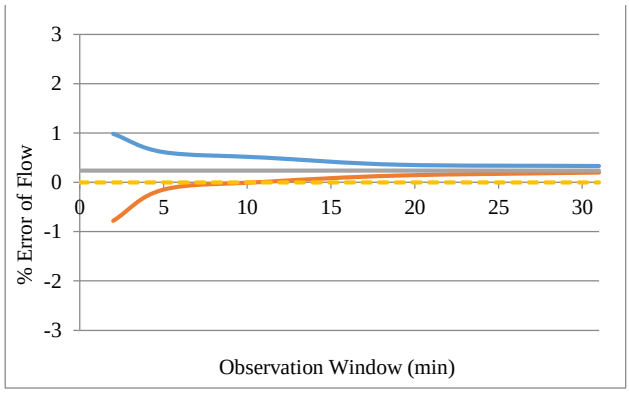

Figure 10. Trumpet curve during last hour of test period.

IV. Discussion and Conclusions

An accurate, miniature, low cost and single use micro-pump has been designed, manufactured and tested for providing controlled dosing and timing in drug delivery. Improvement of the pump performance and accuracy compared to the currently used infusion pumps was demonstrated via in vitro testing. The results for the longevity test show that the volumetric accuracy for a large dose size is consistent with small dose size. The results for longevity test revealed no evidence of wear or the formation of particulates in the fluid path. Torque reduces at the higher rps at the higher viscosities. This may be due to: glycerol is a better lubricant than water and also the pump is seeing higher internal pressures that may be reducing the interference fit between rotor and housing. The results demonstrated that the new design of the pump allows patients to receive therapy in the comfort of their own home, in a very cost effective way, which is easy to conceal. The pump is reliable, very inexpensive, and a simple construction that is easy to manufacture. It is next intended to conduct clinical trials.

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