Bio - Microelectromechanical Systems

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Module No. # 01

Lecture No. # 32

Hello and welcome back to this lecture 32, on Bio-Microelectromechanical systems.

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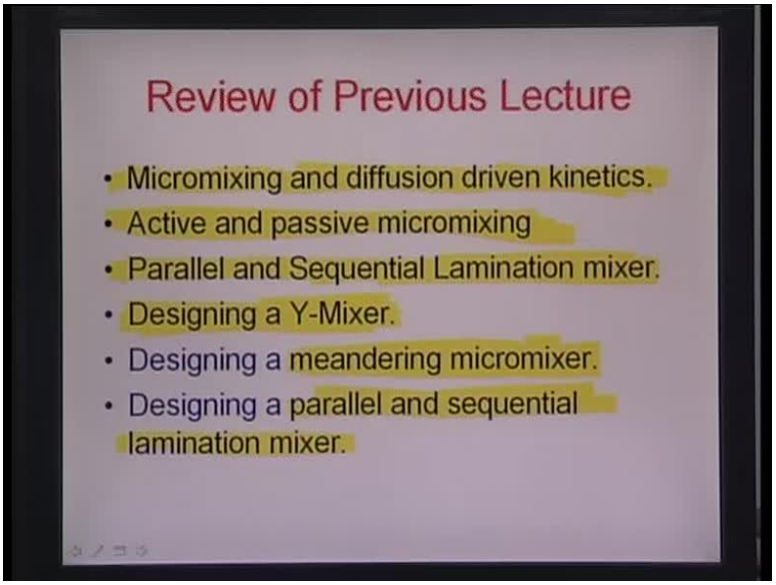

Let us quickly review what have been done last time, we talked about the principles of micro-mixing and diffusion driven kinetics, just like to retreat that in a micro-scale mixing, really is a diffusional phenomena. There is no as such other mass transport except inter molecular diffusion, therefore diffusion time, residence time and their comparison becomes a very important criteria for establishing proper mixing.

We talked about active and passive micro mixers. Active mixers, to may recall, are essentially those mixers where energy is been supplied externally by mechanical or non-mechanical means. Passive mixers are pure mixing by intelligent design, where you can actually split the flows apart many times and join them back, so that they form lamina and that way you reduce the inter diffusion length between the different mixing layers.

Therefore, the time of diffusion can be reduced in this manner and taken much below the residence time, so that there is mixing till and until the flows reside in the chip, by the time it gets out they are totally mixed.

We talked about parallel and sequential lamination mixers; parallel mixers are where you will have multiple streams focusing on to a small cross sectional area. So that the diffusion length becomes equal to the actual length divided by n, where n is the number of such streams. Sequential being, basically trying to go out of plane and then back in plane, so that you can physically laminate the flows together. Therefore, there is a reduction in the diffusion time by about 4 to the power of n minus 1; where n is the number of stages where this mixing takes place.

We also tried to do some numerical examples, designed a Y shaped mixers, found out what is the length of the micro channel for the mixing to be spontaneous and proper. We also designed a meandering micro mixer, reduced it into a small 6 mm by 6 mm area, try to define number of turns that a channel of finite length would take if it comes into this particular area. Then, we also talked about different designs for parallel and sequential lamination mixers, based on this diffusion time reduction concept compared for a single - for a certain amount of mixing time ratio both the stages. We found out certainly that in case of sequential lamination mixers, lesser number of stages is needed, in comparison to there, in the case of parallel lamination mixer for proper mixing.

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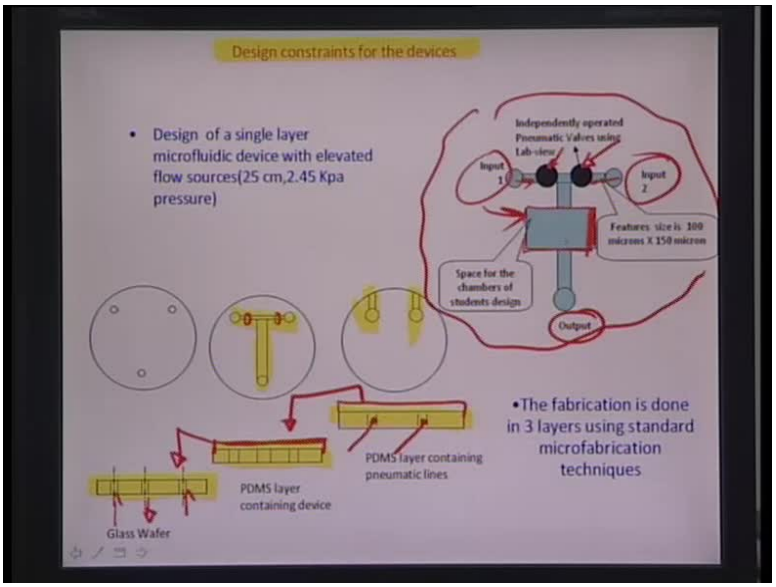

Let us look at today a little more of experimental micro fluidics. I would like to discuss a small example here, which is actually performed by graduate students. Essentially, student starting new in this area were left astounded by the counterintuitive nature of the micro flows. Definitely they are much way beyond the concepts that you can have formed the macro scale mixing or macro scale turbulence driven phenomena.

Here, in this design experiment, we were supposed to design a three layered mixer with a glass layer at the bottom, as you can see here. This glass layer has been drilled with some holes, so that you can have these holes for inlets and outlets of the fluid R.

We have another layer at the top of it, which is actually a PDMS layer with a T section channel, as you can illustrate in the top view (Refer Slide Time: 04:03). Here, there is some sort of a continuous layer of PDMS at the top of these channels; so that they are covered, they are replicated on the lower surface of the PDMS. This is bonded to this glass layer and then further you have another layer, where we have these blisters - the third layer. In this particular layer again, you have certain thickness of PDMS over the blisters.

The blisters can be demonstrated or represented here, are actually air pockets. This further is bonded back into the top of the first PDMS layer. You have a three layered or a stack of three layer devices, here what you do is, you use these ports for feeding in and out the fluid flow. Essentially, you feed these particular ports here, to inflate and deflate the air blisters. So, you can actually compress these channels on both ends, as you can see here in this particular figure.

Here, as you see, there are these two blister valves, which would actually regulate the flow coming from these two inlets input 1 and input 2. The idea is that these two: one may be dye, one may be water and they may be mixing along this output here, of the steam or outlet of the steam. They will be mixing along this particular zone of the stem of a T. This particular T again, you have an area, which is unknown. The idea was, the students in this particular experiment was supposed to design a certain features and structures in this area, so that they could promote quick and rapid mixing in this particular architecture.

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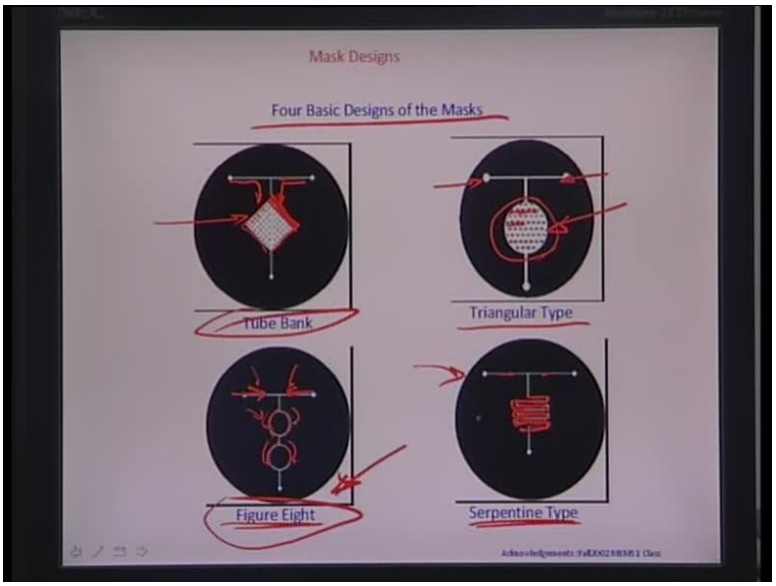

Several designs were proposed at the first instance by the students. The four most prominent designs which were proposed are tube bank like structures. Basically, they would be small pits in this particular area; this would be corresponding to small pillars of the PDMS, which gets replicated on the top of this. So, this was one illustration, where this central area here is introduced with this kind of a feature or a structure; the flows go from the arms of the T, mix along this particular area, which is also along the stem of the T.

Similarly, here, if you see in this particular example, you have flows coming from both ends here and mixing along this circular area, which is essentially made up of small triangular pieces of resist. If you just do the replication on the top of this, it becomes triangular pieces of PDMS pillars on the top surface of the reservoir, which is shown here in this particular example.

The third design was a figure eight type of a system, where there would be repeated amount of mixing and split ups of the flows; so you have flow one and two coming from both these directions. They would mix and split up again, then again mix and so on, so forth; this is the figure eight kind of design. Then, in the fourth design, the students planned a meandering shaped micro mixer, where they would have this entire fluid goal lengthwise, side by side, with a certain interfacial area along a bigger track, which is defined by this meandering channel between the small regions on the stem of the T.

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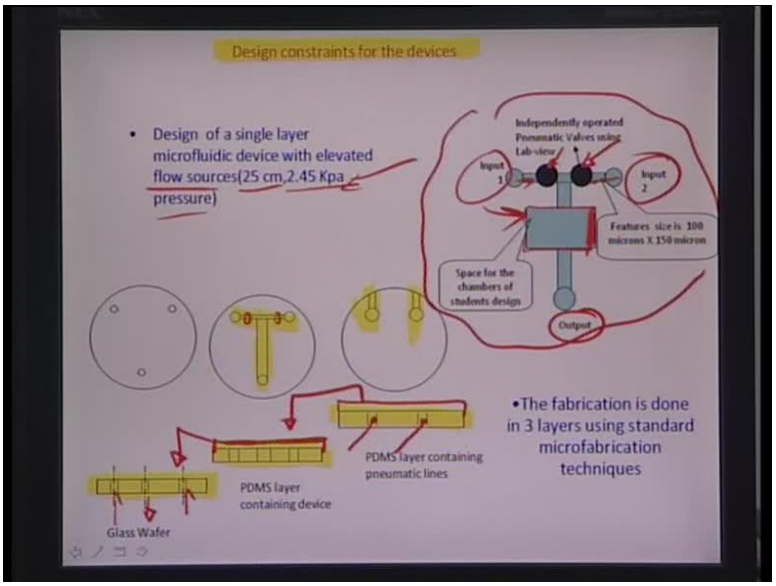

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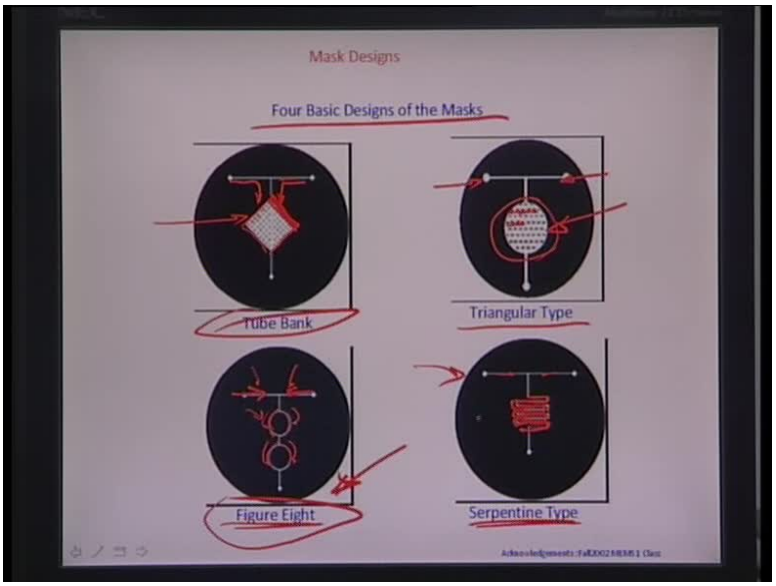

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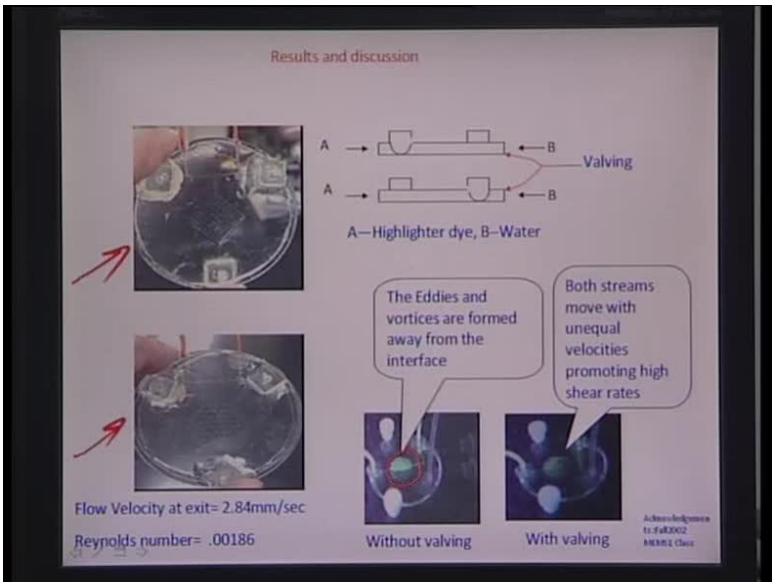

All this four designs are investigated for mixing effects, fluorescence dye was used from one end and it was gravity fed. The idea was which was given in the constraint that it should have the flow sources, elevated it to about 25 centimeters, which means that pressure of equal to 2.45 Kilo Pascal's is to be generated. After flown, this kind of a pressure driven flow through these micro channels, following results came and they were really counterintuitive.

When you actually use these structures, whether it is the tube bank kind of structure or the triangular type, you have really, as you see here, the fluorescence dye and the water flowing parallel. What is interesting to observe here is that although there is lot of these small PDMS pillars around, which the fluid should rotate and it should create local votices or eddies, they are too small to cause any mass transport, particularly across the interfacial layer.

Even though there exist some eddies, some votices away from the interface, they are not significant to contribute any kind of mass transport across the two layers. Therefore, until and unless you really start operating the blisters and block, one flow at a time, try to move this cross section as you are seeing here, in this case, the flow line with the dye has been pinched, so the flow of the dye has reduced to one side. Similarly, when the other inverse effect is done with the waterline pinched, the dye would try to go. So, it switch passed - the interface switches passed in a very serpentine like manner, as you actually

try to open and close the dye, and the water one at a time, through this automated blister cell that does result in some amount of mixing, although not very appropriate.

The take home message here for the experiment was that even though you have structures and features, which could cause eddies to develop, they would be too far away from the interface or they would not be much significant effect of these eddies and votices away from the interface, to cause the mass transport between the two streams.

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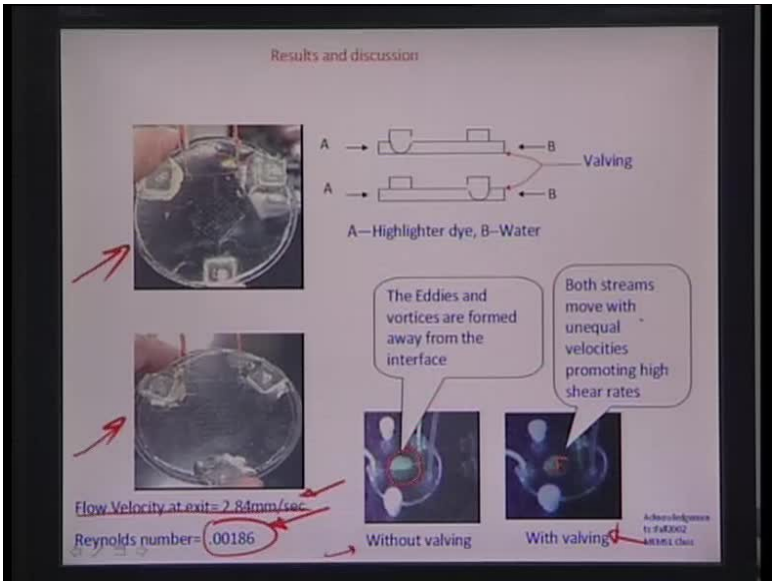

A couple of things here to be mention; one is the flow velocity at the exit, is roughly about 2.84 millimeters per second that is how it was defined. Corresponding to that the Reynolds number was really low about 0.00186, so this is without valving and this is with valving (Refer Slide Time: 10:50).

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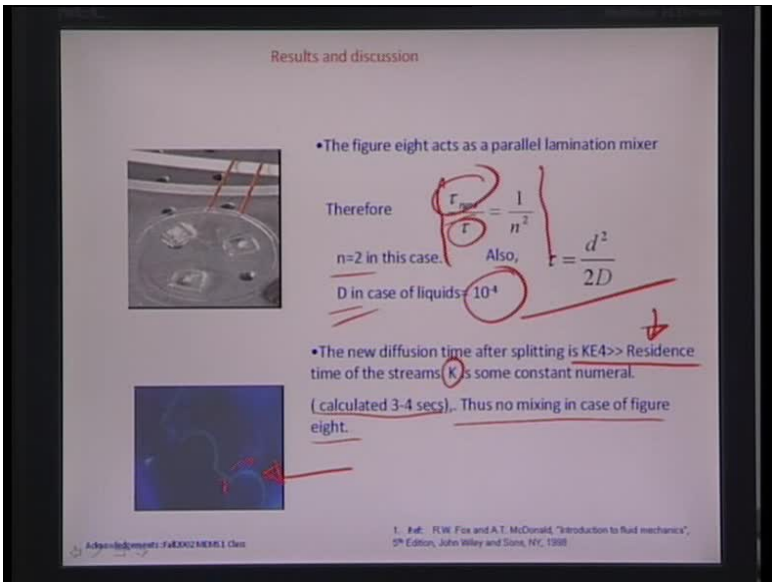

The other two designs also were evaluated, the figure eight type design, basically would be something like a parallel lamination mixer. As we all know, for in the case of parallel lamination mixer, the time new would be actually equal to 1 by n square times of old time and n in 2 in this case. D essentially in this case is about 10 to the power of minus 4, being a fluid.

If you see here, this illustrates how these fluorescent dyes would behave with water. You can see there is some portion of the dye or steam of dye, which actually goes in here, in the second illustration. Similarly, we find that there is a component of water, which goes on in this second illustration here. The new diffusion time after splitting is about some numerical constant K times about 10 to the power of 4 seconds. In this particular case, it is much greater than the residence time of the two streams; K has been calculated to be about 3 to 4 seconds in our particular case.

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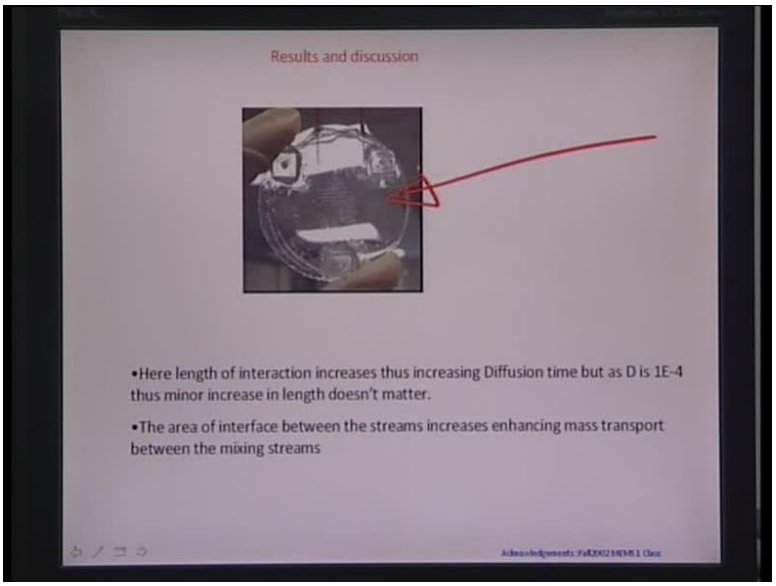

There is no mixing, though but then there is a lamination effect, which is created. There is some streaks of dye, which goes into the waterside, some streaks of water, which goes into the dye side, particularly in the second junction, which is pretty much what we expected in case of a parallel lamination mixer. The only design, which worked in our case, in this particular experiment, was a serpentine design. You can see this design here; although, I do not have the mixing illustrations in this particular slide, but then what effectively we observed here is that there would be an interaction between the green dye and water, if you pass it through a serpentine length. One of the reasons is that is source, because we think that the interfacial area, which is also proportional to the mass transport at a constant flux of diffusion being more in this case, because of more length of the serpentine would promote a mass transport rapidly between the two flowing streams.

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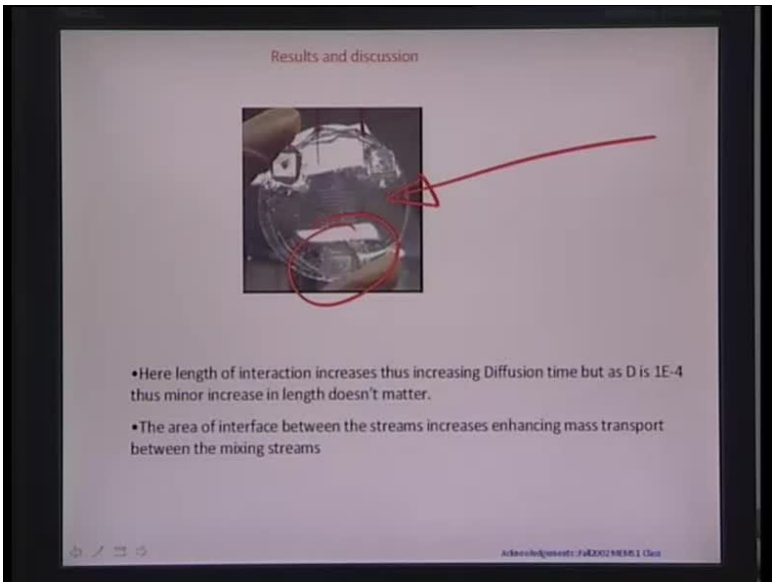

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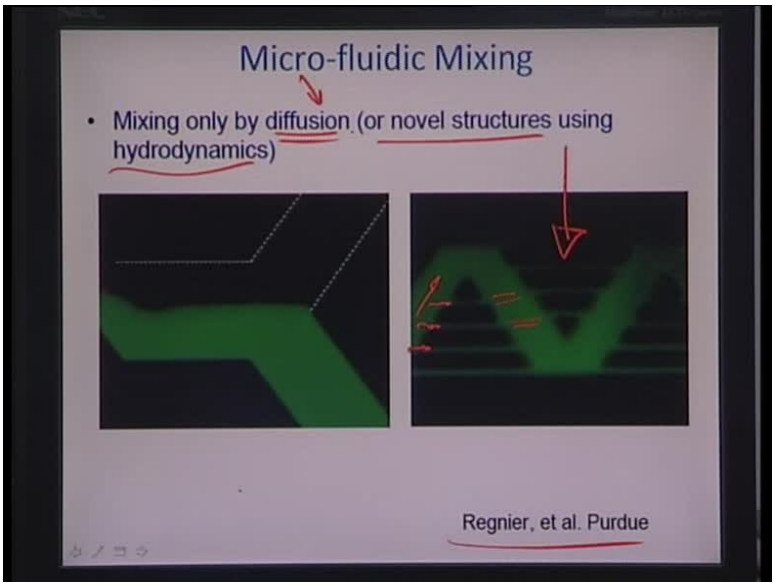

This rapidity of mass transport continues for a longer time, because of the meandering shape and the long length of the chance. You are actually trying to reside the flows for a longer time, so that this overall interfacial area gets significantly higher. We did see some mixing at this end here, in this particular kind of illustration or design. Some other examples, this has been quoted from this paper by the Regnier et al all from Purdue, who talks about how you can actually introduce novel structures in order to promote mixing, whereas, you can see there is a small amount of fluid being taken away from this flow path and being put into the next zone here, as laminates or bands.

As you go along this, is really a very rapid form of micro mixing that can happen, although, all these novel structures, using this different concepts from hydrodynamics, the mixing principle is really only diffusion.

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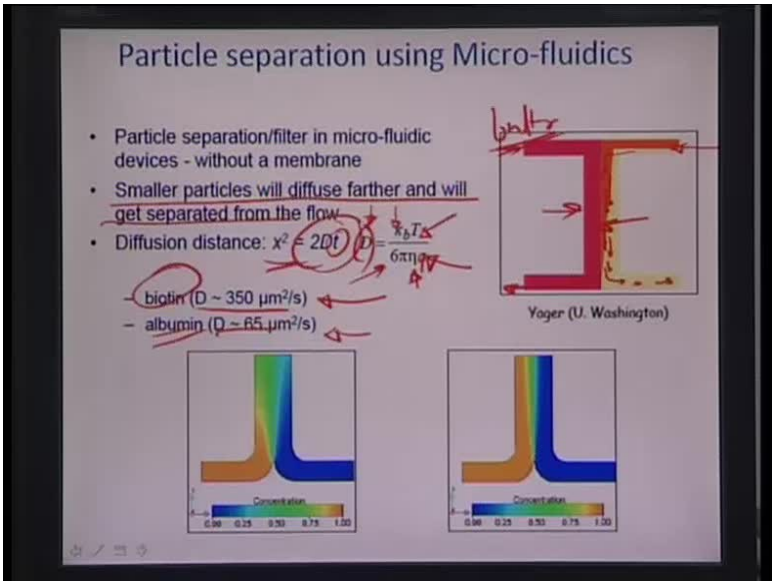

This is another example by Yager et al, which talks about the separation or particle separation, just by using a micro fluidic set up, without any filtration mechanism, here without any membrane.

Here, the concept is very simple, as we know, the diffusion length x square that a molecule would traverse is also proportional to the diffusion time. The constant of proportionality is 2 D, where D is the diffusion constant. If you really calculate the value of D, it depends on several factors like the size of the molecule, the viscosity of the medium. This K b T is basically a Bozeman constant into temperature T of the particular medium.

If suppose diffusion constant is given in this manner, you have diffusion constant higher in case of one species and lower in case of the other species, then by virtue of equality here, the distances that it may have to move would be much higher in the same kind of time. Therefore, the velocity of a heavier molecule, the one with diffusion coefficient more - I mean the velocity of the molecule with the higher diffusion coefficient is definitely more.

Also, interesting is the fact that lighter the molecule is or lesser the size of the molecule, more would be the diffusion coefficient, so biotin in this case is much smaller in comparison to albumin, which both of them are proteins essentially. So, because of its smaller size, the diffusion constant in case of biotin is about 350 micrometers square per second in comparison to albumin, which is about 65 micrometer square per second, suppose one fifth. So, the size effectively is also about five times in case of albumin.

Smaller molecules diffuse faster that is very natural tendency, this effect can be in principle used for separation, how? Let us say, you have two fluids here. This is the buffer solution, which you are running through this particular channel. There is also some kind of a mixed solution, which you are running here essentially from this end. The idea is that as the flow goes past this, there is a phase separation, because the idea is that the heavier molecule will not be that mobile in comparison to lighter molecule.

Lighter molecule here, as we assume, would diffuse towards this end actually, will flow in this direction. So, assuming that to happen, ones which are left behind in this yellow region as you can see are the heavier molecule, so the heavier molecules typically go out, when the lighter molecules can be separated, can be extracted, as carried by a transported by this buffer at this particular end. Therefore, smaller particles diffusing further will get separated from the steam, by virtue of a higher diffusion constant or a higher velocity in comparison to the smaller size particle. So, you can use these mechanisms really for particle separation.

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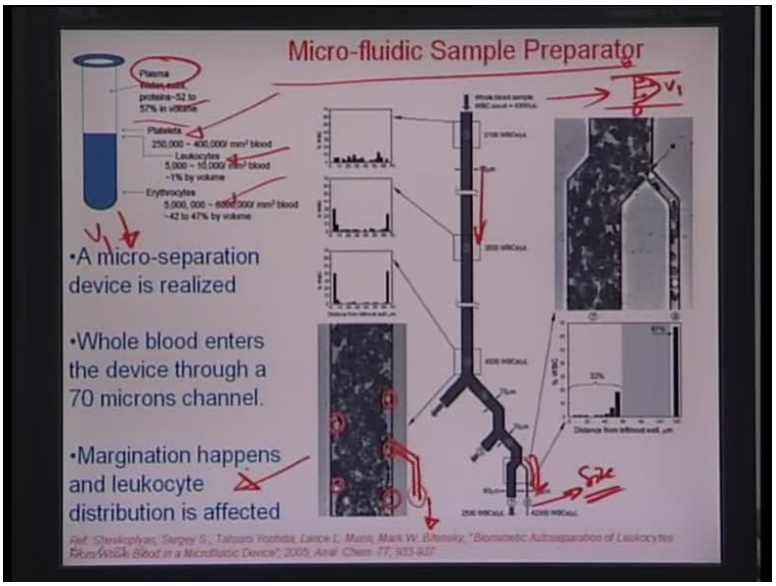

Another interesting example of where - what micro fluidics can do? Comes from this paper by Buttons Kiddo, where they talk about a biometric auto separation of Leukocytes. Leukocytes are, if you compare the sizes, are pretty much higher in size or diameter. Then, the most abundant species inside the human blood is red blood cells.

If you look at really human blood contains plasma, plasma is again bunch of ions present in solution, this is liquid plasma, so you have different salts and waters, about proteins 52 to 57 percent by volume, which is immersed or immersed species in this plasma are platelets, which are about 250000 to 400000 per millimeter cube of blood. Leukocytes which are about 5000 to 10000 per millimeter cube of blood, this is about one percent by volume again. Erythrocytes red blood cells, which are about 5000000 to 6000000 per millimeter cube of blood and about 42 to 47 percent by volume.

What you are seeing here is that blood is really a very diverse component, with a lot of different sizes of these cells, whether its red blood cells are Leukocytes which are flowing around. Therefore, we can really use micro fluidic principles for separating the blood flows. As you know, as we have been talking about, before that there is a parabolic flow profile, which develops in channels, which have fixed or static walls on both sides and the flow within is driven by a constant pressure gradient.

Let us suppose, you have a parabolic profile, you have a channel here and you have a parabolic profile like this (Refer Slide Time: 19:17). Obviously, it means that more

towards the center, the velocity would be more. They would be a 0 velocity or a 0 slip on both ends. Therefore, from site to center, the V should actually continuously reduce. If suppose you have a case, where you are flowing through this end, through a pressure driven flow, a bunch of different cells immersed in a solution; one, which is bigger; another, which is smaller. What do you think will happen? The bigger cells would try to move towards the edges by the principle of conservation of momentum, because the velocity that they will encounter by doing so the edges, are much smaller in nature in comparison to the velocities at the center.

Automatically, by the principle of conservation of momentum, there would be a tendency of these bigger cells to marginate towards the sides, smaller cells to come towards the center. This is actually a principle, which happens inside the human body also; particularly for Leukocytes, this principle is known as leukocyte margination, because of smaller sizes of the capillary and the flow of different constituents using a pressure driven flow. The pressure is created by the heart, by the by within the human vasculature system, because it is a pressure driven flow, there is a tendency of the Leukocytes to marginate by virtue of their sizes and go more towards the edges of the vasculature in comparison to the red blood cells, which get accumulated more towards the center.

If I can actually biometrically represent this in terms of a micro-channel, let us say like this. Then, we pull out different branches of these micro-channels, let us say, this is the branch, which we are pulling out here at this particular end. You have to ensure though that this has a size base selection, it only selects the size of the nuclear Leukocytes; so the size of this channel is pretty much that of the Leukocytes.

What would happen is that these Leukocytes, which have already marginated, you can see these by white cells here. They are more abundant towards the site, somewhere in the center. Of course, the red blood cells are more abundant towards the center. You can actually pull this out by designing a channel, which is of the same size as this leukocyte is. So, you can make leukocyte rich samples by this kind of a separator.

There are lots of uses for this technology. Normally, Leukocytes are separated from the human blood by process called a Buffy coat process, where you use a density gradient causing agent like if you call plaque to make a density or different sizes of the cells

within a solution. This is essay, which can actually replicate that Buffy coat process, which is too intensive, both in terms of labour, as well as in terms of contamination prone. Here, with a small assay, you could have a rapid throughput, where you can get really a leucocyte rich sample towards the end collection here, in this particular case of the smaller channel, you could investigate that for further purposes.

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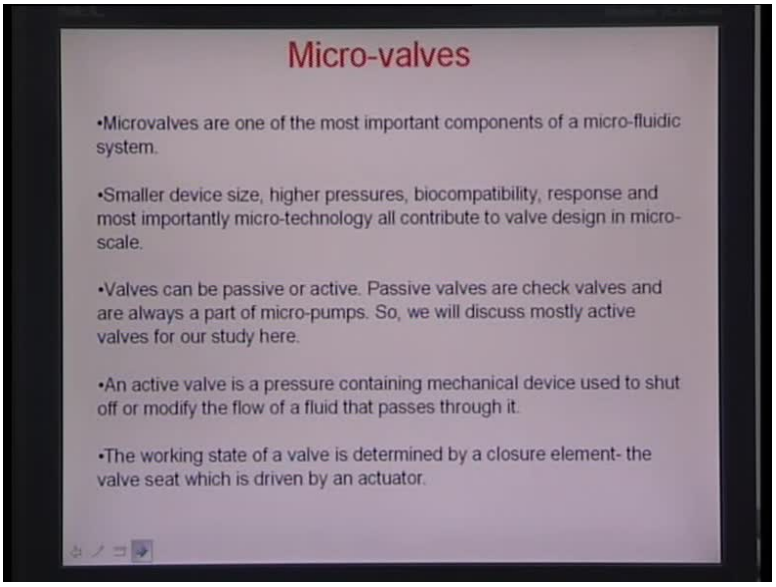

These are some of the wonders that micro-fluidics can do to different systems and processes. I would like to now actually go into a little different domain of micro-valves, when we are talking about micro-fluidics; a very important mechanism that we must ascertain is control of fluidic flow. For doing that we need to design a different valving systems, which can actually stop or block or gate the flows running, past this small channels or capillaries. Also, would be able to meter or control the flows in the flow rates that we desire in certain sections of a particular microchip.

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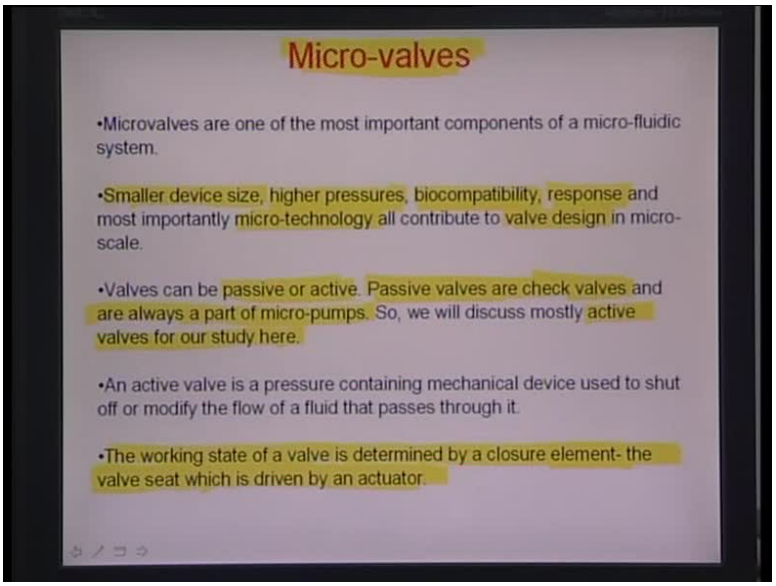

Micro-valves are definitely used for those purposes; they have one of the most important components of a micro-fluidic system. What all is important for valve designing, you have to consider the size of the device, MEMS devices are very small. You have to consider pressures, which are very high that is why most of the pressure sources sometimes are off chip, the mechanism of valving involves energy sources or pressure sources, which are not on the same chip.

We should be aware of biocompatibility issues, if you are designing valve materials or choosing valve materials. The valve should have a proper response to the flow processes, are most importantly micro-technology and should be used for fabricating such micro-valves, particularly for using or gating the flows in micro-scale.

Valves can again be dependent on the mechanism of their closing or working can be classified into passive or active micro-valves. Passive valves can be like check valves, where by virtue of a change in property of a material, maybe it swells and blocks - the valve swells and blocks or alternately, by motion of fluid, let us say, you have a one directional valve. The pressure generated from one side to another, the valve closes one side automatically and opens when this pressure reverses back to the other side, so they are passive valves. Where there is no mechanical energy or any other form of energy which are used for operating these valves, so they are like check valves.

Passive valves are normally a part of micro-pumps. We will study this micro-pumps topic in more detail later. Active valves on the other hand, are essentially those which are dominated by some kind of energy input to the valving system. An active valve is a pressure containing mechanical device, normally used to modify the flow or stop the flow or seize the flow. Although, it can be mechanical as well as non-mechanical, there may be instances where you can use electronically to close a particular channel or electro chemically to close or gate a particular channel. We will actually consider some of these illustrations and designs later on, where we design these different valving techniques.

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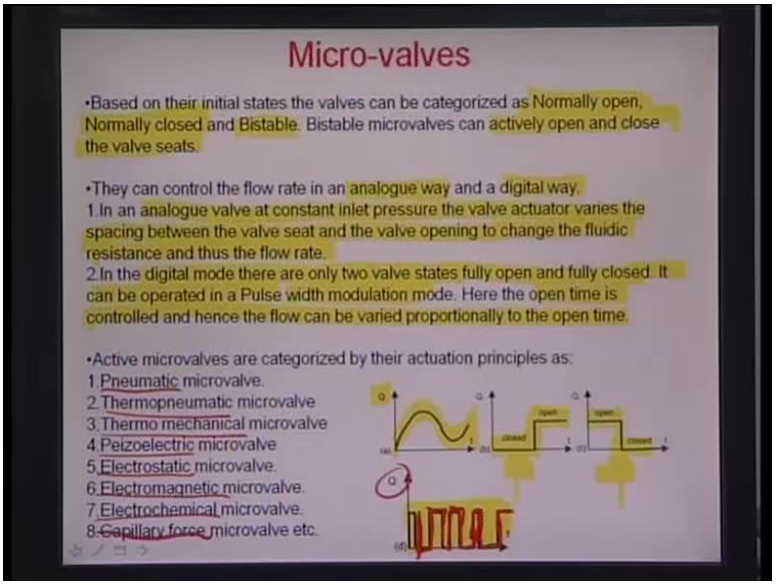

The working state of a valve really, as you can see here, is determined by a closure element, the valve seat over which this closure element would typically set for the valving to be on; this is driven by an actuator. So that is how the valving process can be done, in case of micro-valves. Some of these micro-valves again can be based on what their initial states are, whether they are normally open valves or normally closed valves, can be classified into these two subtypes or bistable valve, which can close and open actively their valve seats.

You have three categories here, normally closed, where valve normally after actuation opens up - normally open, where the valve actually is normally open, but it closes on actuation; bistable, which can also - which can actually close and open both, on some active energy being pumped in the ball valving system, the valve seats. Valves can

further control flows, either in an analog manner or a digital manner, which means digital is just an often kind of mechanism; you have the fluid flowing at one instance and not flowing at another instance, whereas, in analog mechanisms, it is actually a slow closure of the valve, slow decrease in the flow rate across a valve and vice versa.

You can categories them into an analog way and a digital way. As I illustrated that in an analog valve, at constant inlet pressure, the valve actuator varies the spacing between the valve seat and the valve opening. This changes the fluidic resistance and thus the flow rates, so it is analog control on the rates. In the digital mode, however there are only two valve states, fully open and fully closed. It can be operated in a pulse, actually with pulse width modulation technology. Here, the open time is controlled, hence the flow can varied proportionally toward the open time is.

Essentially, if you look at some of these output responses, this is again what an analog valve would typically look like. This is discharge rate versus time plot of that is with time. As you see, as the valve is fully opened, the flow rate is maximum, now closes in the flow rate goes down slowly, but then there is an analog response.

There is a slow convergence of the flow rate to a maximum and a slow convergence of the flow rate again to a minimum, it is not rapid one shot close and open phenomena. This is the digital micro-valve again, where you have only two states, the close and the open. Then the interconnection between this is really very small in times. So, you do not need much opening time or closing time, it automatically either opens or closes.

Here, this is a normally open system; this is a normally closed system. By pulse width modulation what you mean is, you can actually use this technology to vary the flow according to your own interest. If you have more open time here, as you are seeing, this small loop that this discharge is formulating is the open time. This is a smaller open time, this is slightly larger open time and this is the largest open time. Similarly, these are the closure times, the time for which the valves are closed. You can see there are all different times of closures (Refer Slide Time: 29:09).

Pulse width modulation basically would mean that if you have the pulse width to be more, in this particular case, you have the flow going on for a longer amount of time. Pulse width modulation effectively therefore means that if you increase the open time,

the flow rate would be more, if you reduce the open time, the flow rate would be less. That way you can have a differential flow rate Q with respect to T.

Active micro valves again characterize on the basis of their actuation principles. The actuation actually means the way that you are delivering the energy, you can actuate a micro-valve pneumatically and you can do that thermo mechanically. You can also use a peizo electric crystal to actuate micro-valves; you could use electro static method for micro-valve actuation, electromagnetic, electro chemical, capillary force, surface tension. All different kind of forms can be used for actually closing and opening a valve system.

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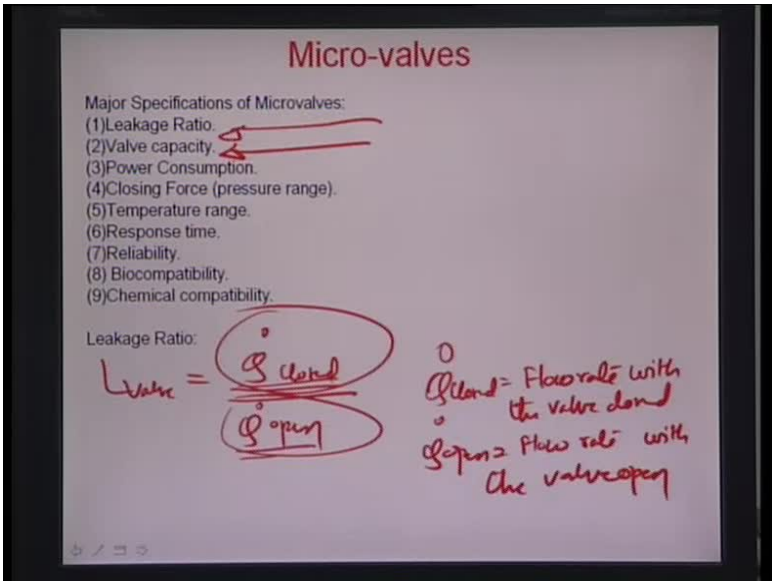

If we look at some aspects of micro-valves, the way that you have to specify or characterize the systems, you have different parameters, which you use for finding out the performance criteria's, of such valves. One is leakage ratio, so the leakage ratio essentially can be represented as the flow rate of the closed system divided by the flow rate of the open system. This is the fully open mode, what is the flow rate? This is the fully closed mode what is the flow rate? So, this is the ratio between what is the flow rate at the closed mode in respect to what is the flow rate at the open mode.

Basically, this represents, what kind of leakage is the valve would have. This ratio is more than 0; that means definitely in the closed mode, the valve is leaky. There is some flow, which can be a fraction of the fully open flow and that fraction effectively cannot be done away with, the valve would still be in its fullest operating on a more stringent

operating condition, would be able to lead this flow passed it or through it. So that is what leakage ratio would mean.

This Q dot closed is flow rate with the valve closed and Q dot open is the flow rate with the valve. The pressure head that is actually driving the flow, through this valve or the under root of or the square root of the pressure head that is driving through this particular valve, let us illustrate this a little properly. Therefore, valve capacity C of the valve is defined as Q star maximum, this is the maximum flow rate divided by root of delta P maximum, which is the pressure difference across the valve divided by rho g; rho is the density, g is the acceleration due to gravity.

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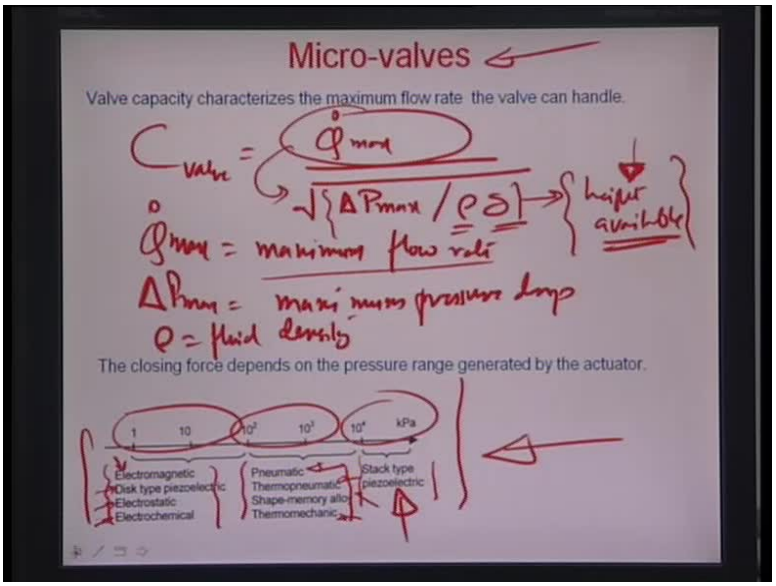

Therefore, this really is indication of what is the pressure height, what is the height available causing the pressure gradient; what is the height of pressure that is available across both ends of the valve. The ratio between the maximum flow rate times the root of this particular available height is what defines the capacity of the valves.

Therefore, if the pressure head that can be which stood by this system is more, the valve capacity is lower and vice versa. If the pressure head that can be withstand by this system is higher, the valve capacity is higher. The closing force depends on the pressure range generated by the particular actuator sequence. Here, in this particular illustration, you can see some of the different actuation mechanisms and the way that the pressure range can be generated by different actuator mechanisms.

Here, for example, let me just write these things down, Q dot max is maximum flow rate, delta P max is maximum pressure drop, rho is the fluid density, g of course, as you know is the acceleration due to gravity. For electromagnetic, disk type is piezoelectric, electrostatic and electrochemical actuation, the pressure ranges that can be generated by these actuator in roughly about 1 to 10 kilo Pascal's. If you go a little bit higher to pneumatic thermopneumatic shape memory alloy based on thermomechanical actuators, the pressure ranges would be in the range of about 100 to 1000 kilo Pascal's. The highest pressure range that can be actuated is given by the stack type peizo-electric crystal. There are multiple peizo-electrics, which are stacked one on another; the effective pressure that is felt is a result of series pressure of all these different peizo-crystals taken together.

Here, the actuation pressure can be as high as about 10,000 kilo Pascal's. So that is what essentially the valve capacity would characterize. Therefore, what it means is that the maximum flow rate, which it can hold per unit pressure head available, is for the capacity. Therefore, if suppose the pressure head is fixed, you have a valve, which can hold, let us say, 1 centimeter cube per minute of flow rate, another which can hold 10 times this value, so 10 centimeter cube per minute flow rate.

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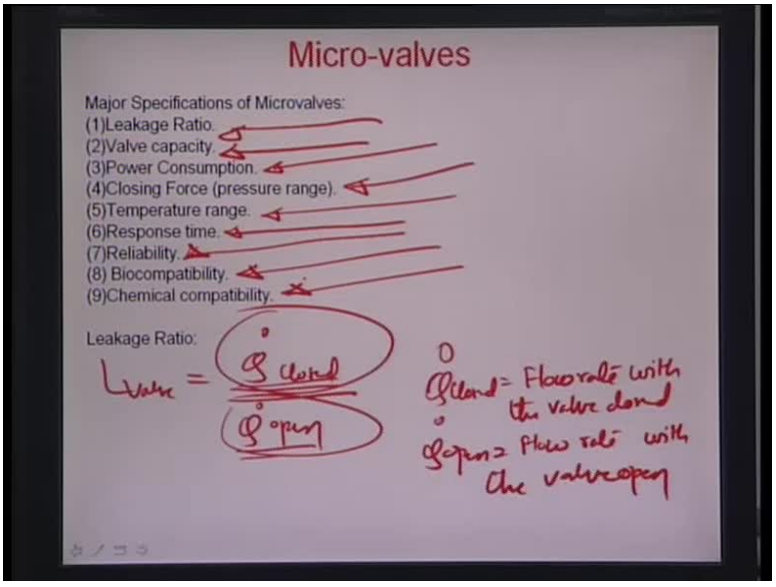

For a similar kind of pressure head on both ends, you know C valve - the valve capacity would be much more in case of the 10 centimeter cube per minute flow closing valve, in comparison to the 1 centimeter cube per minute flow closing valve. So that is what valve

capacity is characterized as. There are some other performance factors, which include what kind of power is consumed by the system, what are the closing forces in terms of pressure ranges, which you need for the valve to fully close. What kind of temperature ranges, you can use the valve; so these are some of the specifications if you design such valves.

What is the response time? It is a big criterion, how quickly the valve can open and close particular flow? Is there any reliability issue or the valve operation is perfectly unique or not every time or repetitive or not every time is what a very important design paradine is. Biocompatibility aspects itself, the material is a very major concept, which is important for designing or selection of materials, which would make these micro-valves. Then, finally, chemical compatibility again is a highly important parameter for designing specifications of these micro-valves.

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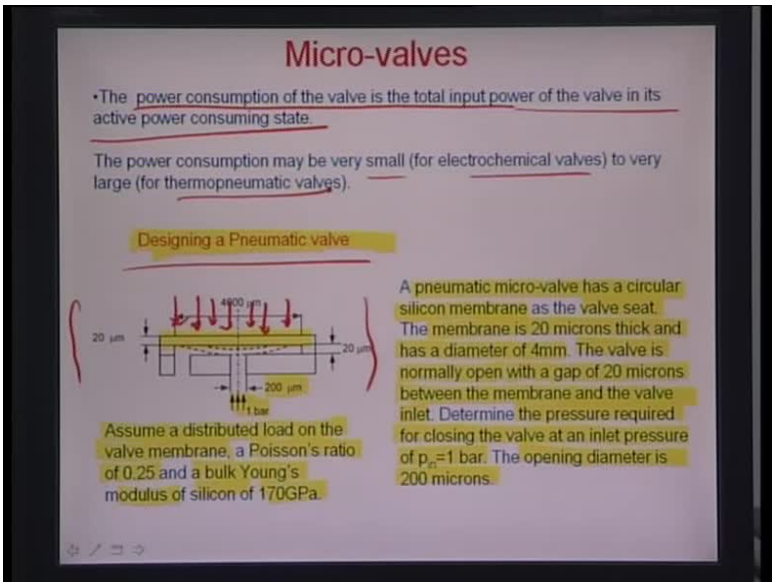

Let us do some real examples of valve designing, to see a little more of how this can really be designed for flow systems. The power consumption of the valve really is the total input power of the valve in its active power consuming state. It may be small for electrochemical valves, very large for thermopneumatic valves depending on what the actuation mechanism is and how things can be actuated.

In this particular example here, we want to design a pneumatic micro-valve system. Let us now solve this example of designing a pneumatic valve system. We have a pneumatic

micro-valve here, which has a circular silicon membrane. This right here is a membrane, which is circular as seen in the top view, this is a side view - the cross-sectional side view actually. The silicon membrane essentially is the valve seat in this particular valve, which means that the silicon membrane bends and deforms like this, it blocks the particular valve.

The valve seat is actually just below this membrane, so normally it is flat; it bends and blocks the entry of the air flow into the system. The membrane is about 20 microns thick; it has a diameter of about 4mm 4000 microns. The valve is normally open, so there is no pressure over this particular region here, it is as if the valve is open at the very outside. The gap in that particular case is about another 20 microns, between the membrane and the valve inlet.

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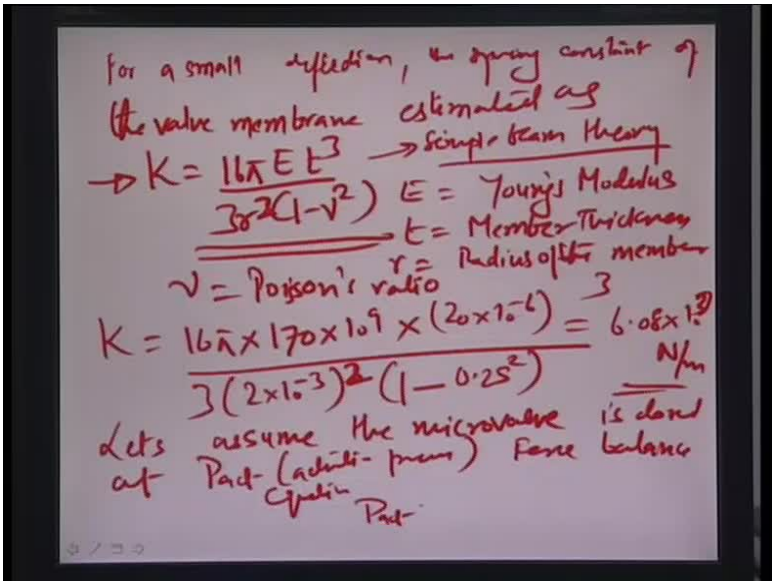

You have to determine the pressure required for closing the valve, at an inlet pressure of about 1 bar, as you can see here. The opening diameter is 200 microns, which is right here. We have to assume that the load is distributed on the valve membrane. This Poisson's ratio of about 0.25, there is a Bulk Young's modulus of silicon, which is about 170 giga Pascal's, so we have to design the system.

Let us actually look at how we will design this. For a small deflection, the spring constant of the valve membrane can be estimated as K equal to 16 pi E cube by 3 squares minus 1 minus nu square. Now, this comes from a simple beam theory. Any standard

mechanics of solids textbook would be able to demonstrate, how the spring constant in case of thin beams can be reported.

The various parameters are E, is the Young's modulus, t is the member thickness; the member which is deforming and is really the valve, r is the radius of the member, nu is the Poisson's ratio. You have pretty much everything among this, so let us calculate what the K value would be, its equal to 16 pi times of E, which is 170 10 to the power of 9. The Young's modulus of silicon is 170 giga Pascal's, so his Pascal's times of thickness cube, so it is 20 10 to the power minus 6, 20 micron is the thickness of the particular member in question.

So, cube of thickness divided by 3 r cube, r is basically 4000 microns or 2000 microns, so this is about 2 10 to the power of minus 3 meters cube of this times of 1 minus, sorry square of this, times of 1 minus nu square, nu is 0.25 Poisson's ratio. This comes out to be effectively equal to 6.08 10 to the power of 3 Newton per meter that is what the K value is.

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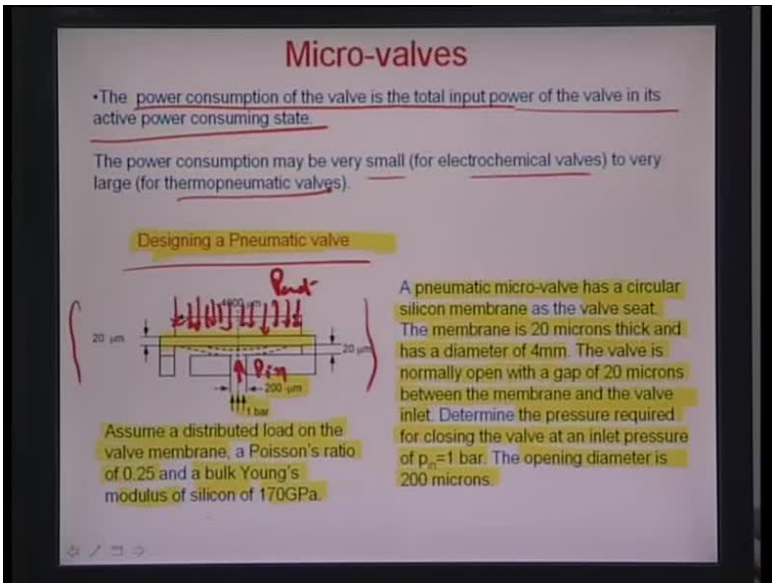

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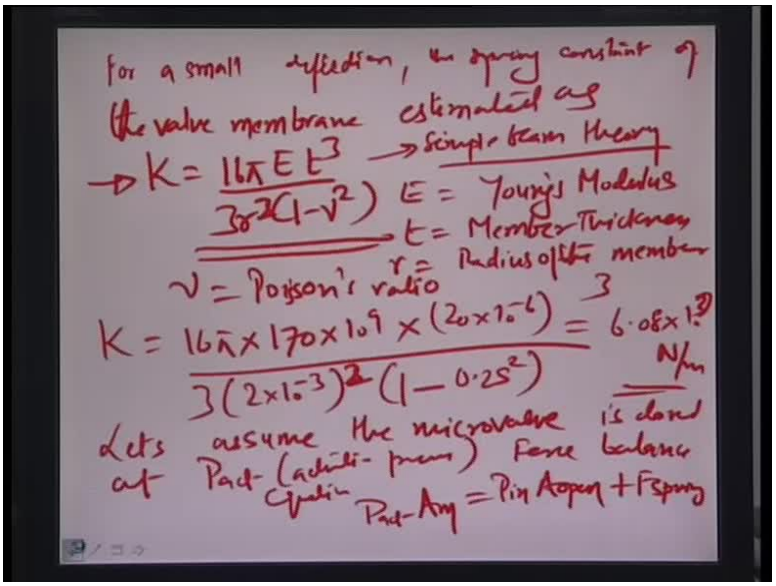

Let us assume that the micro-valve is closed at an actuation pressure of P act. Let us assume the micro valve is closed at P act actuation pressure. So, the force balance equation for this valve, because of the distributed load is basically the actuation pressure, which is actually from this side as you see here, this is the P actuation, this is the P inlet. You are left with P actuation times of area of membrane times of P inlet times of area of the open system.

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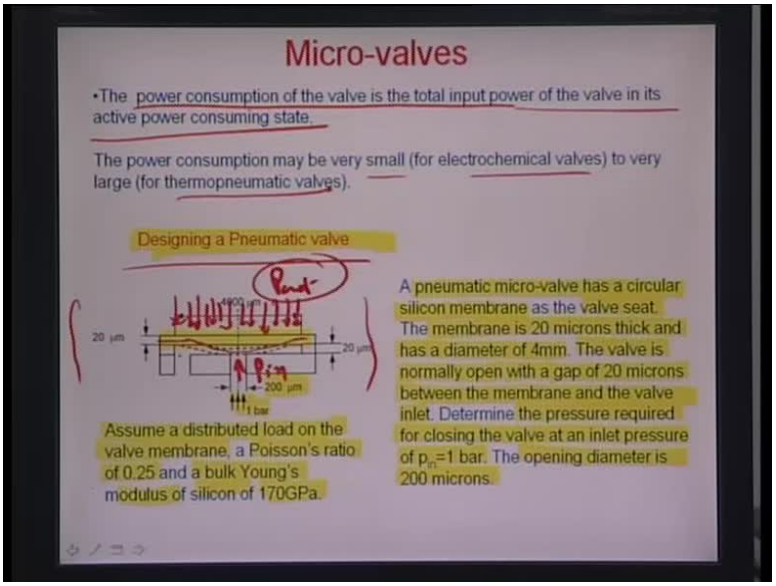

Essentially, in this case, the open system is nothing but the area of the membrane itself again. So, P input times of A open plus the spring force. Mind you, the pressure here is actually being resisted by this downwards spring like movement of this valve. So, whatever force this P input would have on the membrane plus the force spring is essentially equal to the actuation pressure times the area of the particular membrane. So, this is essentially at equilibrium.

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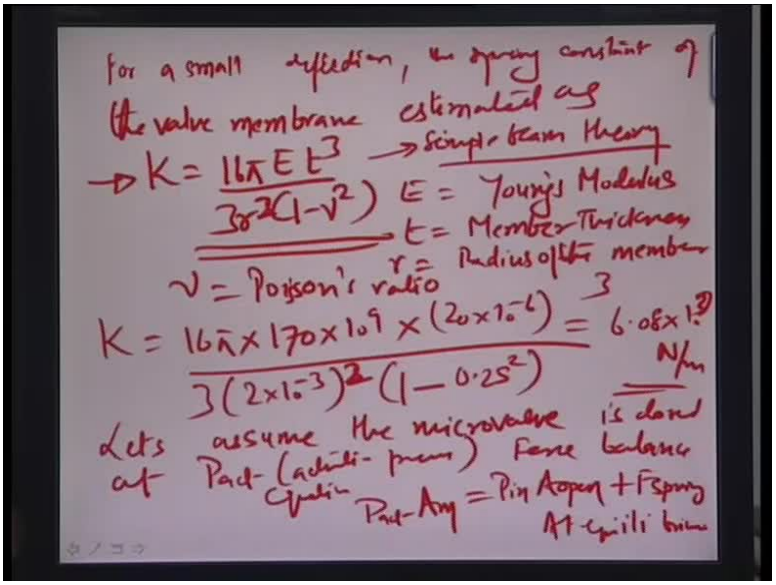

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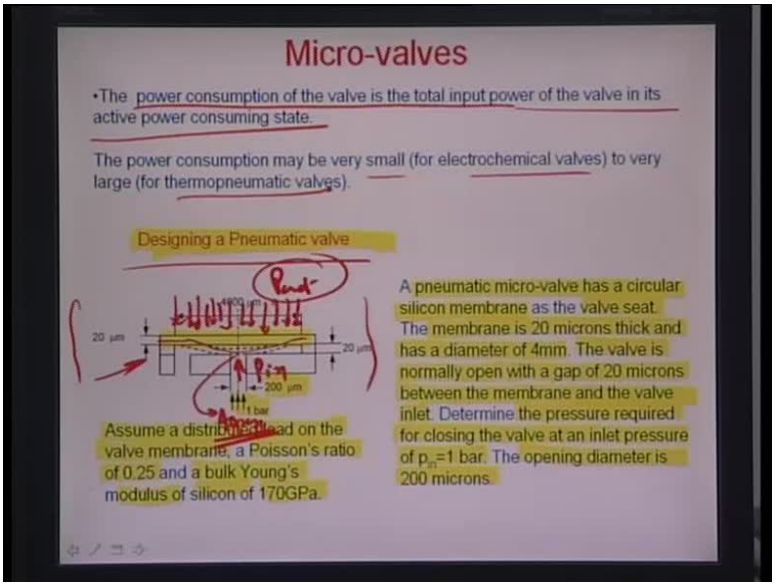

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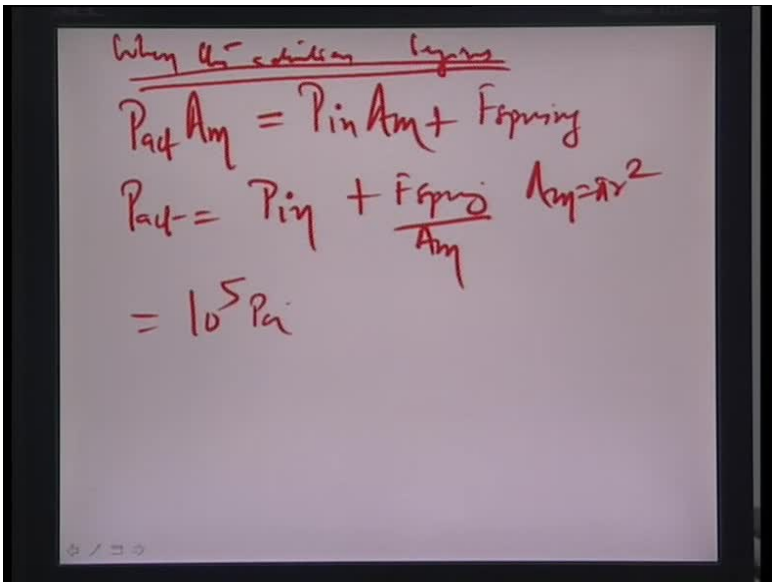

This right here again, is the A open, the area of cross section which is open. Equilibrium means that the membrane is fully seeded over the inlet hole, thus valving the flow. A open is the area of the inlet hole, P input is the inlet pressure and that plus the spring force, is essentially equal to the force by the actuation pressure on the other side of the membrane. Just about when the actuation begins, this equation slightly modifies, so you have now P act times of m on one side, which is also equal to P in times of A m in the same membrane, because it is in the open position plus the F spring; spring force is still there, when actuation just about begins sorry already written this thing back on the top when the actuation just about begins, as you can see here (Refer Slide Time: 44:58).

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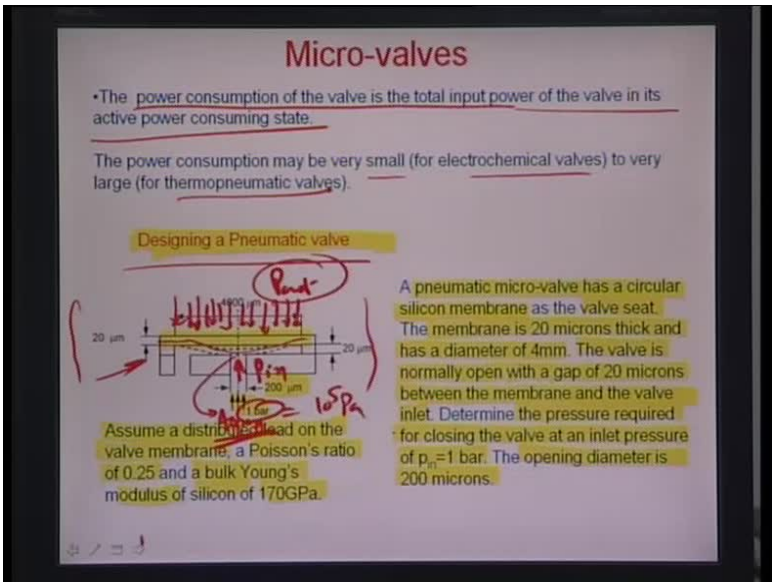

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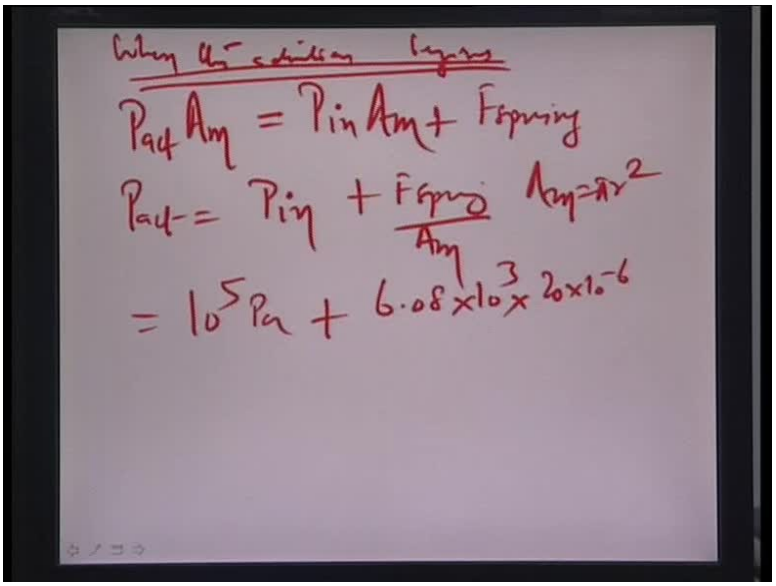

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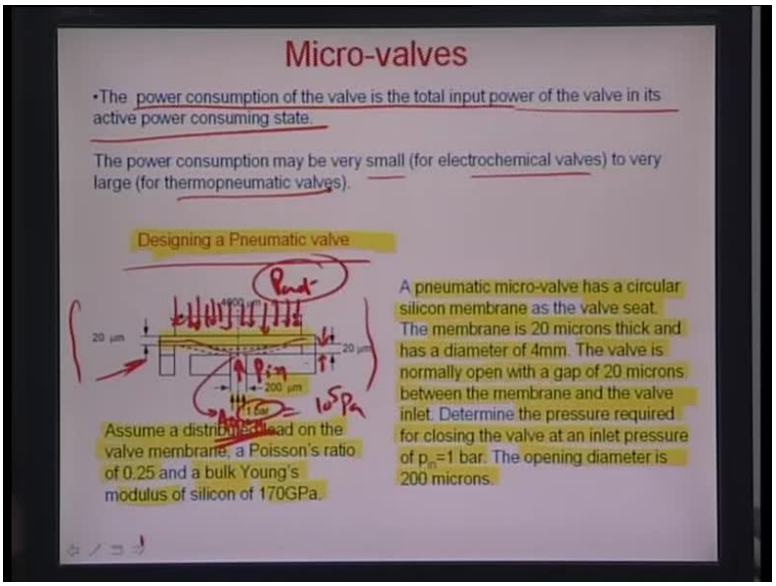

Let us actually calculate some of these values, let us see, what happens when the actuation begins. You have P actuation; actuation pressure needed is P inlet plus F spring times 1 by A m. A m essentially is pi r square, r is about 2000 micrometers, therefore the actuation pressure would be 10 to the power of 5 Pascal's. So, the pressure in the inlet side is about close to 1 bar, which is nothing but 10 to the power of 5 Pascal's plus the force spring, which is 6.08 to 10 the power 3 times of the deflection, which is again about 20 10 to the power of minus 6 that is essentially the 20 micron displacement that would happen here as you see.

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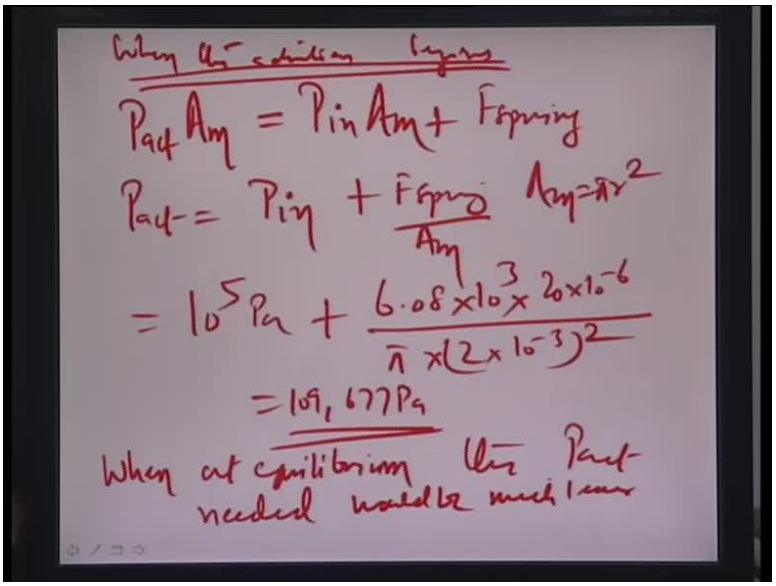

Let us assume that the port has not yet blocked and the pressure is released onto the entire lower surface of the seat membrane, this divided by A m, which is pi r square, r is 2 10 to the power of minus 3 square. So, this value comes out to be about 109677 Pascal's.

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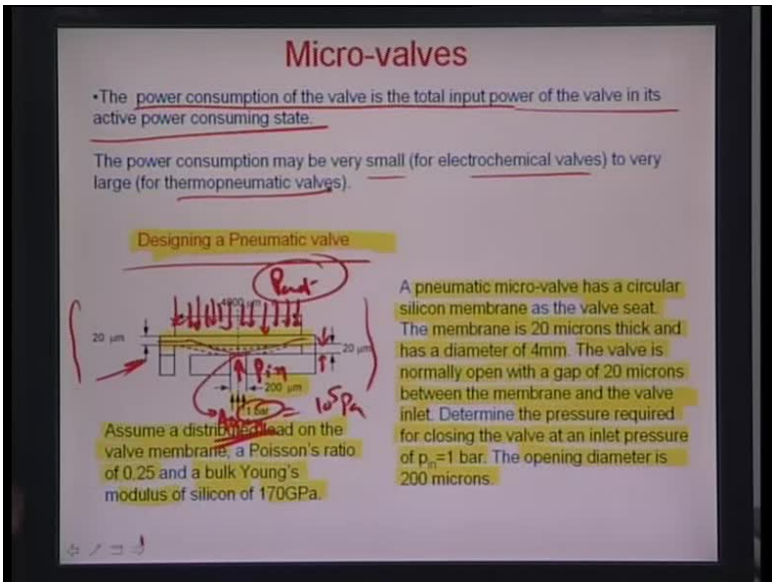

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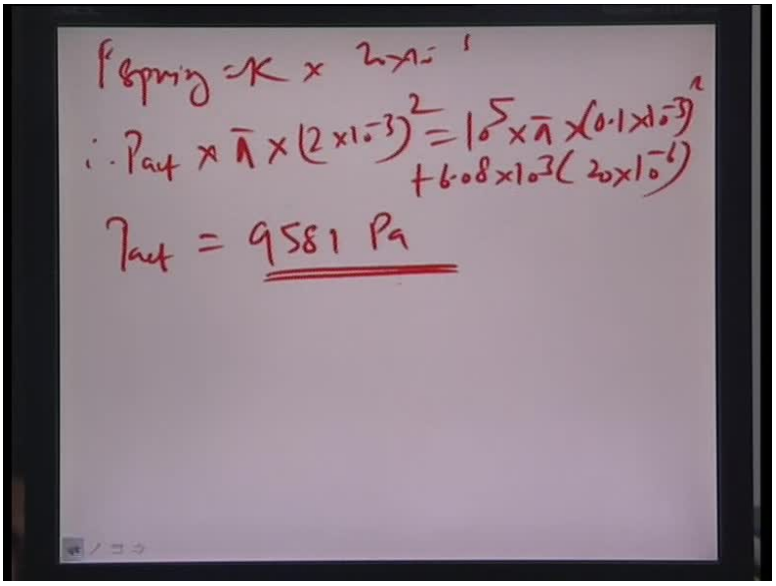

Now, when at equilibrium, though entirely this paradine changes. When at equilibrium, the P actuation needed would be much lesser, more so because, now the active area over which this 1 bar pressure is available, is only this little area here that is the P, the area of

the opening. Let us see, what that would be force spring in this particular case, is again K times of 20 10 to the power of minus 6. P actuation times of pi times of 2 into 10 to the power minus 3 square, would effectively V equal to 10 to the power of 5 times of phi times of 0.1 10 to the power of minus 3 square plus 6.08 10 to the power 3 times of 20 times of 10 to the power of minus 6. P actuation here would come out to be effectively equal to 3.1; let me just write down the final value here, so it comes out to be about 9581 Pascal's.

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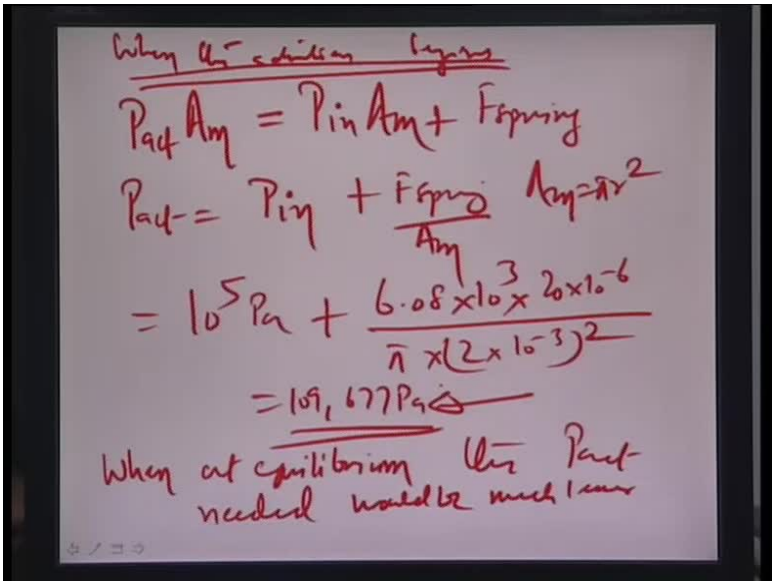

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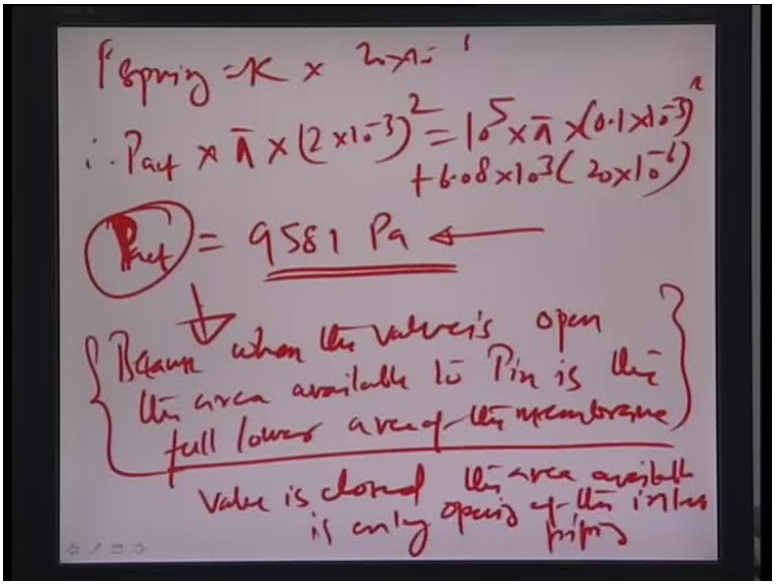

In comparison to the earlier value here, which was about close to 109677 Pascal's, this has reduced to almost two orders of magnitude and become about 9581 Pascal's.

Why that is so again, because when the valve is open, the area of membrane or the area available to the inlet pressure is the full lower area of the membrane. When the valve is closed on the area available, is only the inlet or the opening of the inlet piping. So that is probably the reason why the actuation pressure in the second case, where the valve is closed, is much lower in comparison to the actuation pressure in the uppercase.

This is though an advantage, because being at the micro-scale, the area of cross section of such valves is very small. So, effectively the amount of force that is generated by the inlet side design, which is also pressured into area by virtue of the area being very small, remains small and it necessitates a lower amount of actuation pressures. Therefore, in case of pneumatic valves, in this kind of configuration, it is always desired to make the inlet side opening as small as possible, which can rime with the micro-systems and that can help us to have lower amount of actuation pressures to control the valving action.

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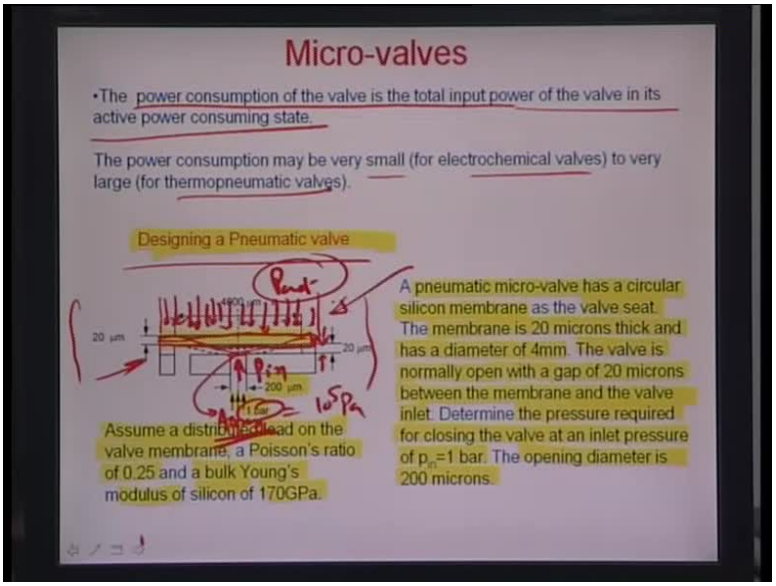

If you design the same valve with the thermopneumatic perspective; that means, whatever this load distributed load here, was obtained on this particular membrane, is done through heating some kind of a gas entrapped over the membrane, so that is called thermopneumatic system. Therefore, on a fixed volume, if you actually keep on adding electrical work or energy, there is an expansion of the gas which would cause a pressure

to develop. The pressure would be good enough for the membrane to bend, the valve, the inflow of the air. So, those systems are known as typically thermopneumatic systems.

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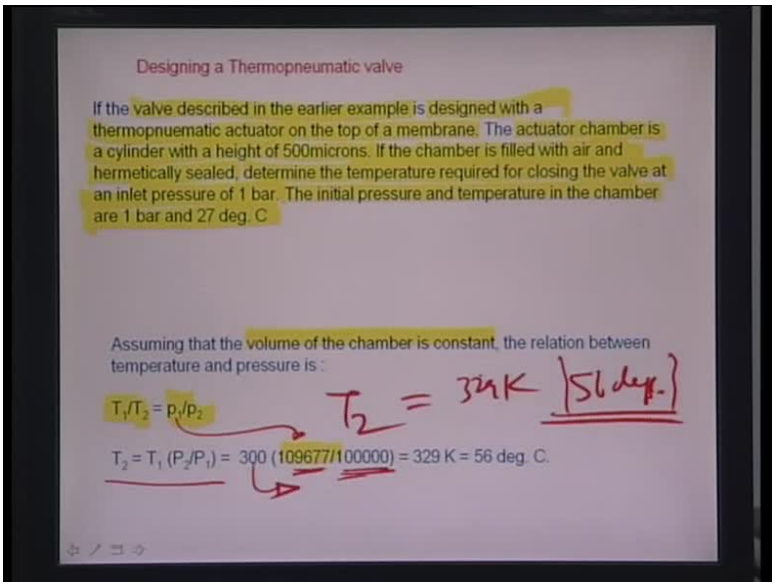

For a similar kind of arrangement, let us see, what would be the amount of temperature that is needed to be reached for sufficient amount of actuation pressure to be generated, so that the closing action of the valve takes place. Let us suppose, you have the valve described in the earlier example, this is designed with the thermopneumatic actuator on the top of the membrane. It is given that the actuator chamber is cylinder with a height of about 500 microns. This is sitting on a diameter of about 400 microns, which is also the outside diameter of the valve, the silicon membrane valve, which is in consideration right now.

If the chamber is filled with air and pneumatically sealed, we have to determine the temperature required for closing the valve at an inlet pressure about 1 bar. The initial pressure and temperatures was in the chamber of 1 bar and 27 degree Celsius. Again, we use the Charles law on a fixed volume; you have pressure proportional to the temperature. Therefore, T 1 by T 2 is temperature in the beginning, by temperature after the expansion, is taken place is pressure 1 by pressure 2.

Assuming the volume of the chamber is constant. Here, the inlet pressure is about 1 bar, we found out that the amount of actuation pressure that is needed from the opening to the closing position or opening position is about 109677 Pascal's. So, this is the P 1, which

is needed. The inlet pressure P 2 is about 1 bar, which are about 100000. Therefore, if the inlet pressure 1 bar is at 27 degree celsius temperature, which is about corresponding to about 300 Kelvin.

We find out from this relationship that T 2 - the temperature required for the pressure to be 109677 Pascal's, is about 329 Kelvin, which is close to 56 degree Celsius. Therefore, you have to really heat, the constant volume on the cylinder with a height of 500 microns, by to almost about 56 degrees for the valve to selfclose using thermopneumatic means. So that is how you design a thermopneumatic valving system.

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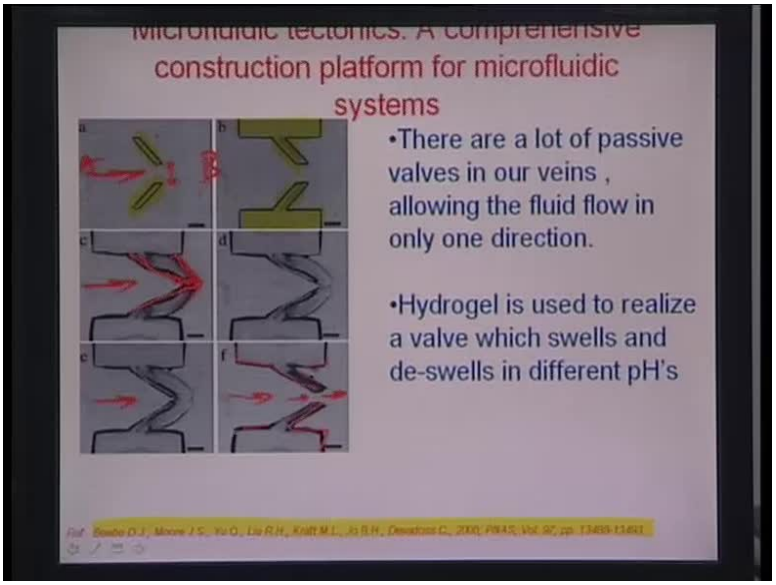

They can be also passive valves as I recorded in my last lecture. These passive valves essentially are based on either the expansion of a structure, blocking of flow because of this expansion, etcetera. These valves are very beautifully illustrated by Beebe et al in this particular paper, appeared in PNAS, where used hydrogels. Essentially, as you see here, these hydrogels are fixed to certain fixed members in a flow channel, so direction of flow being from direction a to direction b through this channel.

Typically, these are unexpanded and you have a gap in between here, from where the flow can go from a side to b side, but hydrogels are essentially things which expander or swell on different PH's, particularly when the PH is more like acidic, the valves would swell in volume, so it will be expand. If we have these valves expanding in volume, they become this big by virtue of their size, they can block the flow. Now, the flow is blocked

because these hydrogels of swelled, they are blocking the flow direction in this particular channel.

As the result of which, these valves can be very good PH sensors, so small change in PH can result in swelling and flow stoppage of the particular valve. You can calibrate the change of PH to the volume of swelling or the volume difference that the valve would make by swelling. Therefore, the volume of flow, past the valve, can be calibrated to the corresponding PH.

Here, as you are seeing, you can reversibly change the orientation by flowing back a basic PH, so that the valve shrinks and gives way or opens the flow. These were being made up of hydrogel and soft materials, they can be biomimetically patterned, they are based on some of the flow valves that our heart probably possessors their expansion and contraction keeps on, either confining the blood onto the chamber of the heart or releasing the blood at a certain pressure.

This brings us to an end of this particular lecture. In next lecture, we are going to talk about some other value mechanisms, particularly non-conventional mechanism, like electrochemical or surface driven valving or flow closing mechanisms. So, I would like to close this lecture, thank you.