System-on-a-Chip (*SoC*) Model of a Micropump

A M. Hodge-Miller and R.W. Newcomb
University of Maryland College Park
Department of Electrical Engineering
2405 AV Williams Building
College Park, MD 20742, USA

***Abstract*—This paper brings System-on-a-Chip (SoC) design principles to the field of micro- electro- mechanical systems (MEMS). Herein presented is a system-level approach to the development of SoC models that characterize fluid flow in BioMEMS devices. A detailed analysis of fluid flow model development is provided.**

I. INTRODUCTION

Recent IC designs have seen the integration of sets of complete yet complex devices in the development of systems-on-a-chip (SoCs). SoC designs are ameliorating the advancement of image processing, communications, bio-medical, and network systems. [1,2] Such systems typically contain many processors, ASICs and thousands to millions of lines of embedded code. Essential to today's semiconductor industry is rapid delivery of these systems. As such, early and accurate modeling of an entire system is essential for lowering the time-to-market of complex embedded SOC systems. Modeling SoCs requires the development of system-level design specifications that define both the hardware and embedded software contained within SoCs. [3]

This paper shows a set of SoC models for micro-fluidic structures used in fluid transmission for bioMEMS devices. Typically, full system modeling of micro-fluidic devices is divided into one of two methodologies. [4] The first involves the development of differential equations to describe the micro-pump's working procedure after all individual elements have been functionally construed. The second applies mechanical-electrical equivalent network representation to build system transfer functions, and then determine the micro-pump's state or process parameters by computer-aided electrical circuit analysis software such as SPICE and AMS. The models presented here differ from previous work [5,6] in that they are developed to be system modules that can be applied to SystemC Library and Matlab/Simulink environments. Moreover a comparison of the effectiveness of each of the traditional techniques, towards the development of system modules, will be made.

II. SYSTEM LEVEL DESIGN OF A MICROFLUIDIC PUMP

System-level designs typically contain sub-components showing sequential, alternative, concurrent, repletion and exception handling behaviors. [7] Such systems are specified with description languages that support these behaviors: VHDL, Verilog HDL, HardwareC, StateCharts, Esterel, and SystemC. HDLs make it easy to combine multiple IP blocks into a single simulation, where the focus is on concurrent models and their interactions.

As such, the model proposed herein is a high-level description of a micro-pump. The approach to module development (system level design) of the micro-fluidic pump is to create the full model of the pump and capture inputs and outputs between each stage of the system. The output of the systems modules are then compared against simulated outputs of electrical equivalent circuits and equivalent differential equations. System module development will augment traditional methods of developing models of micro-fluidic devices and facilitate the incorporation of micro-fluidic bioMEMs devices in SoC designs. The process of system-level modeling begins by understanding how multiple processes can be synchronized by statistically scheduling operations. To that end, we must develop an understanding of all of the degrees of freedom associated with the modeled device.

In figure 1, we present the cross section of a microfluidic pump fabricated by the AMANDA process. [8] Synthesizing the micro-pump requires that we sub-divide each of the components of the device into an equivalent block diagram as shown in figure 2. The pump shown is made up of two membrane valves and a pump thermopneumatical actuator. The membrane of the two valves and the diaphragm of the actuator are made of flexible polyimide that allows for large deflections. The large deflection provides high

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flow rates due to the fact that the output flow rate (Q) of a diaphragm micro-pump is proportional to the amplitude (ω) of the diagram deflection and is further described by:

$$Q \;\alpha\; \eta \Delta V f \;\alpha\; \eta f \omega A \qquad (1)$$

Where η is a coefficient relating to the valve leakage or efficiency, f is the actuating frequency, ΔV is the difference in volume between the pump mode and the supply mode, and is proportional to ωA, where A is the diaphragm area within the pump chamber. The technical specifications for the micro-pump are given in table 1 below.

Fluidic pumps are typically comprised of a cavity including outlets/inlets located on separate sides of the cavity. In the described micro-pump there is a flat cable extending into the cavity for contact to a heater coil that resides in the actuator chamber. In many micro-pumps there are at least two vapor permeable membranes, which are contiguous to one of the pump outlets. These membranes prevent unwanted fluids from passing through the pump outlets while allowing the target fluid to pass through the pump outlets regardless of the pump's orientation with respect to gravity. The problem we address in this work is the synthesis of the individual micro-pump components subject to the timing constraints imposed by the other concurrent models.

III. SYNCHRONIZATION OF CONCURRENT PROCESSES OF A MICROFLUIDIC PUMP

Communication in micro-pumps includes interfacing between the following systems: inlet valve, outlet valve, and pump chamber. Interfacing between these systems can include heating the actuator chamber coil, passing fluids through inlet valves into the pump chamber and then out of the valves. Each of these systems is modeled according to functionality.

In developing a system-level equivalent model of this device, we focus on the actuator chamber. The actuator chamber (AC) interacts with the inlet valve and the outlet valve. Both valves can be modeled as a fluidic diode. [9] To synthesize a valid actuator chamber, we must consider the constraints imposed by the inlet and outlet valves. The process, actuator_chamber works as follows, once the actuator chamber pressure exceeds the fluid delivery pressure, the inlet valve is shut-off and the outlet valve is opened. Opening the outlet valve allows flow out of the micro-pump. The specification of this process is shown below.

```
output […] shut_off_inlet, open_outlet
 …
  while ($Pressure_{AC}$ > $Pressure_{Inlet}$)
   begin
   pumpfluid_in = 0;
   shut_off_inlet = 1;
   if(shut_off_inlet == 1)
         open_outlet = 1;
   end
  end
```

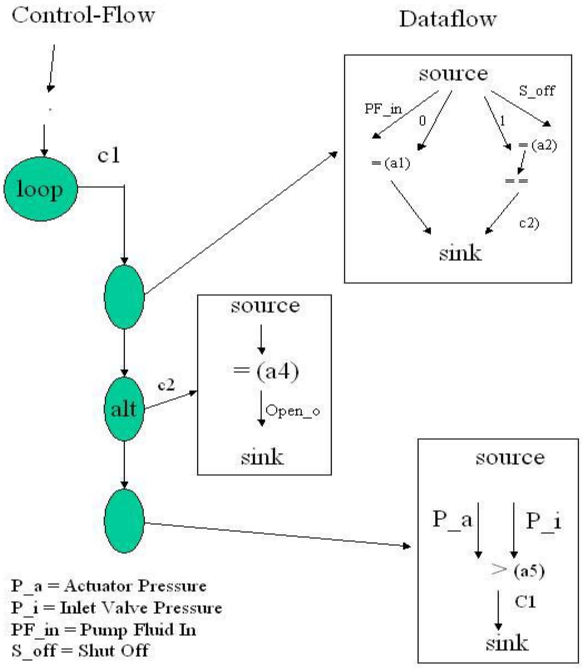
The volumetric flow rate out of the micro-pump is therefore equal to the volume displaced by the actuator chamber's diaphragm, whose movement depends upon the force induced by the electrostatic charge applied to the heating element. Thus, all timing constraints associated with the actuator chamber process are related to the following force equations:

$$F(t) = F_k(t) + F_e(t) \qquad (2).$$

$F_e(t)$ is the motive force to expel the fluid at any time t and Fk is the diaphragm spring force given by $F_k(t) = -k_s * y'(t)$. Any scheduling for the operations in the program states of actuator chamber, will have to consider the timing constraints crossing basic blocks that are imposed by the valves. The result of this effort is the development of timing expressions to capture synchronizations of models. These timing expressions contain timing relations between a set of signals that are expressed using traces of executions. Thus the micro-pump system can be specified by a set of timing expressions and the synchronization of the system specified by a set of constraints that must be satisfied. This, however, does not capture the intricate relations that are present in higher-level descriptions. As such, we now focus on the development of concurrent synchronous systems models.

IV. MODELING CONCURRENT SYNCHRONOUS MICROFLUIDIC SYSTEM

System-level descriptions are specified as sets of concurrent components interacting with each other and with the environment. The characterization of the constraints for synthesis can be obtained only if we understand the underlying behavior of the system, and its relationship to the environment. Here we us a control-flow model for system-level descriptions developed in [10]. Below we show the partitioning of the actuation_chamber process into its control-flow and data flow equivalent models of the micro-pump specification described earlier.



The above algebra of control-flow expressions (CFEs) are defined by the abstraction of the specification in terms of the sensitization of paths in the dataflow, and by the compositions that are used among these operations. The communication between the dataflow and control-flow are viewed as an event generation/consumption process. The output events generated from the control-flow are called actions. Each action is assumed to execute in one-unit of time (a single cycle). Actions that are executed in multiple cycles are handled by a composition of single-cycle actions. The input events of a control-flow are represented by conditionals. These conditionals will enable different blocks of the specification to execute. The set of Boolean formulas over the set of conditionals is referred to as guards.

Using the above control-flow description, we model systems by a set of operations, dependencies, concurrency and synchronization. Sub-behaviors of this system are thought of in terms of processes, which are represented by control-flow expressions and correspond to an HDL model.

V. Conclusion

In this work we present a system-level modeling, analysis and synthesis technique for a micro-pump. Included in this work are models of the micro-pump specification, control-flow, and dataflow. To best capture the degrees of freedom for this device we employ the use of a modeling technique for control-flow dominated specifications.

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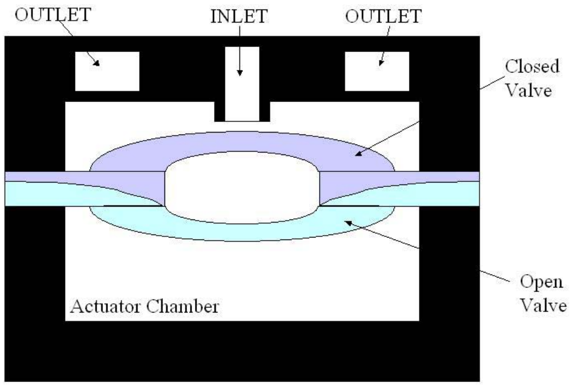

Figure 1. Cross-section of a Micro-pump

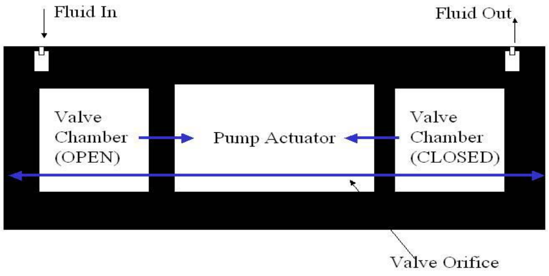

Figure 2. Block Diagram of MicroPump

Table 1. Technical Specifications for Micro-pump

Pump Attributes:	Values
Dimensions of the pump (mm)	9.3 x 10 x 1.2
Maximum height with fluidic ports (mm)	7.9
Outer diameter of fluidic ports (mm)	0.91
Flow rate without back pressure (µl/min)	150-250
Maximum pressure generated (hPa)	70-120
Life <Lab Conditions> (load cycles)	> 315 million
Drive voltage (V)	10-15
Drive pulse width (ms)	1-2
Drive frequency (Hz)	5-30
Power Consumption (mW)	~50
Pump case	Polysulfone
Pump membrane	Polyimide
Adhesive bonds	Epoxy Resin
Heater coil	Gold
Fluidic connection tube	Stainless Steel

Note: Values for Operation at 22-degrees C