Design, modeling and simulation of MEMS-based silicon Microneedles

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Abstract: The advancement in semiconductor process engineering and nano-scale fabrication technology has made it convenient to transport specific biological fluid into or out of human skin with minimum discomfort. Fluid transdermal delivery systems such as Microneedle arrays are one such emerging and exciting Micro-Electro Mechanical System (MEMS) application which could lead to a total painless fluid delivery into skin with controllability and desirable yield. In this study, we aimed to revisit the problem with modeling, design and simulations carried out for MEMS based silicon hollow out of plane microneedle arrays for biomedical applications particularly for transdermal drug delivery. An approximate 200 µm length of microneedle with 40 µm diameter of lumen has been successfully shown formed by isotropic and anisotropic etching techniques using MEMS Pro design tool. These microneedles are arranged in size of 2 x 4 matrix array with center to center spacing of 750 µm. Furthermore, comparisons for fluid flow characteristics through these microneedle channels have been modeled with and without the contribution of the gravitational forces using mathematical models derived from Bernoulli Equation. Physical Process simulations have also been performed on TCAD SILVACO to optimize the design of these microneedles aligned with the standard Si-Fabrication lines.

1. Introduction
The area of miniaturization and micro fabrication in the past few years changes tremendously in numerous engineering and scientific fields due to extensive preponderance progress in micro electromechanical systems (MEMS) and Microsystems. The BioMEMS based devices increase its efficiency and robustness with the advancement of technology in both micromachining tools and design structure [1]. BioMEMS as a whole at micro scale integrates many biomedical systems and biological sciences disciplines and applications. Sampling and extraction of biomolecules and fluids, drug delivery, therapeutics and diagnostics etc are the major areas of biomedical industry where MEMS based microstructures have been successfully implemented and more research is still in progress to connect miniaturized products with the biological environment [2]. Biocompatibility and broad protocols of experiment procedures are primarily required for the manufacture of biomedical microstructures [3].

Patients have been delivering drugs through hypodermic needles for about 160 years [4]. For the transdermal drug delivery and transdermal blood extraction hollow needles have played a very vital role...
since they were manufactured in 18th century and later shortly first injection was then administered [4]. In 1991, the foremost out-of-plane microneedle array, consisting of 100 microneedles with a length of 1.5 mm was reported [5]. Microneedle arrays offer great application and advantage over conventional needles. Microneedles have been developed for different applications with different variety of materials, fabrication techniques, design, dimensions and tips shape etc.

For the very purpose of the extraction of biological fluid and drug delivery for skin permeability, the BioMEMS based microneedle array technology provides many advantages and applications over conventional needles, including (i) The length of microneedle can be controlled for the precise parenteral injection of therapeutic agents, by considering skin anatomy; (ii) It is minimally invasive, due to smaller diameters at the tip; (iii) Microneedles are arranged in arrays in order to transport large volume of fluid and penetrate invivo the skin over large distributed area rather than using single needle [6]; (iv) Provide painless drug delivery due to the fact that microneedles do not reach the nerves deep below the skin; (v) Single wafer is being used for the fabrication of arrays of microneedles due to its small size which tends to be a good design metrics including reliability, biocompatibility, fabrication cost, accuracy and etc [7]; (iv) BioMEMS based microneedles array devices also find its application tremendously in biomedical sciences such as: Treatment of cancer; DNA transportation; Introducing biomolecules in living cell; and Diabetes.

In this work, an approach has been made to design and simulate fabrication steps for microneedle arrays for biomedical applications that would be hollow and out of plane in structure.

2. Types, Shapes and Materials of Microneedles

Based on the fabrication process techniques, microneedles are generally classified into two types that are: Out of plane microneedles and In-plane microneedles. In out of plane microneedles, the structure of the flow channel and shaft is normal and protrudes out of the surface of the wafer substrate [8]. In in-plane microneedles, the structure of the flow channel and shaft is parallel and in plane to surface of the wafer substrate [8].

With reference to the physical structure microneedles are also comprises of two types that are [9]: Hollow microneedles and Solid microneedles. Hollow microneedles are the microneedles that direct the flow of biological fluid and drugs through its internal lumen. Solid microneedles are the microneedles in which either the lumen surface is coated with particles of drugs and applied to the skin or since the base of solid microneedles is non dissolving, it is sequentially removed after the dissolution of microneedles [9].

The factors such as skin permeability and penetration along with the strong biocompatibility require an appropriate shape and selection of materials of the microneedles which is very critical during the process of fabrication and design. Any specific applications particularly in biomedical sciences, the selection of both shapes and materials is very important simultaneously. The shape of microneedles can be categorized into two types with respect to the Overall shape and Tip shape. Table 1 summarizes various designs of microneedle shapes and materials [10].

| Overall shape | Tip shape | Materials       |
|---------------|-----------|-----------------|
| Pyramidal     | Tapered   | SU-8            |
| Spear         | Beveled   | Silicon         |
| Candle        | Snake fang| Metals          |
| Cylindrical   | Volcano   | Polymers        |
| Spike         | Canonical | Silicon dioxide |
3. Skin Anatomy and Microneedle Design Considerations

The human skin is the most multifunctional and largest organ of the body that interacts with the surrounding world [11]. The superficial epithelium layer epidermis is approximately 50-150 µm thick which largely consists of cells called keratinocytes [12] which are formed in the basal layer of epidermis [11]. The major barrier for drug delivery is the stratum corneum, about 10 to 20 µm thick, [13] is the outermost layer of the epidermis adherently consisting of stacked cornified cells of hexagonally flattened shape [11, 12, 13]. The deepest layers of the stratum corneum contain lamellar granules that act as major part of the permeability layer [11, 13]. The dermis is the second thick layer distributed with connective tissues [11]. The dermis mainly consists of elastin, network of blood vessels and capillaries and collagen matrix [11, 13]. The hypodermis also called subcutaneous fat is a fatty layer which is attached loosely to the superficial dermis [11, 13]. So, the structure of skin offers the great opening for the transdermal drug delivery as well as drug absorption [14]. The microneedle should therefore not reach the dermis but it may tend to penetrate the stratum corneum into epidermis of the skin. This suggests that for transdermal drug delivery, the length of microneedle fabricated must be in range of 150 – 250 µm size, as devised in other such experiments [14].

4. Modeling Experiments – Hollow Microneedle fluidic analysis

It is very significant to develop appropriate numerical model for microneedles with the characteristics of fluid flow and to account for “lumen dimensions” as the liquid drugs invivo into the skin through these arrays of microneedle. Therefore; mathematical models are developed and reported in literature [6] to evaluate and comprehend the working performance of microneedle arrays theoretically.

4.1 Flow Characteristics through Microneedle Arrays

Bernoulli equation (BE) [15], which is one of the most useful equation for fluid mechanics has been utilized in the form of Modified Bernoulli equation (equation 1) as reported in some studies [6], with the gravitational forces are ignored in calculations and analysis. We also considered the contributions from gravitational force and re-named the modified equation as Extended modified Bernoulli equation (equation 2) for ready reference, as shown below:

\[
P_1 - P_2 + \rho g(Z_1 - Z_2) + \rho \left( \frac{V_1^2 - V_2^2}{2} \right) = \rho \frac{V^2}{2} \frac{fL}{D} + \sum K \frac{\rho V^2}{2}
\]

(1)

\[
P_1 - P_2 + \rho g(Z_1 - Z_2) + \rho \left( \frac{V_1^2 - V_2^2}{2} \right) = \rho \frac{V^2}{2} \frac{fL}{D} + \sum K \frac{\rho V^2}{2g}
\]

(2)

In equations (1) and (2), \(P_1\) is inlet pressure, \(P_2\) is outlet pressure, \(V_1\) is inlet velocity, \(V_2\) is outlet velocity, \(Z_1\) is inlet height, \(Z_2\) is outlet height and \(\rho\) is fluid density [15]. The first term on the right side of the above equations describes the flow in the microchannel and second term on right side tells about the minor losses encountered during travelling via fluid particle [15]. \(g = 9.81 \text{ m/s}^2\) is the gravitational force.

4.1.1 Analysis for pressure drop

Properties such as microneedle geometry, viscosity of fluid, microneedle density and flow rate are mainly acquired by the pressure drop in order to allow the flow of fluid through microneedles, as previously reported [16]. The above mention equations (1) and (2) are exploited for analysis of pressure drop dynamics for an individual needle.
4.1.1.1 Pressure Drop Analysis from Modified Bernoulli Equation

An individual microneedle has been modeled using water as a “model liquid” as shown in figure 1[6] with Modified Bernoulli Equation, which is used to depict fluid flow through microneedle. In case of considered needle, the channel with a “constant” cross section area, equation (1) can be further reduced to:

\[ \Delta P = \mu \frac{128 Q L}{\pi D^4} + \rho \frac{8 (K_1 + K_2) Q^2}{\pi^2 D^4} \]  \tag{3}

Equation (3) tells that inside a circular tube the viscous shear force of Poiseuille flow induced the pressure drop [17] and secondly corresponds to inertia effects along with the minor losses \( K_1 \) & \( K_2 \) at the entrance and exit of the needle respectively [18]. Furthermore, the equation (3) illustrates that the contribution from the gravitational force ‘g’ are neglected. In the above calculations Gravitational forces were not considered because in previous studies it was assumed that at the time of flow, they might be negligible being relatively small [6]. This, however, is revisited in our studies conducted during the design and modeling of such microneedle devices, governed by equation 4 below:

\[ \Delta P = \mu \frac{128 Q L}{\pi g D^4} + \rho \frac{8 (K_1 + K_2) Q^2}{\pi^2 g D^4} \]  \tag{4}

This equation tells about the influence of gravitational forces ‘g’ along with the pressure drop inside a circular duct of Poiseuille flow due to the viscous shearforce which in turn also corresponds to inertia effects along with the minor losses \( K_1 \& K_2 \) under the influence of gravitational forces ‘g’, which may be of greater importance when using the devices on altitudes and/or deviations from a standard “g” value systems.
4.2 Analysis for Microchannel Diameter

Diameter is one the very important parameter for the design of microneedles because it describes the “channel inner” from where the fluid flows and injected into the skin. This parameter is dealt with full appreciation in our design. Again, two approaches for microchannel diameter design have been employed by using the equations (3) and (4) which in turn obtained the following two equations that are,

\[
D = \left( \mu \frac{128}{\pi \Delta P} Q L + \rho \frac{8 (K_1 + K_2)}{\pi^2 \Delta P} Q^2 \right)^{0.25} \tag{5}
\]

\[
D = \left( \mu \frac{128}{\pi g \Delta P} Q L + \rho \frac{8 (K_1 + K_2)}{\pi^2 g \Delta P} Q^2 \right)^{0.25} \tag{6}
\]

Equation (5) illustrates microchannel diameter under the pressure drop dynamics where the gravitational forces “g” are ignored, whereas equation (6) illustrates microchannel diameter under pressure drop dynamics with the gravitational force contribution considered.

The summary of the underlying assumption for our design are as follows:

For equations (3), (4), (5) and (6), \( f = \frac{64}{Re} \) is the friction factor which has been used for laminar flow. Standard macroscopic values for an edged inlet \((K_1=0.5)\) and for an exit \((K_2=1.0)\) were chosen in line with similar studies conducted [6] to model the flow characteristics of the \( L = 550 \) µm long microneedle lumen [6]. \( \mu = 1.006 \times 10^{-3} \) Pa s and \( \rho = 1000 \) kg/m³ were assumed as viscosity and density respectively as standard values for the model [6, 15] and \( g = 9.81 \) m/s² is the gravitational force which is subject to change on standard calculations for altitudes.

5. Design Experiments I – Design and Fabrication processes of Microneedles

5.1 Experimental tool: MEMS PRO and Process Steps (Design Strategy)

The simulations for design and fabrication processes of microneedles were carried out on MEMS PRO design tool which is a computer aided design tool provided by the SOFTMEMS Company that is reliably utilized for the scalable design, modeling and analysis of Micro Electro Mechanical Systems (MEMS) [19].

The properties of the needles are generally reflected by the structure of microneedles [20]. The primary considerations for our proposed design of microneedles are:

(a) That during the insertion into skin they must not break or bend. This demands a needle with sharp or pointed tip along with a wide base must be taken into account;

(b) That during the injection period, there are some lateral forces that tends to act on the microneedles, which in turn produce bending moments on the needle body. This demands that the microneedles must have such good design that they withstand with all certain forces;

(c) That one needs to organize the microneedles in the form of arrays because by designing them in arrays the fluid obtained would be in large volume and also be able to have a penetration onto the skin over a large distributed area [21]. For successful resulted microneedle, it must have pointed tips, hollow and out of plane shape.
In order to sketch out the 200 µm long hollow out of plane symmetric microneedle array for biomedical applications, simulations for fabrication process on process simulator MEMS Pro were executed and illustrated in the following steps:

STEP 1: The design process started with initial substrate of silicon wafer with 550 µm thickness shown in figure 2(a), having <100> orientation and polished on top and bottom side.

STEP 2: The second step is to deposit mask material on the top (surface side) and bottom (bulk side) of the silicon wafer shown in figure 2(b) and for this purpose 6 µm thickness of silicon dioxide (SiO$_2$) is deposit by applying a process of Low Pressure Chemical Vapor Deposition (LPCVD).

STEP 3: Photoresist is then deployed on the bottom side of the wafer and by applying photolithography, process pattern was transferred on the oxide layer (SiO$_2$) for the hole of 40 µm in diameter shown in figure 2(c).

STEP 4: The anisotropic etching which accomplish by Deep Reactive Ion etching (DRIE) is applied to form the needle channels deep down the silicon substrate to a depth of 556 µm from bottom side stopping close to the top oxide layer as shown in figure 2(d).

STEP 5: 4µm thickness of silicon nitride by applying a process of LPCVD is deposited on the top and bottom along with the channels shown in figure 2(e) which will act as a protector during the next steps of etching including isotropic etching of silicon.

STEP 6: On the front side of the wafer photoresist was deployed and circular pattern was transferred by applying photolithography process on the silicon dioxide and silicon nitride masks shown in figure 2(f) which could be utilize in the next step to create the needle shaft.

STEP 7: Now to craft the outer shape of the needle, isotropic underetched is performed asshown in figure 2(g). The isotropic undercut parameter is primarily utilized to optimize the etching of silicon beneath the circular masks horizontally. Here the silicon nitride plays its critical role and protects the side walls of channels. The obtained outer shape of the needle has a 200µm long shaft with a wide base.

STEP 8: In the last step the remaining silicon dioxide and silicon nitride are stripped resulting in a final structure of microneedle as shown in figure 2(h).

Figure 2. Fabrication Processes Steps for Hollow Out of Plane Microneedles (a) Silicon Wafer (b) SiO$_2$ Deposition (c) Backside Etching of SiO$_2$ (d) DRIE for Needle Channel (e) Depostion of Si$_3$N$_4$ (f) Pattern transfer for Front Side (g) Isotropic Etching for Needle Shaft (h) Removal of SiO$_2$ and Si$_3$N$_4$.
5.2 Microneedle Design Dimensions
The modeled symmetric and pointed tip hollow out of plane microneedles are shown in figure 3(a) and figure 3(b) respectively, consisting of following main design specifications. (i) Silicon wafer thickness = 550 µm, (ii) Needle length = 200 µm, (iii) Lumen diameter = 40 µm, (iv) Center to center spacing between needles = 750 µm, (v) Array size = 2 x 4.

6. Design Experiments II - Process selection optimization for the fabrication of microneedles
6.1 Experimental Tool: TCAD SILVACO
Process simulator ATHENA given by SILVACO TCAD has been used for the fabrication of microneedles to gain optimized process selection parameters. This tool offers process simulation and modeling of semiconductor devices with a variety of modeling options [22].

6.2 Broader Process Parameters Steps (SILVACO) Specific to Si-CMOS fabrication (Fab) Line
Since the SILVACO is primarily based on Si-CMOS processes, therefore the processes fabrication steps for microneedles simulated is slightly different from the processes fabrication steps designed in MEMS PRO design tool. This in turn facilitates the device design engineer to make the fabrication and batch production of MEMS based microneedles easy, cost effective, reliable and flexible on standard Si-CMOS fab lines, without altering the experimental set up a great deal. The processes used in SILVACO are invoked by simple physical process parameters rather than using machine parameters. The process steps for the optimized fabrication of microneedles are provided in the following steps. The selection of the optimized process parameters and final design shapes (refer to figures 3(a) to 3(h)) are based on numerous routines run on the simulator and in close proximity to the design maintained in MEMS PRO, detailed in section 5.

STEP 1: The first step is the selection of appropriate substrate or material and preferred material is silicon as shown in figure 4(a).

STEP 2: Background doping is then carried out in the silicon wafer with p-type boron impurity to modulate its electrical properties and also to provide a region for preferential etch.

STEP 3: Now in order to craft the outer shape of the microneedle at a height of 200 µm shown in figure 4(b) geometrical etching of silicon is performed by invoking arbitrary points.

STEP 4: In this step silicon dioxide is thermally grown by dry oxidation in order to protect the side walls and top of the needle shown in figure 4(c).

STEP 5: Following the thermal oxidation it is then annealed to strengthen the sidewalls by activating the silicon atoms and also to create a sharp tip of the microneedle during subsequent oxidizing process as shown in figure 4(d).
STEP 6: Now the grown silicon dioxide is selectively geometric etched away as shown in figure 4(e). The remaining portion is covering the needle top.

STEP 7: Silicon nitride ($\text{Si}_3\text{N}_4$) is then conformally deposited shown in figure 4(f) mainly to cover the entire channel of the needle and top of the needle tip.

STEP 8: Nitride is now annealed as can be seen in figure 4(g) to harden the channel of the microneedle.

STEP 9: Now the silicon nitride is selectively geometric etched away as shown in figure 4(h) remaining on top of silicon dioxide.

STEP 10: As a last process, minor steps were performed including mirroring the structure and stripping of remaining silicon dioxide ($\text{SiO}_2$) and silicon nitride ($\text{Si}_3\text{N}_4$), as can be seen in figure 4(i). In the end microneedle with sharp tip shape is formed as shown in contours of the figure 4(j).

![Figure 4](image-url)

**Figure 4.** Process Simulation of Microneedles on SILVACO Process Simulator, (a) Silicon substrate (b) Geometrical etching for microneedle outer shape (c) Deposition of $\text{SiO}_2$ (d) Annealing of $\text{SiO}_2$ (e) Geometrical etching of $\text{SiO}_2$ (f) Deposition of $\text{Si}_3\text{N}_4$ (g) Annealing of $\text{Si}_3\text{N}_4$ (h) Geometric etching of $\text{Si}_3\text{N}_4$ (i) Final shape of microneedle (j) Contour of final shape microneedle with sharp tip.

7. Discussion on salient features of results

The design and simulation results are detailed at length in sections 6 and 7. Based on the design assessments made and modeling of the underlying mechanics deliberated in section 4, we provide an insight to these results in this section and discuss them in line with the objectives set forth for this study:

The microstructure needles were designed to inject or extract out the drug or fluid from skin via stratum corneum. Mathematical models were re-visited in form of Modified Bernoulli Equations (1) and Extended Modified Bernoulli Equation (2) which are further reduced to equation (3) and equation (4) respectively. These revised models simulated for fluid flow characteristics through needles. The physical relationships between several parameters such as flow rate, pressure drop and needle diameters were also developed. Modified Bernoulli Equation was re-visited in such areas where gravitational forces are either considered as negligible or ignored, whereas in Extended Modified Bernoulli Equation gravitational forces are taken into consideration, for applications of devised microneedles on ambients where the contribution of gravitation forces are appreciable.
7.1 The contribution of gravitational forces

The gravitational forces were mainly not considered during previous studies of microneedles [18] and treated as negligible during the numerical and experimental analysis of microneedles. In our study, the gravitational forces (g) were taken into account during the numerical analysis in order to see its effects on design and flow characteristics of microneedles. During the comparison of two models, it was found that gravitational force has appreciable impact on geometric aspects of the design and subsequently on its flow characteristics. Gravity is known to have effects and impact on human body as well as on other biological entities, therefore the gravitational forces may become a highly considerable parameter during the design of biomedical systems when used in special environments such as altitudes. This has been a missing link so far, largely in the literature and very little is known on this account. One very recent study published to address this issue partially [23] claimed that during the design of microneedle arrays for biomedical systems, gravitational forces may also be taken into consideration for the transport of drug formulation [23].

7.1.1 Relationship between Pressure Drop and Flow Rate

The pressure drop as a function of flow rate, diameter as a function of flow rate and diameter as function of pressure drop with microneedle geometry has been simulated in our study for an individual microneedle using water as a model liquid with the help of equations (3) and (4). Standard macroscopic values for an edged inlet ($K_1= 0.5$) and for an exit ($K_2= 1.0$) were chosen to model the flow characteristics. Other parameters are: $L = 550 \mu m$ long microneedle lumen, $\mu = 1.006e^{-3}$ Pa s and $\rho = 1000 \text{ kg/m}^3$ were assumed as viscosity and density respectively and $g = 9.81 \text{ m/s}^2$ is the standard gravitational value.

![Pressure Drop vs Flow Rate](image)

Figure 5. Variation between Pressure Drop and Flow Rate

Figure 5 illustrates the relationship between pressure drop and flow rate simulated at a fixed 40 $\mu m$ channel diameter of the designed microneedles. From this result, one can notice that the diameter of channel is determined by desired flow rate and saturated pressure drop. The curves show that as the flow rate increases, the pressure drop also increases initially for both conditions of “g”. The pressure drop however decreases dramatically in case of gravitational force accounted for as compared to the pressure drop without g contribution. It is due to the physical implication of gravity on the viscous and dense medium and directly impacts the behaviour of pressure dynamics due to a certain maintained flow rate within the device.
7.1.2 Relationship between microneedle diameter and pressure drop

Figure 6 discusses the relationship between the variations in diameter and pressure drop simulated with constant flow rate at Q = 400 µl/min and 800 µl/min. As the pressure drop increases, the diameter gets decreased in both the conditions i-e with and without taking the contribution of ‘g’ into account. The curves of diameter vs pressure drop are saturated after the drop of pressure reaches a certain value. The diameter of lumen increased with the increased flow rate at certain values of pressure drop which is true for both the conditions. In this graph the curves are becoming flat with the increase of the pressure drop. The region of flattened lines illustrates that the diameter is not very sensitive to the variation of the high pressure drop. Hence a smaller diameter is required for the higher pressure drop at the desired flow rate in the microneedles.

![Diameter vs Pressure Drop Graph](image)

**Figure 6.** Variation between Diameter and Pressure Drop

It can be noticed that by comparing this graph for both conditions of ‘g’, the diameter gets much reduced in case where value of “g” is accounted for as compared to the diameter modeled without “g”. This also suggests that at a certain value of pressure drop, the increase in the flow rate makes the diameter of lumen increased.

We gathered from these experiments that the designed diameter of the microneedles can be determined at the saturated region of diameter vs pressure drop curves. The slope of the graph presents the variation of the diameter with pressure drop. After the pressure drop reaches a certain value, the variation of diameter with pressure drop is negligible, which is true for both the conditions of gravitational forces.

Further, it can be gathered from the results presented in Figure 6 that the velocity of liquid flow also goes to a decrease with the increase in the diameter, which results in lower pressure drop across a microneedle. Since the water has low viscosity so the flow rate will increase readily. This is true for both conditions of “g”. However, the values become smaller for all above parameters when they are compared with the condition where experiments are modeled without taking the contribution of “g” into account.

7.1.3 Relationship between diameter of the microneedle and flow rate

Figure 7 illustrates the relationship between the variation in the diameter and flow rate simulated with two separate constant pressure drop at 75.9 kpa and 304 kpa. The diameter gets increased with decrease of pressure drop at certain desired flow rate for both conditions of “g”. This means that the pressure that is needed to flow the fluid rises dramatically when channel diameter decreases. In figure 7, it can further be
seen that a very small decrease in the diameter reduces the flow drastically. This is due to the fact that cross section area (which is directly proportional to the flow rate) gets impacted for both conditions of “g”. By comparing the value of diameter with and without the consideration of value of “g”, it is found that by including the value of “g”, the diameter becomes much smaller at higher pressure drop at a desired flow rate, which suggests that there is higher velocity experienced at that particular point.

Figure 7. Relationship between Diameter and Flow Rate

7.2 The Geometry of microneedle & Physical Processes
The pressure drop is mainly dependent on the geometry of microneedle in combination with fluid viscosity and density for a smooth fluid transportation through microneedle channel. Therefore appropriate selection of needle geometry leads to an effective biomedical microsystem device. The Deep Reactive Ion Etching (DRIE) is suggested to be performed to create the microneedle lumen. The outside shape of microneedle is suggested to be created by isotropic undercut of silicon dioxide and silicon nitride masks.

The geometry of the design is adjusted in such away to provide a solution using which a process engineer may fabricate MEMS-based microneedles with the following characteristics:
(a) A miniaturized cross section (controlled by diameter modeling)
(b) A substantial capillary flow rate (controlled by the modeling of pressure drop, contribution of gravitational force and optimized diameter)
(c) Full strength not to break during the skin penetration (controlled by physical process optimization)

Physical process based simulations carried out as shown in Figure 4 are aimed to optimize the structure and design of the microneedles. The purpose of the process simulation of microneedles is to predict and manipulate the design structure knowledge that has been obtained after applying specified process steps which was not possible to obtain from the MEMS PRO design tool (refer to figure 3 and its constituents, for example). Since the ATHENA simulator is primarily based on the CMOS fabrication process, the dimensions and design used for microneedles in these simulations are complex, relevant to the fab-less solutions and cost effective in nature. The selection of parameters (refer to figure 4 and its constituents and steps elaborated in section 6) is purely iterative in behaviour and provides a lot of insight while optimizing the design of MEMS-based microneedles by the optimization of both the modeling and physical process parameters. Our study is somewhat different as the utility of both the powerful design and modeling tools for the design and consequent fabrication of MEMS-based microneedles is focused on (a) the cost effective Si-CMOS fab specific solutions, and (b) the physical process parameters rather than machine...
parameters in TCAD SILVACO environment, which makes it viable for design optimization studies and applications rather than an ‘embedded-only’ solution in a mechanical application.

8. Conclusion
This work was focused to the design and fabrication process of “hollow out of plane” silicon based microneedle arrays for biomedical applications in MEMS environment. The purpose of such microneedles was to offer a design solution which would in turn control physical parameters for a successful painless epidermal drug delivery and cause no trauma at the site. The design of microneedles presented in this work is carried out on the MEMS PRO design in conjunction with TCAD SILVACO process simulation tool. The MEMS PRO design provided the basic geometrical definition of the device. With the help of MEMS PRO, it was found that Bernoulli Equation (and its revisited versions) had been a good approximation for liquid flow through microneedle lumens. TCAD SILVACO was utilized to improve the specific design in line with standard Si-CMOS fab lines. A trade-off process optimization was thus achieved between fab-less and in-fab solutions for MEMS-based microneedles for effective applications in biomedical industry. The design in this work has been specific to transdermal drug delivery so the dimension of microneedle shaft was set to be 200 um long which might prove to be consequently sufficient to easily penetrate the skin barrier stratum corneum and reach the epidermis. An appropriate selection of microneedle diameter was needed that could deal with the design limitations. With large value of diameters, the microneedle channel got widened making the base of microneedles wider. But for pointed microneedles (our design) with small diameter, sharp tips were created intentionally. Therefore diameter ranging between 35 µm to 45 µm was found suitable for the microchannel created with both types of microneedles.

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