

THE PHACO FLUIDICS BOOK.

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FORWARD:

Within this book the determinants of Phaco-Emulsification Machine Fluidics are examined and analysed. This analysis is based on basic Mathematical and Physical Science and is also supported by experimental investigations to verify that the physical models are accurate.

Pressure within the anterior chamber is defined for *steady flow* conditions for both the Peristaltic and Venturi/Vacuum based machines. Also anterior chamber pressures, under *dynamic* conditions, such as *flow occlusions* and the *post occlusion surge* are also examined and defined. It will be shown that the mathematics and physics of these events are those of *damped simple harmonic motion*.

A new model has been created by the author to explain the unique properties of Phaco Fluidics. This analytical system helps to identify the true determinants of the capacity of irrigating instruments and the properties of the aspiration systems in our phaco machinery. This analysis leads to new insights on how to improve anterior chamber pressure stability during surgery and therefore improve anterior chamber geometry stability and in turn improve surgical safety.

It transpires that the entire phaco fluidics system may be emulated or modelled very accurately by using *equivalent flow circuits* which take *flow resistance, compliance* and fluid *inertia* into account. These three features determine the overall fluidics behaviour of phaco machine fluidic systems, rather than the properties of the machine microprocessors and software as sometimes claimed. Also incorporated for the first time in the study of phaco fluidics is the concept of the aspiration tube as a fluidic *transmission line* with a *characteristic velocity* and features of *transmitted* and *reflected pressure waves* and *flow delay* phenomena. This is critical in the understanding of the events which occur in the eye during surgery and also determines the delay between the occurrence of fluidic events within the eye and any data relating to those being available at the machine's sensors for any compensatory actions by the machinery.

The model created readily generates graphs which are printed to show the anterior chamber pressure and the anterior chamber inflow and outflow rates during fluidic events which occur during surgery as well as the vacuums present at the machine's sensor. Also an analysis of *lens fragment holding forces* in the Vacuum/Venturi and Peristaltic phaco machine system is included. This has been made possible for the first time by the use of equivalent flow circuits, mathematical modelling and the deployment of flow circuit simulator software.

Finally, practical advice is given for safe settings for the surgeon when using existing phaco machines. The ultimate goal of understanding phaco fluidics is improved surgical safety during phaco-emulsification cataract surgery and the total avoidance of any tissue trauma such as posterior capsule rupture.

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CHAPTER 1.

(1-1) OVERVIEW AND INTRODUCTION TO THE FIELD OF PHACO FLUIDICS.

An understanding the fluidics of the phaco machine and its fluid management systems will help the Surgeon get the best out of the machine in practice. An understanding of the machine's fluidics helps to reduce surgical complications and improve surgical outcomes because the Surgeon can tailor the machine settings to suit the individual eye and identify the reasons for chamber instability when it occurs.

Anterior chamber geometry instability can lead to corneal or iris damage and posterior capsular rupture. This may lead to secondary complications which increase the probability of retinal detachment, macula oedema, glaucoma and result in a less predictable refractive outcome. The Phaco Machine's settings for both the fluidics and the ultrasound energy need to be individualised to suit each eye. It is the Surgeon's responsibility to understand the implications of the machine settings and adjust them to suit the particular eye and not merely to rely on "standard settings" which might only be suitable for the average eye.

Due to the variability of individual eyes and orbits, a surgeon may get the impression that one type of machine or instrumentation set might perform better another, when this might not be the case averaged over a very large series of eyes. In addition a more experienced surgeon can cope with a higher degree of anterior chamber instability than junior surgeon. Variables such as the compliance of the individual's eye, the anterior chamber depth for a given anterior chamber pressure, wound leakage and external globe pressure all result in a "spread" of anterior chamber geometries and stability, all other things being equal.

The ideal way to compare our Phaco Machine's performance is with laboratory testing under controlled conditions with model eyes and standard materials to emulsify. For fluidic analysis this is a very straightforward process and involves placing the probe in a test chamber with a pressure sensor. The chamber must have the same average compliance properties as the human eye. This way the various machines can be objectively compared for fluidic performance under identical test conditions. However with accurate enough modelling, as will be shown, this can also be done in "computer space". This requires that the geometries of all the fluidic system objects and the compliance of the soft materials measured and known.

There is a significant difference in the behaviour of the vacuum (venturi or electric vacuum pump) based Phaco Machine, compared with *flow* (peristaltic pump) based machines. These differences will be examined, as each type has its own advantages and disadvantages as the technology stands currently.

There are generally two distinct types of fluid flow situations to consider in phaco fluidics, which may exist independently or be combined at times. These are constant or *static* flow situations and *dynamic* or transiently changing flow situations. In the *static* or stable flow case there is uninterrupted or continuous flow via the fluidic system and the pressure gradient driving the flow is in equilibrium with the forces of fluidic friction. In this instance, in the *peristaltic machine*, the anterior chamber pressure is easily determined from the bottle height, flow rate and irrigation flow resistance.

In the case of the *vacuum machine* the sum of the bottle pressure and vacuum applied to the total system flow resistances determine the flow rate and again the flow rate and irrigation resistance and bottle height determine the anterior chamber pressure. The method explaining how to calculate this will be given for both types of machine.

A point of interest in both types of machine is that the anterior chamber pressure in a *static* flow situation depends only on the applied pressures and flow resistances alone. On the other hand when the flow is stopping and starting in *dynamic* conditions, the eye pressure depends not only on the driving pressures and flow resistances but also two other factors. These two factors are firstly the *compliance* (springy nature) of the elastic structures. These include the eye itself, the plastic parts of the disposables, the soft parts of the peristaltic pump and the vacuum sensor area at the machine and the cassette air in the venturi machine. Secondly, the inertia (reluctance of a mass to change its existing velocity) of the fluid within the fluidics system which is being accelerated and decelerated at times. In addition the moving fluid has "stored energy" in the form of kinetic energy. Also the elastic structures store "elastic energy" analogous to a spring. The stored energy in both cases is a major determinant of the behaviour of the fluidics system under dynamically changing flow conditions such as occlusion make (an occlusion forms in the phaco needle entry area) and occlusion break (the occlusion clears and flow is re-established). Under these occlusion make and break circumstances the flow is obstructed and subsequently clears later resulting in anterior chamber pressure and therefore anterior chamber geometry fluctuations.

There are fundamentally 5 basic sources of anterior chamber geometry variability in phaco, and these are related to the 4 characteristic fluid flow variations with respect to the inflow limitations.

The four flow variations are related to:

- 1) Aspiration flow.
- 2) Wound leakage flow.
- 3) Occlusion make flow interruption.
- 4) Occlusion break flow resumes.

In addition to this there may also be partial and variable degrees of occlusion which elevate the flow resistance in the aspiration pathway in the entry area of the phaco needle.

The aspiration flow rate is machine pump driven in the *peristaltic* machine. This is regulated by the peristaltic pump rotation speed up to the maximum flow rate the surgeon's has set on the

machine's control panel. In the *vacuum/venturi* machine the aspiration flow rate is driven by the combination of cassette vacuum and bottle height applied across the total fluidic system flow resistance. Wound leakage flow is variable and depends on the anterior chamber pressure and the geometry of the wounds and instruments and how they are manipulated within the wounds.

Flow occlusions, by substantial blockage of the phaco needle with lens fragments, result in increases in anterior chamber pressure toward the bottle pressure over time. This can result in overall unstable and fluctuating chamber geometry when the anterior chamber pressure without occlusion is relatively low at high flow rates prior to occlusion. The differences between the occluded and un-occluded state can then represent large variations in anterior chamber pressure and geometry. This is a problem seen in *venturi/vacuum* machines where the flow rates are often high and unknown to the surgeon.

On the other hand the driving force for post occlusion surge flow in the *peristaltic* machine is unrelated to the peristaltic pump's flow rate and depends, as it will be shown, on a combination of factors which include the ratios of the compliance of the disposables on the aspiration side, the compliance of the eye, the aspiration system flow resistances, the fluid inertia in the aspiration system and the sum of the bottle pressure and the vacuum within the aspiration system prior to the occlusion break. The energy source driving the post occlusion surge in the peristaltic machine can be regarded as energy stored in the elastic structures of the aspiration system prior to occlusion break. Stored elastic energy is then converted to kinetic energy of fluid flow (surge flow) during these post occlusion events. The resistance to fluid flow damps down *oscillations* or continuous exchanges between elastic and kinetic energy, limiting the time course of the post occlusion surge event, dampening it out after one half cycle of oscillation.

The 5th factor determining anterior chamber geometry stability or instability is the inflow limitation or the limitations of the irrigating instrument irrigation fluidic pathway. The inflow is limited due to three primary factors. The first is the fluid flow *resistance* in the entire irrigation pathway. The second is the *inertia* of the fluid in the entire irrigation pathway and the third factor is the *compliance* of the eye. These latter two factors along with the flow resistance are important in determining the anterior chamber stability in *dynamic* flow situations when the flow is stopping and starting. A good part of this book deals with *flow resistances* in the fluidic flow pathway as these are essential in the understanding of *static* flow and eye pressure situations, and also have significant impact on *dynamic* eye pressure and flow situations.

(1-2) FLUIDICS ENERGY AND ULTRASOUND ENERGY IN THE BALANCE.

The cataract is "aspirated" from the eye primarily by the fluidics system and it is this system that determines the anterior chamber stability during the surgery. Unstable chamber pressure and therefore geometry may result in tissue damage, one example being posterior capsule rupture by phaco needle laceration.

Ultrasound energy allows lenses of hard consistency to be fragmented to a diameter or a size small enough to enter the phaco needle lumen and be aspirated. Soft lenses may require no ultrasound. One of the reasons why longitudinal (conventional) ultrasound is so effective is that it disrupts the fragment on the face of the needle tip where vacuum (reduced pressure) on the phaco needle lumen side of the fragment and bottle pressure on the other side of the fragment, is attempting to force the fragment into the mouth of the phaco needle against the repulsive forces of the oscillating phaco needle.

There is no such thing as "suction" which pulls on the fragments. For example when you "suck" a milkshake through a straw you create a positive pressure gradient and atmospheric pressure "pushes" the milk up the straw to the zone of lower positive pressure in your mouth. The same applies to phaco fluidics in that the cataract particles are not "sucked" from the eye, but pushed out of the eye by a positive pressure gradient that exists across the entry area or mouth of the phaco needle. These fluidic forces encourage contact between a loose lens fragment and the oscillating needle. This aids emulsification or *fragmentation* of free chunks of lens matter by the ultrasound energy vibrating the phaco needle. Without these forces the fragments would be pushed away by the ultrasonic needle oscillations and float around in the anterior chamber. During initial sculpting of the lens body, the forces encouraging contact between the lens material and the phaco needle edge are provided by the zonules supporting the lens, not by the fluidics system.

New ultrasound modalities such as various pulse width modulation schemes for the ultrasound energy or rotational modes of ultrasound tip motion will need to be carefully evaluated in a laboratory setting so that it can be determined if the phaco-emulsification is more efficient in terms of reducing ultrasound power and heat delivery to the anterior chamber for a given hardness of a reference artificial lens material removed at a specific rate under controlled fluidic conditions. The fluidic settings influence the rate and efficiency of cataract extraction provided the ultrasound is effective enough that occlusion is not occurring for long periods of time interrupting the flow. Therefore the ultrasound energy and the modality of the delivery, either longitudinal or rotational, need to be *untangled* from the fluidic settings by keeping these identical during any comparative study. This has to be done to reliably conclude if any new ultrasound modality is an improvement or not. Also, as physicians and advocates for our patients, we need to be sure that any small incremental improvements in Phaco Machine technology are cost effective, as ultimately the cost of increasingly expensive machines and more complex multi-crystal phaco probes are passed on to the patient. If there is a not a significant practical improvement to be had, then the additional cost of these modifications or "improvements" may not benefit the patient. Significant improvements in phaco-emulsification technology and the surgical safety will be likely to occur via further incremental improvements in the fluidics systems, with assistance from improvements in ultrasound delivery modes.

In the balance of energies required to remove a cataract the fluidics forces are better optimised to their maximum values within safe limits of anterior chamber stability, while the ultrasound should be minimised due to its toxicity and thermal effects. New laser modalities for lens fragmentation will ultimately be helpful as less overall ultrasound will be required for the removal of a cataract and in some cases with soft to medium hard lenses no ultrasound may be needed at all with adjunctive laser. It is likely that the fluidics system will be required for many decades into the future.

(1-3) EQUIVALENT "FLOW CIRCUITS" IN PHACO FLUIDICS – A NEW MODELLING SYSTEM:

Pressures and vacuums in phaco fluidics are measured with respect to atmospheric pressure, defined as the zero reference point. Vacuums, unless equal to the full atmospheric pressure, are still a positive pressure equal to the atmospheric pressure less some amount we call the vacuum. For example the bottle pressure is measured to be positive with respect to zero, and the vacuum negative.

Although there is no actual fluid communication to return the fluid from the collection bag or cassette back to the bottle fluid, the fluid ultimately passes between points of zero atmospheric pressure after passing via the fluidic system. Therefore the fluid can be "imagined" to be returned via a continuous flow circuit to an imaginary volume of fluid at atmospheric pressure. In the case of the venturi machine, the waste fluid is sequestered in the cassette, and exposed to the varying vacuum there, depending on the pedal position at different times, and this has no effect on the flow circuit concept. The waste fluid can still be imagined to have been returned to the surface of a fluid filled vessel at close to atmospheric pressure when there was no vacuum at the cassette. The same applies to the fluid sequestered into the waste bag of the peristaltic machine. This analogy is possible because once the waste fluid is sequestered "somewhere" no more physical work is done upon it, transporting it anywhere else, and it has no kinetic energy of motion so it can be thought of as returned to a vessel of fluid at atmospheric pressure, or *recycled* to a vessel at atmospheric pressure.

The "flow circuits" concept provides distinct advantages in analysing the flow system properties for phaco fluidics. The author has created a flow circuit system so as to model phaco fluidics in a similar manner to the way electrical circuits are modelled. This also allows known laws for electrical flow circuits, such as *Ohm's law*, *Kirchoff's law*, the *Thevenin Theorem* and formulae for series and parallel *resistances* and for *compliances* (capacitances) to be applied to the fluidic circuits. In addition the concept allows for effortless addition of features such as external pressure acting on the globe, or wound leakage flow. The flow circuit concept also allows for the inclusion of compliant structures, such as plastic tubing and the eye itself which is a compliant structure.

In the case of plastic tubing, there is a further interesting quality in that the three important properties of *resistance, inertia* and *compliance* are evenly distributed along the length of the tubing. This is results in some very important and interesting phenomena which are involved in phaco fluidics. Another property of *flow circuits* is that they can be solved directly simply and quickly by running them in a computer engine known as a circuit simulator. There are a number of these on the market which perform this function. The graphs in this book were printed by the Super-Spice simulator. This means that some of the extraordinarily long and tedious equation solutions, which are derived from the flow circuit's fundamental differential equations, can simply be plotted as a graphical solution. Using this methodology is particularly appropriate when dealing with flow circuits which are undergoing transient or *dynamic* changes in pressure and flow, as they do in phaco cataract surgery.

The flow circuits involved, with the three qualities of *flow resistance*, fluidic *inertia* and *compliance*, are examples of systems which display *damped simple harmonic motion*. These are equivalent to many electrical systems and man-made or naturally occurring mechanical systems which have these three qualities. The damped suspension in your car being a good example where the spring represents the compliance, the shock absorber represents the resistive losses and the mass of the wheel and suspension supporting it representing the inertia.

While hydraulic or "fluidic calculation software" does exist, it is not greatly specialised to the transient flow situations of phaco fluidics primarily because the critical property of *compliance* of the fluid carrying system objects is completely neglected. A number of the exact flow and pressure equation solutions are presented in this book for completeness so as to confirm that they agree perfectly with the circuit simulators graphical models and also agree with practical experiments. The model is also such that any small part of the fluidic circuit can be altered and the result of that on *static* anterior chamber pressure and *dynamic* anterior chamber pressure can be visualised immediately. This aids in the design of new fluidic surgical instrumentation.

The reader should not be discouraged by the use of the equations as they are explained step by step throughout this book and are essential to the accurate understanding of phaco machine fluidics. It will be explained how various proportions of flow *resistance*, *compliance* and *inertia* relate to each other. From these equation solutions we find the true factors which influence specific sets of fluidic events during cataract surgery. Some of these findings are quite surprising. If the equations are disagreeable, the important results are described and graphed for the reader which makes for much easier reading and interpretation.

(1-4) UNDERSTANDING PRESSURE and VACUUM:

Pressure, or force per unit area (Newtons per square meter, N/m^2 in metric units) exerted by a fluid on a surface, can also be regarded as *Energy per unit Volume* of the fluid, by multiplying both force and area by a linear dimension. The units are then Joules per cubic meter or Energy/Volume. Therefore in Fluidics, understanding "pressure" as a concept is aided by thinking of pressure as "potential energy density" or an *energy source* capable of doing work on a volume of fluid flowing via fluidic pathways between points of higher pressure, toward points of lower pressure. These pathways are typically tubes or apertures, which have a "pressure gradient" across them. The pressure gradient is the difference between the two pressures (Higher pressure – Lower pressure) at either end of the tube or fluid passage under consideration. The fluid is driven or "transported" from the side of higher pressure toward the side of lower pressure.

Fluid pressure can be generated by a number of means, typically a pump, however a vertical column of fluid generates pressure. This potential energy density you get for the price of lifting the fluid column, or BSS bottle (balanced salt solution) against gravity. The pressure generated is proportional to the height of the fluid column, the density of the fluid, and the gravitational field. The product of these three is $\rho \ge \alpha \ge \alpha \le \beta$, where ρ is the density of the fluid (water 1000 kg/m³), and h the height of the fluid column in meters, and g is gravity or 9.8. N/kg. From this point on the symbol Pb in this book will be synonymous with the term "Bottle Pressure" For example a 70 cm bottle height generates 6860N/m² or close to 51.5 mmHg. BSS, as a fluid, has only a slightly higher density than water because of the solutes it contains, and we can regard water and BSS as nearly identical for our purposes from the point of view of density and viscosity.

Vacuum, typically generated by the Venturi device, or due to an electric pump, is measured also with respect to atmospheric pressure but it is still a *positive pressure* unless the vacuum value is equal to atmospheric pressure of 760mmHg, then its positive pressure value is zero. Atmospheric pressure, close to 760mmHg, is our "zero reference pressure". Obviously then, a vacuum can never exceed atmospheric pressure or 760mmHg. Typically the highest vacuum value generated by a phaco machine is 600 to 650mmHg, which is a *residual positive pressure* value of 160 or 110mmHg respectively. As surgeons we express the vacuum as a positive value in conversation but we need to remember it is actually a positive pressure equal to 760mmHg less the vacuum number we are talking about. The symbol used for vacuum is Pv in this book.

Consider the higher bottle pressure Pb on the left hand side of a fluid filled pipe and a vacuum Pv (a lower than atmospheric pressure value) on the right hand side of the pipe. The pressure gradient, driving the fluid from left to right through the pipe is: Pb - (-Pv) or simply put: *Bottle pressure value* + *vacuum value*, both expressed as positive numbers. For this reason, all the vacuum values in every equation in this book, are formatted a positive vacuum value, as we would refer to vacuum in conversation.

(1-5) ANTERIOR CHAMBER PRESSURE:

If the eye's anterior chamber pressure remained constant at all times during surgery, so would the geometry of the anterior chamber. If the anterior chamber pressure is too high, the chamber can be too deep, if too low, the anterior chamber shallows, and if very low to negative the eve can collapse, with the cornea buckling inwards. The eye is a compliant (elastic) vessel and its overall volume and anterior chamber volume depends on the intraocular pressure less the value of any external globe pressure. For any given average anterior chamber pressure, there is a range of anterior chamber depths, which depend on the compliance and geometry of the particular eye, the presence of any external forces such as the speculum acting on the globe or tense orbital tissues generating "vitreous pressure" as it has been called. Highly myopic eyes tend to have very deep chambers for a given anterior chamber pressure, while the reverse may be the case for hypermetropic eyes. Anterior chamber volume may also be lower in eyes with large cataracts that have crowded the anterior chamber over many years. In most cases during surgery, the anterior chamber pressure, symbol Pe in this book, is run at higher than physiological values, in the range of 30 to 70 mmHg. This helps to allow for the pressure reducing effects of the four forms of fluid flow variation and the inflow limitations to be described.

Under stable or constant un-occluded flow conditions the eye pressure reduces primarily due to flow *resistance* in the fluid inflow or irrigation pathway. This irrigation flow resistance is dominated by the small internal calibre of the *irrigating chopper* (irrigation needle) or the *annulus* geometry for the coaxial irrigation device formed by the loose fitting sleeve around the exterior of the phaco needle. There is also a significant contribution from the resistance of the irrigating tubing and other fittings in that pathway. In general, due to flow resistance, the pressure is dissipated or lost along the irrigation pathway with fluid flow. Dissipation is this case is an irreversible loss of pressure (energy density) to heat, by the process of fluidic friction.

The fluidic system properties of *compliance* or springy nature of flexible structures such as the eye and tubing and *inertia* or reluctance to change velocity of the fluid mass are very important with *dynamic* flow situations. These two properties do not have the energy dissipating properties of *resistance*. Compliant structures may store energy by their elastic nature and fluid may acquire more or less kinetic energy as it is accelerated and decelerated during surges and flow occlusions. The stored energy in elastic structures and the kinetic energy of moving fluid may exchange with each other, or exchange with pressure energy in the fluidics system in a nearly lossless manner, however pressure energy lost through dissipation as heat in the flow resistances is never recovered.

When performing any fluidic calculations at all, even the simple ones, it is best to keep all of the parameters in standard metric units. This avoids errors and allows quick checking of units to see that they are accurate. However as surgeons we are used to ml/min or cc/min flow and pressures in mmHg. Throughout this book all the calculations are performed in mks (meter kilogram seconds units) and converted back and forth to the *surgeon familiar* mmHg and ml/min as required. The conversion is done for the reader throughout the text. All that is required is a scientific calculator. Table 1 below is set up so multiplication only is needed for a conversion.

TABLE 1:

Convert from:		To:	
mmHg N/m ² ml/min m ³ /s ml m ³	\uparrow \uparrow \uparrow \uparrow \uparrow \uparrow \uparrow	N/m ² mmHg m ³ /s ml/min m ³ ml	multiply mmHg by 133.3 multiply N/m ² by 7.5×10^{-3} multiply ml/min by 1.667×10^{-8} multiply m ³ /s by 6×10^{7} multiply ml by 1×10^{-6} multiply m ³ by 1×10^{6}

Also 1cm $H_20 = 0.733$ mmHg which is a useful conversion.

(1-6) UNDERSTANDING FLOW:

It may seem that fluid "flow rate" is self evident. It is important to remember the flow we are referring to, for example ml/min or m^3/s , is a *bulk flow rate* called *volume rate of flow* (symbol I in this book, symbol Q in some texts). Generally the volume flow rate is used to describe a volume of fluid being transported from one location to another at some rate where the units are volume per unit time. This also corresponds to a mass of fluid being moved at some rate where the *mass flow rate* is the product of the volume flow rate and fluid density. So we can consider *flow rate* as either a *volume transport* or a *mass transport* phenomenon.

For any considered cross sectional area, at some point in a fluidics pathway, the fluid will also have an *average velocity* \mathbf{v} . The product of average velocity and cross sectional area is the flow rate. The average values of fluid velocity are used in the calculation of *flow resistances* and *Reynolds numbers* (see below) and for the calculation of the fluid's *inertia* and *kinetic energy density*. In all cases the driving force or the source of energy for this *mass transport phenomenon* is the applied pressure gradient.

In general, if a bulk rate of flow I, in a tube, is divided by the cross sectional area of that tube then the average velocity \mathbf{v} , of the fluid will be obtained. This is because the volume of the fluid in any length of the tube is equal to the product of length and cross sectional area:

 \mathbf{v} = Flow rate / Cross sectional area, or for a tube of radius \boldsymbol{r} :

$$\mathbf{v} = \frac{\mathbf{I}}{\pi r^2}$$

Equ.1

(1-7) PRESSURE LOSSES WITH FLUID FLOW- FLOW RESISTANCES:

When fluid flows via conduits, energy density (pressure) is lost due to *flow resistance*. Resistance itself occurs due to frictional forces between molecules and layers of fluid. In general *frictional* pressure losses result from the interaction of the molecules between the fluid layers which move at different velocities.

If we look at *laminar flow* in a tube for example the flow can be visualised as a series of thin concentric hollow cylinders of fluid, sliding on one another, Burns¹. It is as though the central cylinder of fluid is being pulled forward, with cylinder the around it being dragged at a slightly lower speed, and that cylinder drags the larger cylinder around it and so on at progressively lower speeds. The thin layer of fluid on the wall of the tube is not moving. This results in a parabolic fluid velocity profile where the fluid near the centre of the tube has the highest velocity and the velocity decreases towards the wall of the tube. Fluid *friction*, and resulting *flow resistance*, can be imagined as occurring between the concentric cylinders of fluid, as they slide on each other with different relative velocities. *In this instance the flow resistance is independent of the tube wall and independent of the flow rate*. The frictional energy or pressure losses in this case are *dependent linearly on the first power of the flow velocity or flow rate*. Frictional losses cause energy dissipation, in this case pressure energy, which is irreversibly lost as heat.

The pressure gradient applied across a length of tube or flow resistive pathway under consideration is therefore the pressure energy dissipated by the resistance and the relation is widely known as Pressure = Flow x Resistance or P = I.R where P is the applied pressure gradient, I is the flow rate and R is the flow resistance. This is an example of Ohm's law. The applied pressure is the pressure energy density lost or dissipated as heat due to the flow resistance. The units of resistance are Pressure/flow or units of J. s. m⁻⁶. When resistances are specified in this book these are the units.

If the resistance is independent of the flow rate, as it is with laminar flow, you can calculate a flow resistance value for a tube and the flow rate can be zero or a real value. So the resistance value in this instance is a property of the fluid filled tubing and has one constant value for any flow rate from zero to some figure, provided laminar flow does not break down. This is an interesting concept the notion of a fluid flow resistance which still has a numerical value even at a zero flow rate where there is zero energy dissipation. In this case however the practical resistance property really only exists at some flow rate greater than zero.

On the other hand in a *turbulent* flow situation, where laminar flow within a tube has broken down, the fluid particle interactions increase. The fluid acquires velocity components which are not just parallel to the wall of the tube but also have a transverse component in direction with respect to the flow lamellae. Areas of fluid exchange momentum with each other increasing the pressure losses and energy dissipation. This is known as turbulence. The fluid also then has interactions with the tube's wall such that the roughness of the wall can become important too. In this instance the flow resistance becomes dependent on the flow velocity or flow rate. The pressure losses then become proportional to the square of the flow rate. Turbulent flow within tubing occurs at high Reynolds numbers > 2000(see below).

Another similar type of pressure loss due to turbulence to be considered in phaco fluidics is local pressure loss due to *local turbulence*. These pressure losses are caused by local disturbances of flow which occur when there is a sudden change in cross sectional area of the fluid flow pathway, typically a constriction in the flow passage. For example local pressure losses occur at tube fittings and joiners and at the entry areas of needles or orifices, or places where the flow bends or branches. These pressure losses result from local disturbance of the flow, separation of flow streamlines from the vessel walls and the formation of vortices whorls and turbulence. Again there are transverse velocity components of the fluid, in this case toward the centre of the constricted area. Local pressure losses have the typical property of turbulent flow behaviour in that the pressure loss is proportional to the square of the flow rate.

Now we can consider the two fundamental types of resistance behaviour seen in phaco fluidics: Firstly a form of resistance, which we calculate for a "laminar flow" This is typically within a long tube. The flow rate has no effect on the resistance value (provided the flow rate is not so high that turbulence has occurred). We call this *linear resistance* to signify that the resistance is a constant which is independent of the flow rate and that the pressure loss with flow is equal to the product of flow rate and the resistance value as predicted by Ohm's law.

Secondly, we can now consider turbulent flow situations within tubing where laminar flow breaks down, or local turbulence, where pressure losses occur at local constricted points within the fluidic pathway. These local losses occur at orifices and fittings. They also occur with short fluid carrying objects with abrupt entry constrictions. The pressure losses in this instance are proportional to the square of the flow rate. To signify this we can call this sort of resistance a *square law resistance*. This is to signify that the pressure losses, on account of this type of resistance, are proportional to the square of the flow rate linearly. This is the case because of the general relation between pressure, flow and resistance given by Ohm's law above. If the resistance itself is flow dependent, or in other words R is proportional to the flow rate.

Some objects in the fluidic system such as the irrigation and aspiration tubing are dominated by linear resistance. On the other hand the short objects with entry points such as the coaxial irrigating sleeve annulus, irrigating choppers and phaco needles are dominated by square law resistance. Square law resistance behaviour dominates at typical surgical flow rates in these objects. Square law resistance behaviour determines the resistance of short small apertures such as the hole of the IA (irrigation-aspiration) handpiece or holes in irrigating sleeves. The implications of this will become clearer as the fluidics system is described.

The square law flow resistances, which are a significant proportion of the total irrigation flow pathway resistance, are responsible for the widely known fact that doubling the bottle height(or applied pressure) will not result in doubling of the flow rate in a free flow irrigation test on irrigating instruments. In general the flow rate, on account of the square law resistances, starts to become proportional to the square root of the applied pressure or bottle height, rather than the bottle height directly as the flow rates increase.

In general laminar flow, which has linear resistance, is not established immediately as the fluid enters the end of a tube, but at a point downstream of this. Near the tube entry the *local* square law resistive losses are proportional to the flow rate. Later when laminar flow is established, the resistance is independent of the flow rate, provided turbulence does not occur. Laminar flow is established within an entrance length of 0.035Re.D, where Re is Reynolds number (see below) and D is the tube's internal diameter, Bird². Reynolds number is proportional to the flow velocity so the faster the fluid is flowing the longer the entrance length will be before laminar flow is established. The entrance length is only a very small proportion of the total length for items such as the irrigation or aspiration tubing with the flow rates used in phaco. For example laminar flow at 30 ml/min is established 3 cm into the length of the irrigation tubing with a flow rate of 30ml/min or just below 1% of the length of the tubing. Therefore, calculations for the flow rate of a minar flow as linear.

For shorter items or objects such as phaco needles, and irrigating choppers, established laminar flow does not occur until a length which is longer than the object itself and the laminar flow formulae cannot be used with any accuracy. The fluidic object then behaves more like an entrance area with pressure losses which depend on the square of the flow rate and a flow resistance which depends on the flow rate. For example a 20mm long 0.635mm internal diameter phaco needle has a measured flow resistance of around 7.78x10⁹ at 30 ml/min, but about double this at 60ml/min. At very low flow rates, when laminar flow is established within most of the needle length, the resistance is flow independent and has a value of $5x10^9$ for example. The flow resistance of the needle therefore increases during high flow rate conditions. The same applies to irrigation devices such as needle/sleeve annuli or irrigating needles or irrigating choppers which are irrigating needles with a projecting spike to assist lens chopping.

Reynolds number Re, is a useful number which can be used either in the calculation of flow resistances or to help predict when established laminar flow in a tube might break down to turbulent flow. Resistance values calculated during laminar flow situations are very accurate for tubing provided the flow is laminar and not turbulent. We can calculate the flow rates at which turbulence is likely to occur in any structure in the fluidics pathway using Reynolds' number. In general though for the shorter objects in the fluidic system where *local* entrance losses dominate there is already turbulence at that point. Re is a dimensionless ratio which represents the ratio of inertial to viscous forces in a fluid flow situation. When the fluid is flowing too rapidly the inertial forces exceed the viscous forces holding the molecules together and streamline flow breaks down. One can imagine the inertial forces trying to pull the streamlined lamellae of fluid apart into a disorganised flow profile of whorls and vortices with transverse velocity components which are not parallel to the flow streamlines.

When fluid is passing via an aperture of diameter D, with average velocity \mathbf{v} then Reynolds number is, Bird²:

$$\operatorname{Re} = \frac{\rho.\mathbf{v}.D}{\eta}$$

Equ.2

Where ρ is the density of the fluid (1000 Kg/m³ in this case) and η the viscosity of the fluid $1x10^{-3}$ Kg. m⁻¹. s⁻¹ (for water) and D, is the hydraulic diameter of the aperture which for a tube is just the internal diameter. As the Reynolds numbers increase the inertial forces which are proportional to the square of the fluid velocity, become more dominant. This turbulent process begins at Reynolds numbers of approximately > 2000.

At high flow rates in phaco fluidics, say 60ml/min, the Reynolds number for typical 1/8 inch internal diameter irrigation tubing is only 400 and for the 1/16 inch internal diameter aspiration tubing only 810, so the flow is laminar and the resistance is *linear* and independent of the flow rate. On the other hand the resistance values for the smaller and shorter objects in the fluidics system depend on the flow rate. These resistances can be either calculated or directly measured which is the preferred method. *These resistances should be determined at flow values similar to the actual flow rates which occur during surgery and not at other higher flow rates which do not occur during surgery.* If this is not done then false assumptions may be made about the fluidic performance of the device.

(1-8) CALCULATING AND MEASURING FLUID FLOW RESISTANCES- OBJECTS IN THE FLUIDIC PATHWAY:

In general flow resistance gives rise to pressure losses. The pressure loss with flow is given by the Darcy Wiesbach Equation: Idelchik³:

$$P(loss) = \lambda \cdot \frac{l}{D} \cdot \frac{\rho v^2}{2}$$
Equ.3

Where ρ is the fluid density, **v** the average fluid velocity, *l* the length of the tube or conduit in question and D the diameter, which is the internal diameter of the tube. For laminar flow (Re < 2000) the value of lambda λ , depends on the Reynolds number alone and is equal to 64/Re. Substituting this value into equation 3 and also substituting in equation 1 for the value of **v** we obtain:

$$P(loss) = \frac{I.8.\eta.l}{\pi.r^4}$$
Equ.4

The reader may recognise equation 4 as the Hagen- Poiseuille equation for pressure loss along a pipe length l with a flow rate I and a radius r. Thus from the general relationship Pressure = Flow x Resistance or P = I.R (Ohms law) then this makes the *linear flow resistance* R:

$$R = \frac{8.\eta.l}{\pi r^4}$$
Equ.5

This is a very steep relationship, in that increasing the internal radius of a tube by only 20% will halve the fluid flow resistance, and doubling the radius will result in a 16 fold reduction in flow resistance. Poiseuille's law is valid only for established *laminar* or non-turbulent flow in tubing. *This resistance value is independent of the flow rate*.

The reason why the Darcy equation was introduced first, rather than going directly to Poiseuille's law, is that the value of lambda λ , and the length to diameter ratio l/D, can be replaced by one symbol K, a *friction coefficient*. The coefficient is a number which can describe pressure loss under turbulent conditions in tubing or due to *local* constrictions within a fluidic pathway.

As we have seen for the non turbulent flow situation in tubing where Re <2000, the value of K merely depends on the Reynolds number and length to diameter ratio of the tube and *decreases* with an increase in the Reynolds' number. However at higher Reynolds' numbers 2000 to 4000 at the onset of turbulence, K *increases* with the Reynolds number. At higher Reynolds' numbers > 10^5 the K value can be regarded as independent of the Reynolds number and dependent very much on the roughness of the pipe walls in the case of flow in a tube.

Once turbulence has occurred, either at local constrictions or with laminar flow breaking down, the pressure losses become dependent on the square of the flow velocity or the square of the flow rate. Pressure is lower in the downstream area after these losses. The *local* pressure loss can be described as follows:

$$P(loss local) = \frac{K}{2} \frac{\rho v^2}{2}$$
Eau.6

The velocity \mathbf{v} of the fluid in this case is the average velocity within the smallest part of the aperture at the point where the velocity is maximal. This cross sectional area is a little smaller than the cross sectional area of the object because the fluid streams contract toward the centre of the aperture after they enter the orifice. Determination off the friction coefficient K for any *local* part of the fluidic pathway is best done by practical experiment at the flow rate of interest. In practice, in the case of phaco fluidics, the calculation of a friction coefficient value is not often needed because the flow resistance at any given flow rate is easily determined by practical experiment. This is done simply by dividing the applied pressure by the flow rate. The benefit of knowing equation 6 is the relationship between pressure loss and flow velocity at *local* areas in the fluidic system or under turbulent flow conditions. Substituting the result from equation 1 into equation 6 we obtain:

$$P(\text{loss local}) = \frac{K.\rho I^2}{2\pi^2 r^4}$$
Equ.7

As we can see from equation 7 the *local* pressure losses are proportional to the square of the flow rate I and the K value for the local area or constriction in question and again inversely proportional to the 4th power of the tube's radius.

For our purposes in phaco fluidics, while pressure losses are an indicator of flow resistance, we need to know what the resistance value is at *any given flow rate*. In the case of the Hagen Poiseuille's law describing pressure loss and volume rate of flow it was easy to take the resistance value directly from the formula to acquire equation 5 for the linear flow resistance. In the case of equation 7, if we were to plot it as a graph, with flow I on the x axis and pressure loss P on the y axis, the graph would simply have the general form of $y = ax^2$ and dy/dx would be the slope of the graph. Resistance at any specific flow rate is always the ratio of change in pressure to change in flow rate and in this case the slope of the graph (tangent) to the graph at any point or any specific value of flow rate of interest. Therefore we simply differentiate equation 7 to find dP/dI which is the value of the square law resistance at some flow rate:

$$\frac{dP}{dI} = R(local) = \frac{K.\rho.I}{\pi^2 r^4}$$
Equ.8

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The result of equation 8 tells us in general that the *local* resistance at a flow constriction is proportional to the flow rate I and this is quite unlike the resistance given by the Hagen – Poiseuille's law where the flow resistance is independent of the flow rate. The K value which determines the local pressure losses and therefore the resistances in flow constrictions are best determined by experiment with the flow rates in the vicinity of the values of interest. We can therefore summarise flow resistance R for an "object" in the fluidics system and its fluid entry area as having two terms which are flow independent (Poiseuille's) and flow dependent resistance:

$$R(object) = \frac{8\eta l}{\pi r^4} + \frac{K\rho I}{\pi^2 r^4}$$
Equ.9

At low flow rates where laminar flow establishes itself along the bulk of a tube length, the second term is not significant. For short tubes with entrance constrictions, or at high flow rates which induce turbulence, then the square law resistance begins to occur and the total resistance to flow will be found to be greater than that calculated by the first term (Poiseuille's resistance) alone. This especially applies to short objects with minimal length l, and with small entrances such as phaco needles and irrigating choppers, or object like holes which are short in length compared to their diameter. The second term of equation 9 describes the flow resistance which is dependent on the flow rate I.

In summary we can say therefore that at low flow rates and low Reynolds numbers the frictional forces due to fluid viscosity dominate and pressure losses are linearly proportional to the flow rate. At high Reynolds numbers and in *local* areas, typically abrupt constrictions of the flow passages, the fluid streamlines are disrupted by transverse velocity components the pressure losses depend on the square of the flow rate. The result of this is that there are two fundamentally different types of flow *resistance*, one independent of flow rate and one dependent on flow rate in the phaco fluidics system. For this reason, when flow resistances of the smaller objects in the fluidics system are specified, they must be specified at some flow rate of interest, typically 30ml/min, and the flow resistances at higher flow rates, eg 60 and 90ml/min are sometimes needed. A number of these values for various objects in the phaco fluidics system are listed in Table 2 in the Data Chapter.

Total Irrigation and Aspiration flow pathway resistances:

One of the most important flow resistance totals we are dealing with in phaco fluidics is the total resistance in the *irrigation* pathway, symbol Ri. This is because it is the resistance that is "in series with" or in the fluid flow pathway between our bottle pressure source, and the eye's anterior chamber. For any "constant flow" of fluid, via this resistive pathway Ri, there will be a pressure loss, and this loss will be the product of the irrigation flow rate, symbol Ii and the total irrigation resistance Ri:

$$P(loss) = Ii.Ri$$

Equ.10

The pressure loss along the irrigation pathway leading to the eye subtracts from the pressure provided by the bottle height. Flow resistance in the irrigation pathway Ri, has a *total value* of about 7.36×10^9 @ 30ml/min in a typical irrigation arrangement, for example with an AMO Yellow sleeve and needle annulus passing through a wound into the eye and a standard irrigating tubing set, bottle fittings, probe and connections. The total irrigation flow resistance without compression of the annulus by the wound is a little lower at 6.32×10^9 @ 30ml/min. The values alter if other irrigating devices are used. Figures for the total Ri with other irrigating devices are listed in table 2.

The eye's anterior chamber resides in a position in the fluidics pathway which is between the *irrigation* and *aspiration* system resistances, therefore it is useful to lump the resistances into two lots, simply the total irrigation flow pathway resistance Ri, and the total aspiration pathway flow resistance Ra, these are what we need for all of our basic *static* flow and anterior chamber pressure calculations. Also the total of these two, the *total system resistance* Rt, is another useful figure:

$$Rt = Ra + Ri$$

Equ.11

We can use the Rt value to calculate the power loss due to fluid friction in the entire fluidics pathway. Using the general relation power = $(Flow)^2 \times Resistance$. With 30ml/min ($5 \times 10^{-7} m^3/s$) flow and the typical Rt value of 2.89×10^{10} the power lost in a typical fluidics system is only about 7.2 milli-watts. The heating of the system's fluid on account of fluid friction is therefore negligible. The bulk of the work done by a peristaltic pump motor in a phaco machine therefore is largely to overcome the friction of the rollers on the tubing and not in moving the fluid.

For the objects where there is a significant flow dependent (square law) resistance the values of resistance are listed at 30ml/min and 60ml/min. 90ml/min resistance values are also listed in table 2 for various phaco needles. The total values for the 3 tube connectors and bottle fitting contribute a little on the irrigation side with some square law resistance, while the irrigation tubing has one value of linear resistance. The values in table 2 for the tubing and probe are typical values for industry standard needles and sleeves. The figures in the table can be regarded as approximate and fairly standard for peristaltic machines. In the case of the venturi machine the Ra values are usually somewhat larger due to higher flow resistance in the phaco needle and longer lengths of aspiration tubing typical of venturi machine disposable sets.

In general flow resistance can be measured by dividing the applied pressure in mks units by the flow rate in mks units. The figures in table 2 are a good starting point to describe the *flow resistance* properties of the phaco fluidic system. These were calculated or measured on repeat experiments by the author using water flow at 25 deg. C and can be regarded as useful approximations. Table 2 also lists values of the *Inertia* and *Compliance* of the components too. These properties will be explained later and we will need these values later when looking at transient or *dynamically* changing flow situations.

The flow resistance via an annulus:

Out of the eye and at very low flow rates, the flow resistance of the annulus of the coaxial phaco system is fairly accurately described by a formula resembling Poiseuille's equation with a correction factor which depends on the two relative diameters which comprise the annulus, Bird². Also the cited formula (not shown here), like Poiseuille's formula, gives a *linear* resistance which is independent of the flow rate. For typical sleeves and needles in phaco the correction factor ends up at a value around 20 to 50 depending on the exact annulus geometry. So we can think in this case of adding the needle to the sleeve, increases the resistance to flow by about 20 to 50 times of what it would be if the sleeve was acting as a tube alone. If the formula however is used to predict the flow resistance of the yellow sleeve/needle annulus the value returned is 3.25×10^9 . When measuring the value by practical experiment the flow resistance @ 30 ml/min is 3.48x10⁹ and @ 60ml/min is 4.09x10⁹. This is because of the *local* square law pressure losses at the input area to the annulus which increase the flow resistance with flow rate. Also the change of fluid flow direction and losses at the sleeve's outlet holes, which also have flow dependent and *square law* pressure loss, are also responsible for the measured resistance value being higher than the calculated value using the formula. In addition to this the sleeve annulus architecture can alter due to wound pressure increasing the flow resistance there.

Practical experiment by the author with needles and sleeves passing through a surgical wound created in the limbus of a pig eye demonstrated that the flow resistance increases with compression of the annulus anywhere between 20 and 50% and typically by 30%. So the real flow resistance of the yellow needle/sleeve annulus, in use within the eye, during surgery is in the order of 4.52×10^9 @ 30 ml min. The grooved B&L micro flow plus needle and blue sleeve is less affected from compression by the wound and its resistance only increases by 10 to 15% with wound pressure. This is unlike the case with the irrigating chopper (metal needle) used in bi-manual micro incision cataract surgery where passage through the wound does not affect the flow resistance. The resistance values of the irrigating devices in listed in table 2 are also shown on a graph of figure 1.

It is also worth noting that the annulus of coaxial phaco is a very touchy fluidic animal. The flow resistance is noticeably affected by the centralisation of the needle and how far the sleeve is screwed on because most needles have a subtle taper. One turn of the sleeve can affect the flow resistance by 5 to 10%. Even slight stretching of the sleeve can change the flow resistance, or a slightly bent phaco needle can affect the resistance by 5 or 10%.



Flow resistance via an Irrigating Chopper:

The flow resistance of a standard (STD) irrigating chopper needle, see table 2, was constructed from a 19g needle shortened to 30 mm length. The measured flow resistance of this chopper is 4.33×10^9 @ 30ml/min and 6.04×10^9 at 60ml/min. The total irrigation flow resistance Ri with the chopper and its handle being used is 7.17×10^9 at 30ml/min. Laminar flow is not established in the chopper at typical flow rates and the resistance is dominated by losses at the entrance to the chopper needle. If the flow rate was very low and laminar flow was established then the calculated *linear* resistance of this chopper is only 2.98×10^9 . The flow resistance of an improved design of chopper is also shown in figure 1. This makes the *chopper system* better than coaxial sleeves, when properly designed, and also immune from compression by the wound. With the improved chopper and handle, the irrigation resistance total Ri, can be reduced to 5.18×10^9 .

Flow resistance via a hole:

The pressure loss via a hole or small aperture placed in a fluid flow pathway is a square law function. Laminar flow is not established because the hole is very short compared to its diameter. The first term of equation 9 (with the length value) is therefore insignificant and the second term, where the flow resistance is proportional to the flow rate is the significant term. Fluid streamlines are directed toward the central axis of the hole and the flow is locally turbulent. The pressure losses also depend of the architecture of the holes' edges where the fluid enters and this determines the K value. The K value for a rectangular or sharp edge hole entrance is typically in the order of 0.5, Idelchik³.

We expect these local resistive losses to therefore be flow rate dependent. Small holes, such as the 0.25mm to 0.26mm diameter hole in the IA (irrigation/aspiration) handpiece have a very high square law flow resistance. Using equation 8 and a K value of 0.5 this predicts flow resistance at 21ml/min for a 0.26mm diameter hole to be 6.2×10^{10} . The measured value for a hole in the IA handpiece needle by the author was 6×10^{10} at 21ml/min.

We can expect twice that resistance at 42 ml/min and three times that at 63ml/min. This relation can also be steeper because the K value itself increases with flow rate. The high resistance of the hole in the IA handpiece helps to stabilise the anterior chamber. This prevents (damps) post occlusion surge flow and pressure oscillation associated with that from being a major problem in IA mode with the peristaltic machine on high vacuum settings. This is why 500mmHg maximum vacuum can easily be run in IA mode without too much anterior chamber instability. Other reasons for this will also be explained in the section on *dynamic* phaco fluidics. The fact that this high flow resistance may inhibit the flow rate for a given vacuum level at the machine is of no consequence in IA mode as no heat of any significance is being generated in the IA handpiece. High flow resistance such as this would be very problematic in phaco mode where heat is being generated by the crystals in the phaco handpiece and the flow rate is being used to cool the ultrasound crystals.

Coaxial sleeve holes are in the order of 1mm diameter. The flow resistance of the hole is affected by the proximity of the phaco needle and the pattern of flow leading to the hole from the annulus. This pattern also changes when the annulus is compressed by the wound. Using equation 8 again as a basic guide this suggests that the flow resistance of a single sleeve hole Rh, is in the order of 1.35×10^8 @ 10ml/min, 2.02×10^8 @ 15ml/min, 4.05×10^8 @ 30ml/min. Measurements by the author of the flow resistance of a 1mm diameter hole in the side wall of a 1/8" internal diameter tube with a 0.2mm thick wall gave 4.70×10^8 @ 30ml/min. Measured values at higher flow rates gave more resistance than predicted with the K value of 0.5. For example 1.36×10^9 @ 60ml/min and 2.14×10^9 @ 90ml/min were found by experiment. This is because the K value is increasing with the flow rate over this range, so the holes' resistance increased by a factor of around 4.5 rather than a factor of 3 when the flow rate was increased by a factor of 3. So for a short hole, the pressure loss with flow is a little steeper than square law.

With a standard two hole sleeve in use at 30 ml/min total flow, then the flow via each hole is 15ml/min making the resistance of each hole on its own about 2.35×10^8 . The resistance of each hole acts in parallel to reduce the total hole related flow resistance to Rh/2, which is 1.17×10^8 . Adding a third hole to a two hole sleeve (upper or lower surface near its tip) and leaving the flow rate at 30ml/min, decreases the flow resistance for *two* reasons. Firstly the flow via each hole has dropped from $\frac{1}{2}$ the total rate to 1/3 the total flow rate or now 10ml/min for each hole. This reduces the resistance of each hole by 2/3. Also a third hole is now acting in parallel with the other two holes reducing the resistance from Rh/2 to Rh/3 or another 2/3.

Therefore the net reduction in *hole related* flow resistance of the 3 hole sleeve versus the 2 hole sleeve is $2/3 \ge 2/3$ or 4/9 which means compared to the two hole case where the flow resistance of the two holes was $1.17 \ge 10^8$ it is now 4/9 of this with three holes or $5.20 \ge 10^7$. The addition of the extra hole has reduced the *irrigation* pathway resistance Ri,

by $1.17 \times 10^8 - 5.20 \times 10^7 = 6.5 \times 10^7$ which is only about 1% of the total irrigation free flow resistance of 6.32×10^9 (annulus out of the eye value) when the flow rate is 30ml/min.

Therefore adding a 3rd sleeve hole, at a typical flow rate of 30ml/min, would be expected to reduce the overall irrigation flow resistance only by about 1%. Adding an extra hole to the Yellow AMO sleeve provides by experiment and the above calculation only an approximate 1% reduction in flow resistance at a flow rate of 30ml/min.

On the other hand the reduction of flow resistance *measured* by adding the third hole to a sleeve can increase when measured at *high flow rates*. This is because of the steeper than square law resistance property of the holes. At 3 x 30ml/min or 90 ml/min one might expect the *hole related* flow resistance values to be 4.5 times higher. Therefore the measured improvement by adding an additional sleeve hole would be around 4 or 5% (not 1%) of the total irrigation flow resistance. Also if the annulus in the area of the holes is more crowded the flow resistance of each hole may be further increased. It has been reported by Akahoshi⁴ that adding a 3rd hole to an Alcon 2.2mm coaxial sleeve and needle assembly, increased the flow rate for a fixed bottle height, from 98ml/min to 104 ml/min. Or a 6% increase in flow rate for a fixed bottle height. This represents a reduction in total flow rate returned to the original 98 ml/min, the total resistance would be approximately another 6% lower or about 12% lower because the resistance is flow dependent.

Therefore in light of the above it is very important to carry out experiments on irrigation apparatus with flow rates that resemble those which *occur in practice with the instrument in the eye*, rather than free flow irrigation tests with high flow rates. The peak flow rates, in the irrigation system when the eye is being irrigated *during surgery* are in fact much lower than the flow rate values attained when the irrigation device is out of the eye during a typical free flow irrigation test.

With a 70cm bottle height (51mmHg) the peak irrigation flow rates reach a value typically of around 38 ml/min only, in worst case scenarios, shortly after the eye pressure has collapsed to zero during large surges. With a bottle height of 1m the bottle pressure is 73mmHg and peak flow irrigation rates can reach 50ml/min and with a bottle height of 140cm reach in the order of 68 ml/min. These values are nowhere near the high flow rate figures of free irrigation flow experiments with the irrigation device out of the eye and running to free air or atmospheric pressure. Performing irrigation free flow tests at high flow rates can therefore be very misleading.

A 1 meter bottle height generates an eye pressure of close to 73 mmHg which equals or exceeds the central retinal artery perfusion pressure with no aspiration or significant wound leakage. At these high pressures, the anterior chamber depth in some patients can be far too deep to allow good stereo vision for safe surgery and it can be like "working down a well". This can increase complications such as posterior capsule rupture simply because the view is poorer. A highly pressurised anterior chamber also increases the forces on the zonules. A bottle height of 140 cm generates 103 mmHg. These bottle heights and pressures have been used to compensate for the post occlusion surge when using high maximum vacuum settings on the

peristaltic machine and to compensate for anterior chamber pressure loss at high un-occluded flow rates such as greater than 50ml/min in both types of machine.

Counter intuitively when using these high bottle settings, the amplitude of the post occlusion surge or pressure drop transient in the anterior chamber is *increased by raising the bottle height*, even though the average anterior chamber pressures are higher with the elevated bottle. How this occurs will be explained in the section on *dynamic phaco fluidics*. As Surgeons we should always use caution until we have scientific data which can guide us in what the safe upper intra ocular pressure value and duration should be during routine cataract surgery. Until then it might be wise to keep the maximum pressures at 3 times physiological or less in the range of 50 to 65 mmHg. This requires currently that the flow rate is kept below 30 to 40ml/min to avoid excessive pressure loss along the irrigation pathway resistances, and that the maximum occlusion vacuums in the peristaltic machine are less than 250 to 300 mmHg, depending on the particular machine, to avoid excessive post occlusion surge.

Often it is remarked that the bottle is elevated to increase the irrigation flow. Assuming there is no excessive leakage from the wounds and un-occluded flow, in the *peristaltic* machine, the irrigation flow is generated by the aspiration flow which is set and controlled by the peristaltic pump and is *independent* the bottle height. This flow generates a pressure drop across the irrigation resistance leading to the eye, below that pressure value set by the bottle height. Elevating the bottle does not increase irrigation flow rate under un-occluded flow situations (ignoring wound leak), but provides a higher positive anterior chamber pressure under any flow conditions. With zero out-flow from the anterior chamber, the eye pressure then assumes the bottle pressure. Elevating the bottle therefore means that the reduction in eye pressure value, due to the product of the pump's flow rate and irrigation flow resistance, does not take the anterior chamber pressure as close to zero as it would have if the bottle were lower. In the case of the *venturi* machine, elevating the bottle does increase the irrigation flow rate in un-occluded flow conditions because in this case the driving force for flow is the sum of the bottle height and machine's cassette vacuum which is applied across the entire system flow resistance Rt.

However under dynamic conditions, such as occlusion make, and occlusion break, the *dynamic* flow rates increase with increasing bottle heights in *both the venturi and peristaltic* machines and again in a manner which depends on the square root of the bottle height rather than the bottle height directly. This will be discussed in the section on *dynamic* fluidics.

Author's suggested protocol for measuring flow resistance of an irrigating device:

Perhaps the biggest trap to fall into, when testing and comparing irrigating devices, it to compare the free flow rates of two irrigating devices at a fixed bottle height. This has been standard practice in Ophthalmology for many years. For example at some bottle height one irrigating device may provide 50ml/min free flow and the "better" device 55ml/min. This seems to suggest that the better device is 55/50 or 1.1 or 10% better. The problem here is that alteration of the flow rate also alters the *square law resistance*, both of the device being tested and of the square law resistances of the other parts of the irrigation pathway. Therefore at the higher flow rate of 55ml/min all of the square law resistances involved, which are dominantly

those of the irrigation device being tested, are higher by a roughly a factor of 55/50. So in the situation where the flow rate increases by a factor of 55/50 at the same applied pressure, the actual flow resistance at the 50ml/min flow rate must have reduced by a factor of $(50/55)^2$ making the improvement, or reduction in flow resistance, more like 17% at the initial lower flow rate of 50ml/min. Another way to understand this is to look at Ohm's law again, where the applied pressure P equals I.R. The resistance value R itself in this case however is a square law resistance and it depends on I and is proportional to I. Therefore with a constant applied pressure the flow resistance R, whatever it is, is proportional to $1/1^2$. Therefore if the flow rate I, increases by 10% for example, with a fixed applied pressure, or bottle height, the flow resistance at the initial lower flow rate must have decreased by around $(1/1.1)^2 = 0.83$, or about a 17% reduction in flow resistance. The only way to avoid this problem is to *compare the irrigating devices at fixed flow rates*. This is also important so that the flow resistances, at a specific flow rate, of the remainder of the irrigation system, can be subtracted to determine the flow resistance of the irrigating device in question.

Also it is important to make the comparison at a flow rate which resembles maximum flow rates in actual use in surgery. In the case of an irrigating annulus, the annulus should be passed through the limbus of a pig's eye with a wound cut to specification for that coaxial sleeve and needle. If this is not time expedient the test can be done without a "dummy wound", bearing in mind that the flow resistance for the typical coaxial irrigator could be about 1.3 times higher than that measured in a free flow test.

Suggested method for assessing the flow resistance of an irrigating device:

Firstly the entire irrigation system excluding the irrigating device (chopper removed from the handle, or sleeve and needle removed from the phaco probe) is assessed. The bottle is adjusted by trial and error to acquire a flow rate at figures of interest, for example 30 ml/min. Then the flow resistances of the system, excluding the irrigating device, are easily calculated by dividing the bottle pressure value by the flow rate (in mks units). The flow resistances for the irrigating system, at the specific flow rates of interest, without the irrigating device are then known. Then the irrigating device is included and the bottle adjusted upwards by trial and error to give the same flow rate for example 30 ml/min and the new total resistances calculated. Then it is simply a matter of subtracting the resistance of the system, without the irrigation device, at the same flow rate, to yield the flow resistance of the irrigation device at that same flow rate. If the above is not done then very misleading results can be obtained, especially comparing the coaxial/sleeve/annulus irrigating system with an irrigating chopper, where the proportions of the linear and square law resistances quite different. It can be seen from figure 1 that a range of irrigating choppers could exist with resistance properties between that of the two choppers listed. In those cases when tested at flow rates of 30ml/min the choppers could be superior to a coaxial arrangement however when tested at high flow rates not seen in actual surgery they would seem inferior to a coaxial arrangement and mislead the observer. Obviously then to test our irrigating instruments meaningfully for *flow resistance*, we need to be aware of the typical peak irrigation flow rates actually occur during surgery, with the irrigation device in the eye, under both *constant flow* and *transient flow* situations. These will be looked at in the section on dvnamic fluidics. We now move on in Chapter 2, to flow circuits to model the pressures, flows and resistances in fluidics under stable un-occluded flow circumstances:

CHAPTER 2.

(2-1) STATIC PHACO FLUIDICS – FLOW CIRCUIT MODELLING APPLIED TO PHACO MACHINES.

"Flow circuits" representing the fluidic flow for phaco machines are now introduced, firstly for the Peristaltic machine constant un-occluded flow.

Peristaltic machine Flow Circuit modelling:

Figure 2 shows an equivalent fluidic Flow Circuit for a peristaltic machine:



The box Pb represents the bottle pressure and the peristaltic pump is also drawn as a box .The pump generates the aspiration flow rate Ia. The zig-zag lines represent the flow resistances Ri and Ra of the irrigation and aspiration flow pathways respectively. The total resistance is Rt = Ri + Ra. The eye is placed in the fluidic pathway at the junction of these resistances, so a circle is drawn there. Pe is the pressure within the eye. The lines joining the boxes to the resistances are imaginary lossless fluid flow pathways. The bottle pressure is 70cm H20 (51mmHg or 6798N/m²). The bottle can be regarded as a pump which picks up fluid from a zero pressure reference level and raises it to 51mmHg pressure.

The peristaltic pump can be regarded as a device which regulates a constant aspiration flow rate Ia, in this example 30ml/min. The arrow heads in the diagram show the direction of the flow of fluid which is *imagined* to cycle around the flow circuit. The box labelled IL represents the rate of leakage flow out of the eye and is an imaginary pump generating some leakage flow rate IL. Irrigation flow rate is represented by the symbol Ii and equals the sum of Ia and IL.

For a constant flow situation with a peristaltic machine we first consider the situation when the leakage flow I_L is zero and the wounds are perfectly sealed around the instruments. The flow rate in the aspiration system is set by the peristaltic pump, for example to 30 ml/min $(5x10^{-7} \text{ m}^3/\text{s})$ and as a consequence generates a vacuum Pv. This vacuum is monitored by the machine's vacuum sensor. As it is in all peristaltic machines, the pump is inhibited (instructed to stop) if this measured value at the sensor exceeds the maximum value set on the control panel by the surgeon. Regardless of any minor features built into a peristaltic machine's fluidic system, this is the main and most important aspect of the functionality of a peristaltic machine.

The overall driving pressure for the fluid flow is the sum of the bottle pressure Pb and vacuum level Pv and this is applied to the total system resistance Rt of 2.89×10^{10} a typical 30ml/min value taken from the disposables and setup listed in table 2.

Pv in this case is a consequence of the flow rate. The vacuum generated by the pump Pv is calculated by (Pb + Pv) = Ia.Rt or:

Pv = Ip.Rt – Pb (Un – occluded Vacuum equation Peristaltic machines)

*Equ.*12

This makes the vacuum value of $Pv = 7652 \text{ N/m}^2$ or 57.4mmHg when the bottle is providing 51mmHg and the flow rate is 30ml/min. An average value very close to this will be found in the AMO Sovereign machine at the vacuum sensor for example with a 70 cm bottle and an unoccluded flow rate of 30ml/min. It is interesting to note therefore that the un-occluded vacuum level generated in the *peristaltic* machine *decreases* with increasing bottle height (bottle pressure) and *increases* with the system flow resistances and *increases* with increasing flow rate.

The pressure loss P(loss) below the bottle pressure Pb, along Ri as a result of the flow rate of $5x10^{-7}$ m³/s, from equation 10 is:

$$P(loss) = 7.36 \times 10^9 \times 5 \times 10^{-7} = 3680 \text{ N/m}^2 \text{ or close to } 27.6 \text{ mmHg}$$

Therefore *the pressure in the eye* with a bottle height of 70cm and a flow rate of 30ml/min and the typical flow resistances of Ri and Rt is:

Pe = 51mmHg - 27.6mmHg = 23.4mmHg.

If there is any leakage flow IL, the eye pressure is further reduced by an amount equal to the product of the leakage flow rate IL and the irrigation resistance Ri. Therefore the eye pressure with leakage included is:

Pe = Pb - Ri.Ip - Ri.IL (Eye pressure equation Peristaltic machines)

Equ.13

As can be seen from equation 13, *the irrigation resistance Ri is the important factor which lowers the eye pressure* in conjunction with either pump flow or wound leakage flow.

Peristaltic phaco machines are configured as noted above, to operate such that if the measured vacuum at the vacuum sensor near the pump climbs to the maximum value the surgeon sets on the panel, the pump is switched off (Stopped). When the vacuum falls below this value, the pump is switched on again. Examining figure 2 again, provided the vacuum Pv generated by the pump near the vacuum sensor does not exceed the value on the machine's panel (programmed by the surgeon) the pump will not be stopped or interrupted and the flow rate will be maintained at, or close to, the panel programmed value. The aspiration flow of course is controlled up to the set maximum on the panel by the foot pedal.

In the peristaltic machine, the eye pressure formula, equation 13, is *not dependent on the aspiration pathway resistance Ra.* When the Ra is increased in the peristaltic system the generated vacuum for a given flow rate increases, because Rt increases. This limits the values of increase in Ra that are permissible while still being able to set low vacuums on the machine's panel, without the pump flow being interrupted for the lower maximum vacuum settings on the machine. Interrupting the pump reduces the average flow rates which have heating consequences in phaco mode when heat is being generated by the phaco probe's crystals.

You can think of the Peristaltic pump as a "current sink" which "sinks" or removes a programmed current or flow from the fluidic pathway, which you can control with your pedal, up to a set maximum value. The peristaltic pump by generating (or sinking) a flow, ends up generating the vacuum as a consequence of the flow rate and system flow resistances. The value of this vacuum is determined by the aspiration flow rate Ia and the total flow resistance Rt and the bottle pressure Pb as given in equation 12.

Flow restrictive devices in the aspiration pathway, or smaller internal diameter phaco needles, both which increase Ra and therefore Rt, will result in an increase in the un-occluded vacuum level at the machine. At occlusion the vacuum climbs from the un-occluded value to the maximum occlusion vacuum specified on the control panel and then the pump is stopped. The time course of this event is known as the "rise time". Both un-occluded and occluded vacuum levels are important in lens fragment holding forces which will be discussed.

Varying degrees of phaco needle occlusion occur when emulsifying the lens fragments and the vacuum levels fluctuate between the un-occluded and occluded levels when that occurs. The flow rate also fluctuates because with near total occlusions the flow resistance is so high that the un-occluded vacuum levels rise above the panel setting and the pump is inhibited. Fluctuating flow rates result in fluctuating anterior chamber pressure. At zero flow rate the eye assumes the bottle pressure with time and with un-occluded flow assumes the bottle pressure less the value of the flow rate and irrigation system flow resistance as per equation 13, ignoring any wound leakage.

The vacuum generated at any time by the peristaltic pump is not perfectly smooth and fluctuates as the pump rollers strike the peristaltic pump tubing generating positive pressure pulses which transiently reduce the vacuum. This can occasionally be observed in the anterior chamber as regular fluctuations in the geometry (chamber volume) with the iris plane moving up and down towards and away from the surgeon's microscope. This fluctuation corresponds to varying anterior chamber pressure, and varying anterior chamber depth. This problem is not of great surgical importance however it is aggravated as the compliance of the peristaltic pump tubing and aspiration tubing is reduced, even though this reduces the post occlusion surge amplitude as will be explained later. These pulsations can be reduced to a lower amplitude and higher frequency pulsation by using smaller diameter peristaltic pump heads and short lengths of smaller diameter pump tubing, so that the pump head runs at a higher rotational speed for any given average flow rate.

It can be seen from equation 13, ignoring leakage, that the eye pressure will fall to zero when the pump flow is such that: Ip x Ri = Pb. In other words at a flow rate where the entire bottle pressure value Pb, is lost (dissipated) by the irrigation flow resistance. With a Pb of 51mmHg (6798N/m²) and the Ri of 8.61×10^9 (60ml/min Ri value with annulus in the eye) this corresponds to a flow rate of around 7.89×10^{-7} m³/s or 47 ml/min. Therefore the anterior chamber pressure will fall near to zero if you could set the pump to that flow rate with the bottle at 70cm. Therefore flow rates around 50ml/min can only be supported if the bottle height is higher than 70cm (28 inches) with typical irrigation apparatus.

When occlusion occurs Ra and therefore Rt climb to very large values and with total occlusion would be an infinite number because no flow is possible. The vacuum builds in the aspiration system due to the pump removing fluid from it. This flow can occur with total occlusion, for a time because the aspiration system is elastic. Over this rise time, atmospheric pressure compresses the compliant parts of the system on the aspiration side. The vacuum climbs until the vacuum at the machine's vacuum sensor and in the aspiration tube equals that set on the machine's panel and then the pump is instructed to stop.

With incompletely occluded flow and at high maximum vacuums which can occur with high flow resistances and partial occlusions, or flow restrictive devices, there is some compression of the peristaltic pump tubing by atmospheric pressure, and therefore there is less pump output per cycle of the pump rollers. This because the pump tube lumen diameter is a little smaller and holds a smaller volume of fluid. The flow rate can fall a little below the programmed value at high vacuums. This can easily be corrected by increasing the pump speed proportionally the measured vacuum. This has been called "volumetric compensation".

It should be pointed out the usual reason for accelerating the pump velocity with increasing vacuum, due to occlusion, is to shorten the "rise time" to reach the maximum programmed panel vacuum. On the whole the reduction in pump flow with occlusion and rising vacuums is not a significant problem, this is because higher vacuum levels which substantially compress the pump's tubing, over 150 to 200mmHg, do not generally occur without a good degree of occlusion when the actual flow is low or near zero and the flow rate is therefore academic. Over the years there has been a move to shorten the rise time to maximum vacuum in the peristaltic machine to help make the lens fragment holding behaviour of the machine better resemble that

of the venturi machine. However the only way to make the initial lens holding forces at occlusion in the peristaltic machine, similar to those in the venturi machine, is to have high flow rates *prior* to occlusion. We will discuss lens fragment holding forces in the section on *dynamic* fluidics.

The use of low maximum allowed vacuum settings, 15 or 20 mmHg, and a flow rate of 30ml/min, on peristaltic based machines, for sculpting the lens results in interruption of the pump's function. This is because the generated vacuum level equals or exceeds the programmed maximum vacuum value and the pump is therefore inhibited. The machine therefore displays erratic vacuum measurements. The actual flow achieved is nowhere near the 30ml/min or thereabouts which might be programmed on the machine's panel. The measured vacuum has to fall to a lower value before the pump is switched on again.

The pump takes an interval of around 100 to 200 milliseconds to resume motion again after instructed to do so, therefore the average vacuums are lower than the panel maximum values. The average flow rate is therefore much lower than programmed 30ml/min (for example) when the pump is continually interrupted and so there is proportionally less probe crystal and phaco needle cooling and higher probability of wound burn.

Adding aspiration resistance, or increasing Ra in the peristaltic machine, has been tried for a number of reasons. The first is to damp down and reduce the post occlusion surge. The second is more of a side effect or consequence of seeking smaller incision sizes for *micro-coaxial* micro incision surgery to compete with *bi-manual* micro incision cataract surgery with an irrigating chopper needle. Referring to the simple flow circuit of figure 2 again and the vacuum equation 12, increasing Ra, which increases Rt, results in a higher vacuum being associated with the same flow rate. This rule applies whether it is a peristaltic pump which generates the vacuum as a secondary consequence of the flow rate, or a venturi machine, or electric vacuum machine which generates the vacuum directly. In the case of the peristaltic machine, then if the maximum allowed vacuum set on the machine's front panel is lower than the generated vacuum at the selected flow rate then there will be continuous intermittent interruption of the pump's rotation and the flow rate will be well below that selected on the control panel. In other words increasing the aspiration flow resistance Ra prevents the use of low vacuum levels while at the same time still maintaining normal flow rates which help cool the phaco crystals and needle.

In the IA mode, on the peristaltic machine, with the addition of the very large flow resistance, $6x10^{10}$ to $1x10^{11}$, over the range of 30 to 60ml/min provided by the small port in the IA needle is of no consequence, as no significant heat is being generated. Also in this mode the machine is usually run at a maximum vacuum of 500mmHg and the flow is not significantly attenuated. This small highly resistive port allows the use of maximum vacuums near 500 mmHg without significant post occlusion surges.

Along with small calibre phaco needles and sleeves for micro-coaxial phaco and along with the increased irrigation flow resistances Ri due to the smaller architecture of the annulus, has also come an increase in the aspiration resistance Ra because the phaco needles have a smaller internal diameter than standard needles.

In the case of the venturi machine the problem is similar. Increasing aspiration flow resistance Ra means on the one hand that higher aspiration vacuums can be used because the flow rate for any given vacuum is lower. However at times when the vacuum levels are low the flow rate falls very low too, again exacerbating the problem of wound burn. To some extent these problems in both machines have been partially offset, but not solved, with pulse mode ultrasound which helps lower the phaco needle temperature.

Typically for the peristaltic machine, the first position on the pedal opens the irrigation pinch valve, the second or middle position linearly controls the flow rate up to the preset panel maximum, and the third position controls the phaco (ultrasound) energy. The surgeon can alter these with the pedal position preferences. The peristaltic machine can also be configured so the middle pedal position is such that rather than linearly controlling the flow rate up to a maximum, the pedal now linearly controls the measured vacuum, at the vacuum sensor, up to a set panel maximum vacuum. The flow rate under these conditions is whatever is required to generate the specific vacuum level at the vacuum sensor according to the pedal position and the set maximum vacuum on the panel. This results in "similar" performance to a venturi machine. This is called the "linear vacuum mode", versus the usual "linear flow mode", and it confers no or little advantage to the peristaltic machine over the venturi machine, except for higher unoccluded vacuums and flow rates which improve lens fragment holding forces. In peristaltic machines, that have this option, be very careful if you switch to linear vacuum (or "venturi mode") that you run a low maximum vacuum, less than 140 to 180mmHg, or just as in venturi based phaco, without flow restrictive devices in the aspiration system, your anterior chamber will quickly collapse. This due to the high flow rates generated by those vacuums and the associated pressure loss in the eye due to irrigation pathway flow resistances subtracting from the bottle pressure as outlined above. For example if your maximum vacuum in peristaltic mode is set to 250mmHg and you switch to venturi mode, the anterior chamber may rapidly collapse as it will be taken significantly below atmospheric pressure.

Venturi Machine – Constant Un-occluded flow circuit modelling:

The flow circuit for the venturi machine is shown in fig 3.



In this case the diagram is similar to the peristaltic machine however the vacuum is generated directly by the venturi device (or electric vacuum pump) which can also be regarded as a pump which picks up the fluid from a negative pressure value below atmospheric pressure, and raises the fluid pressure to zero or atmospheric pressure. Ignoring leakage flow, the aspiration flow rate Ia in this system is dependent on the sum total of the bottle pressure Pb and the vacuum level Pv applied across the total of (Ra + Ri) or Rt. With a bottle generating a Pb of 6798N/m² (51mmHg) and the vacuum being 7652N/m² (57.4mmHg) and Rt being 2.89x10¹⁰ the flow rate will be $5x10^{-7}$ or 30ml/min.

In general the leakage flow acts to lower the anterior chamber or eye pressure Pe, in proportion to the leakage flow rate. In the venturi machine system the effect of leakage flow is more complicated than for the peristaltic machine. The leakage flow in the venturi machine case is *sourced* from the eye, which is linked to two pressure sources, albeit one called a "vacuum" via irrigation resistances and aspiration resistances Ri and Ra respectively. For this analysis we require the formula for two resistances in parallel, Horowitz⁵:

$$\frac{1}{R(total)} = \frac{1}{R1} + \frac{1}{R2}$$
Equ.14

Therefore the total for Ri and Ra in Parallel =
$$\frac{\text{Ri.Ra}}{(\text{Ra} + \text{Ri})}$$
 or;

$$=\frac{\text{Ri.Ra}}{\text{Rt}}$$
Equ.15

The above is the general formula for the net effect of two resistances in parallel. When a current, or flow is extracted from a resistive divider network which is placed across a pressure source (in this case the sum of the bottle pressure and vacuum sources), then the system behaves as though there is one resistance, the Thevenin resistance, Horowitz⁵. This is equal to the two resistance values in parallel, connected to an equivalent pressure source which is a divided down pressure by the two resistances (the Thevenin pressure). For example the Thevenin pressure in this case is the eye pressure at the equilibrium or un-occluded flow rate eye pressure at the junction of the two flow resistances. The anterior chamber pressure loss due to leakage is therefore the leakage flow rate multiplied by Ri.Ra / Rt), in other words IL.Ri.Ra / Rt. This was not the case for the peristaltic machine because the aspiration flow rate, controlled by the peristaltic pump, was independent of the leakage flow rate and eye pressure. In the venturi arrangement the aspiration flow rate *is* affected by the eye pressure.

To summarise the pressure within the anterior chamber Pe, in a "constant flow" situation, with a venturi or vacuum machine, is the bottle pressure, less the pressure drop due to aspiration flow and less the pressure drop due to the leakage flow. The pressure in the eye Pe therefore is:

$$Pe = Pb - Ri.Ia - IL. \frac{Ri.Ra}{Rt}$$
Equ.16

Aspiration flow Ia, depends on the sum of the applied driving pressure gradient, Pb and the vacuum level Pv divided by the total system resistance Rt;

$$Ia = \frac{Pb + Pv}{Rt}$$

In addition;

Equ.18

Using the relations 16 through 18 above algebraically then the eye pressure Pe for un-occluded flow with the venturi machine is:

 $\frac{Ra}{Rt} = (1 - \frac{Ri}{Rt})$ because Rt = (Ra + Ri)

*Equ.*17


It is interesting to compare this with the eye pressure equation 13 for the peristaltic machine which is simpler. Converting these large resistance values to relative resistance helps deal with the large numbers and the ratios are dimensionless. For example using the 30ml/min resistance values from table 2 and using the B&L Micro flow Plus needle and sleeve used in Venturi work then $Ri = 6.50 \times 10^9$, $Ra = 2.39 \times 10^{10}$ and Rt therefore 3.04×10^{10} so Ri/Rt = 0.21 or 21% and Ra/Rt = 0.79 or 79% for the fluidics system outlined in table 2. So one can see there is little relative resistance on the irrigation side and compared to the relative aspiration resistance.

These resistance percentages above are very useful numbers to know as they enable a quick mental calculation of what the pressure in the anterior chamber might be with a certain bottle height and set vacuum, using a *venturi* based phaco machine at least if the flow resistances are similar to those outlined in the table 2. Equation 19 therefore tells us (ignoring wound leakage, therefore IL = 0) that the eye pressure Pe, is approximately 79% of the bottle pressure, minus 21% of the vacuum or roughly:

Pe = 80% Bottle - 20% Vacuum (Rule of thumb eye pressure equation- Venturi)

Equ.20

So the 80:20 is easily remembered. In the case of the venturi machine the flow rate is determined by the irrigation and aspiration resistances and the overall applied driving pressures which are the bottle and vacuum. At a high flow rates around 60 ml/min both the irrigation resistance Ri and aspiration resistance Ra have increases somewhat. The ratios of Ri/Rt and Ra/Rt stay fairly constant. Therefore using equation 20 above and with a bottle pressure of 6798 N/m² (51mmHg), then the vacuum which takes the eye pressure to zero is 25573 N/m² (191mmHg vacuum). If any flow restricting devices (resistive devices) are added to the aspiration tubing then Ra is increased, this increases Ra/Rt, and this decreases Ri/Rt (because Rt is increased). Therefore with the flow restrictive device, for any given vacuum, the flow rate will be lower and Pe will be higher. This is the basis of long lengths of aspiration tubing added to the aspiration line, or flow resistances or flow restrictor devices. Other methods include aspiration tubing with higher flow resistance due to a smaller lumen, which increases the flow resistance, or smaller internal diameter phaco needles with a higher square law flow resistance. Using these devices Ra is increased and the anterior chamber pressure will be higher for a given machine vacuum because the flow rate will be lower for a given vacuum. This reduced flow rate results in a smaller pressure drop along the irrigation resistance Ri, so Pe remains at a more positive value closer to the value of the bottle pressure. The compromise however is that of reduced flow rates at low vacuum levels and if lower vacuum values are used, without

removing the flow restrictor, this reduces phaco probe crystal cooling and increases the risk of wound burn in phaco mode.

Analogous Flow Circuit Systems:

The loss of pressure with a constant flow along the irrigation resistance originating from the bottle's pressure source has a very interesting analogy here: When you draw electrical current flow from a battery (which is a voltage or electrical pressure source), the terminal voltage (or electrical pressure) drops, due to the battery's internal resistance. This is why lead acid batteries, of very low internal resistance are used to start your car. Otherwise the large current flows, drawn from them by the starter motor, would cause the battery's terminal voltage (or terminal pressure) to collapse excessively.

In the case of phaco fluidics most of the *internal resistance* equivalent "in series" with the bottle pressure source is the combination of the irrigation tubing and the phaco-needle and sleeve annulus arrangement, or irrigating chopper lumen, as can be seen from the resistance figures in table 2. Fluid flow resistance is analogous to electrical resistance except that fluid flow resistance dissipates heat with a flow of fluid and electrical resistance dissipates heat with a flow of electrons. Both of these processes dissipate heat and lower the potential energy. This potential energy is Pressure and Voltage respectively for the two physical systems.

In the case of fluidics the pressure "potential energy per volume of fluid" (Joules / m^3) reduces along the resistive flow pathways. In the electrical analogy the pressure equivalent is the "potential energy per volume of electrical charge" known as Voltage. Voltage (units Joules/Coulomb or Joules/volume of charge) reduces along the electrical resistive flow pathway. In fluidics the volume units are cubic meters of fluid. The volume units of charge in the electrical system is an amount of charge Qo, the Coulomb, which is a large quantity of electrons: $6.25x10^{18}$. Poiseuille's law is in fact a version of Ohm's law, where the electrical resistance R, of Ohm's law is represented by $8.\eta.1/\pi r^4$ of Poiseuille's law.

CHAPTER 3.

(3-1) DYNAMIC PHACO FLUIDICS - APPLYING FLOW CIRCUIT MODELLING UNDER CHANGING FLOW CONDITIONS:

We are now ready to move into the more complex area of *fluidic dynamics* which describe *dynamic* rather than *static* flow situations described above. The equations presented above explained what the anterior chamber pressure would be during constant un-occluded flow conditions with a given flow rate, or vacuum level and bottle height when using the disposables and irrigation setup described by the flow resistance values in table 2. However, as useful as that is, it tells us nothing of the dynamic behaviour of the phaco fluidics system when the flow is stopping and starting. The analysis and investigation of this dynamic behaviour, presented here has yielded some important insights in the nature of anterior chamber instability during cataract surgery.

Bernoulli's Equation:

The Bernoulli equation is one of the most well known equations of fluid flow. The equation is a statement about the conservation of energy. It aids in the understanding of this equation that pressure, as pointed out in (1-4), is in fact an "energy density" or energy per unit volume quantity. Kinetic energy (or energy of motion) can also be expressed per unit volume then we call it "kinetic energy density".

Bernoulli's equation states:

Applied pressure + any gravity pressure + kinetic energy density = a constant.

Or in symbolic terms:

$P + \rho.g.h + \frac{\rho v^2}{2} = k$	\mathbf{v} = average fluid velocity ρ = fluid density, g = gravity, h = height
Equ.21	P = Applied Pressure k = a constant
	p.g.n - gravity pressure.

The Bernoulli equation applies to a *lossless system* (no resistive frictional losses) with *rigid* walled vessels at a constant flow rate with uniform streamline (laminar) flow. The Bernoulli equation shows that the applied pressure P and any pressure contribution from the position of the fluid in the gravitational field ρ .g.h, and the kinetic energy density of the fluid $\rho v^2/2$, remains constant. The Bernoulli equation is a useful equation and it explains that as fluid passes via a narrow aperture and the velocity of the fluid increases then so does the kinetic energy density there and therefore the pressure must drop as a consequence. In other words some of the

pressure energy is converted to kinetic energy density, but the total remains a constant. This explains the Venturi effect for example. The equation therefore shows how the fluid has "kinetic energy density" *exchangeable* with pressure or "potential energy density" depending on the fluid's velocity. This concept will help us later in understanding how a fluid's kinetic energy is converted to a pressure when moving fluid in a tube is stalled at a flow occlusion for example.

The Bernoulli equation however is severely limited on its own and won't do for phaco fluidics because a number of important features are missing. These are the pressures associated with *Compliant* (elastic) chambers in a fluidic system. In addition the energy dissipating effects *linear resistance* as described by Poiseuille's equation and *square law resistances* calculated with K values and the Darcy Wiesbach equation are absent. Also the Bernoulli equation is for "constant flow" or "equilibrium flow conditions" and explains little of the processes involved in the time, from when pressure is initially applied, to equilibrium flow being reached. Although the kinetic energy density relates to the fluids' inertia, the equation gives little clue as to the behaviour of a mass of fluid when it is being accelerated and decelerated. We therefore require an equation and solution system for *dynamically* changing flows which include the three properties of *inertia* of the fluid contained in the fluidic system objects, the *compliance* of the elastic structures of the system objects and the flow *resistances* of the system objects.

A suitable model and analysis addressing these issues has been developed by the author to suit the unique properties of phaco fluidics. This has been achieved with "flow circuits" as described above in conjunction with the basic equations for the three important properties. The result creates the basic structural equations for "*dynamic phaco fluidics*". These tell us what is happening to the eye pressure and therefore anterior chamber volume during rapidly changing flow situations. Also these equations, exactly as one might expect, reduce to the constant flow equations described above when the flow rates and pressures and vacuums are constant and not changing with time.

The dynamic phaco fluidics equations, presented in this book can be directly solved for a solution as a numerical value or can be run as a graphical solution from the flow circuit equivalents in a Circuit Simulator which gives identical results. The graphs are certainly more reader friendly. The circuit simulator becomes very useful when the equation solutions become enormously bulky, which they tend to be due to the complexity of the system. Generally the equations are useful up to the point of analysing an irrigation or aspiration system and eye together on their own. When the irrigation system and aspiration systems are linked, both of which have different resonant and damping properties, the equations describing them are too cumbersome and the circuit simulator is not a luxury, it then is essential.

The equations themselves, as it turns out, provide enormous insight into the behaviour of our phaco fluidics systems. It also provides a new methodology to objectively design, test or verify the performance of fluidic instrumentation devices, and therefore improve anterior chamber stability during surgery and therefore patient safety.

In this book the approach to help explain the *dynamics* of phaco fluidics is to first describe the *irrigation* pathway and its dynamic behaviour and limitations. Then describe the *aspiration*

pathway with its dynamic properties and then combine the two systems simultaneously, as they are in practice, working together.

When the *irrigation* and *aspiration* dynamic systems are combined the system becomes "one network" or a fluidic system where everything is interlinked. Understanding the irrigation and aspiration system separately does have some merit as it helps explain some features of the performance of the overall system. The overall pressure changes involved on the irrigation side are relatively small, and at their greatest only about equal to the bottle height pressure. The compliance in the irrigation tubing does have some small measurable effects and these are detectable in experimental recordings. This is not the case with the eye itself where the eye's compliance has significant effects, or indeed on the aspiration side where the compliance of the eye, aspiration tubing, machine pump tubing and vacuum sensor assembly(in the peristaltic machine) have a significant effect on the fluidic properties. These effects are due to significant changes in volume of these objects over the pressure ranges that they experience and the fact that the volume of the anterior chamber of the eye is small in the order of 0.25 ml. Flow circuits will be again used to symbolise the system components and simplify the concepts and find the solutions. The reader is now familiar with the basic constant flow circuit of figure 2 and 3 incorporating the flow resistances so now this can be added to, as required, to stop and start the flow and incorporate the *compliant* structures and the *inertia* of the fluid within the fluidics system.

(3-2) THE IRRIGATION SYSTEM - DYNAMIC BEHAVIOUR.

Firstly we consider a standard bottle and irrigation setup only with an on off-tap running to free air. This could be the irrigation tube pinch valve in the phaco machine is depicted in figure 4. The probe is out of the eye as it would be in an irrigation free flow test.





Figure 5 shows the equivalent flow circuit where the switch S1(or pinch valve opens) is switched on to start the flow at a time t = 0 zero. Pb is the bottle pressure source calculated from ρ .g.h and if the bottle height is 32.5cm then the pressure is 3156N/m² or 23.7mmHg. This pressure gradient drives the fluid through the 3.5m of irrigation tubing the fittings probe and yellow needle/sleeve which has a total resistance together of Ri, from Table 1, of 6.32×10^9 @ 30ml/min with the annulus out of the eye. We know that when the flow has reached a stable value some time after being switched on, that the flow rate will be $3156/6.32 \times 10^9 = 5 \times 10^{-7}$ m³/s or 30ml/min.

If we now consider a time at t = 0 when the flow is turned on and we look at the flow circuit of figure 5: It does not tell us anything at all about how the flow builds up with time to the equilibrium flow rate of 30ml/min. Figure 5 suggests that the flow will begin immediately the switch is closed. We know this can't be true in practice because the fluid in the tubing has a mass and it takes a while after the pinch valve is open (or S1 closed), for the unbalanced force (pressure/unit area) acting on the fluid mass to accelerate the fluid to its equilibrium flow rate.

When the flow rate stabilises the fluid has a constant average velocity **v**. Therefore figure 5 is only any use to explain what the flow rate will be *some time later* after the equilibrium flow rate has been reached and clearly something else is missing. The time delay, whatever it is, in accelerating the fluid up to a constant velocity is important in determining the *dynamic* behaviour in a fluidics system. What we wish to know now is how the flow I changes with time after the flow is switched on, denoted by a function I(t).

By the time the flow is stable the kinetic energy of the fluid with a constant average velocity \mathbf{v} moving in the tubing corresponds to the additional energy taken to accelerate the fluid up to a fixed velocity (constant flow rate). At this constant flow rate the driving pressure gradient in this case Pb of 23.7mmHg is exactly opposed by the forces of friction to fluid movement (flow resistance) and the fluid stops accelerating and reaches a constant velocity. Prior to this time to reach the stable flow rate, or flow velocity, the mass of fluid in the tube is "opposing" the applied pressure with a "*reactive pressure*" (For every action there is an equal and opposite reaction; Newton). The fluid mass while it is being accelerated therefore *generates a reactive*

force or *pressure* opposing the applied (bottle) pressure Pb. This relates to the *inertia* (resistance to change in motion) of the fluid in the tubing. When the fluid reaches a constant velocity or constant flow rate the reactive pressure or Inertia "vanishes" and the forces of friction balance with the applied pressure. Also then the rate of work being done by the pressure source equals that being dissipated by the flow resistance.

The reactive pressure which is a function of time t, we will call Pr(t) for now: a *pressure varying with time* and it is the result of the fluid's inertia. Pr(t) therefore subtracts from the applied pressure or bottle pressure Pb in our Ohm's Law formula during the build up of flow:

$$Pb - Pr(t) = I.R$$

Equ.22

In a very small time increment after S1 is closed (t = some nanoseconds) there is no flow I of any note established yet because the value of Pr(t) equals Pb. Much later when the flow is stable equation 22 reduces to Pb = I.R when the flow I is not changing with time and the Pr(t) function, whatever it is, is zero. Now we need to determine the reactive pressure function with time:

Inertia relates to Newton's first law that a mass will not move unless acted on by a net or "unbalanced force". In general the faster you attempt to change the velocity of a mass (accelerate or decelerate it) the more force is required to do it. There is a relationship we all know given by Newton's second law F = m.a. (force = mass x acceleration).We can regard inertia as the reluctance of any physical object with mass to change its velocity. If we consider a pressure gradient acting on some fluid in a tube with a pressure P1 at one end which is greater than P2 at the other end, the driving pressure is P1-P2 or simply P. If we multiply this pressure by the cross sectional area of the lumen we can obtain the force F, acting on the mass of fluid in the tubing such that:

P x Area = Mass x Acceleration, (as in F=m.a)

The mass of the fluid is equal to the volume of the fluid in the tube multiplied by the fluid's density ρ . The volume of the fluid in the tube is the area of the lumen multiplied by the length of the tube.

For the fluid filled tube we now have:

$$P =$$
 Length x Density x Acceleration.

Equ.23

Which describes the pressure gradient across the fluid in the tube and we can think of the pressure as reactive because the fluid mass opposes the acceleration or change in velocity.

Therefore we may re write equation 23 in symbolic terms as:

$$Pr(t) = \frac{\rho \cdot l}{dt} \frac{d\mathbf{v}}{dt}$$
Equ.24

Where l = length of the tube and $\rho =$ density of the fluid in the tube and dv/dt is the rate of change of velocity with time (or acceleration) of the fluid at any time in the limit where the small interval dt is approaching zero. Pr(t) is therefore the reactive pressure at any time t.

Leaving equation 24 for a moment, we return to the useful equation 1 again:

$$\mathbf{v} = \frac{\mathbf{I}}{\pi r^2}$$

Over a small time increment dt, there is a small velocity change $d\mathbf{v}$, therefore we obtain from equation 1 above:

$$\frac{\mathrm{d}\mathbf{v}}{\mathrm{d}\mathbf{t}} = \frac{\mathrm{d}\mathbf{I}}{\mathrm{d}\mathbf{t}} \cdot \frac{1}{\pi r^2}$$
Equ.25

Therefore substituting equation 25 for $\frac{d\mathbf{v}}{dt}$ into equation 24 we now have the:

(3-3) REACTIVE PRESSURE FLUIDIC EQUATION:

$$\Pr(t) = \frac{\rho . l}{\pi r^2} \cdot \frac{dI}{dt}$$
Equ.26

Where r is the radius of the tube lumen, ρ is the density of the fluid and *l* the length of the tube in question and dI/dt the rate of change of flow with time.

The Reactive Pressure fluidic equation is merely the "fluidics equivalent" of F= ma and has been derived by with the reasoning above.

System analogies:

Again we strike a very interesting analogy with electrical systems in that the reactive Voltage (or pressure) Vr(t) induced in an electrical inductor or coil⁵ is:

$$Vr(t) = \frac{L}{dt} \frac{dI}{dt}.$$
 Equ.27

In this case I symbolises electrical current rather than fluid flow. This reactive voltage Vr(t) opposes the applied voltage in the same way that the reactive pressure of the fluid in our tubing opposes the applied pressure. This makes the $\rho . l/\pi r^2$ of the fluidics system analogous to Inductance L in the electrical system. It is easy to see how this is the case: An electrical inductor opposes a change of current flow, the faster the current attempts to change value, the greater the reactive or opposing voltage (or electrical pressure) generated by the inductor. A mass of fluid on the other hand opposes a change in velocity or flow rate, the greater the attempted flow rate change the greater the opposing or reactive forces.

We will call $\rho . l/\pi r^2$ the "Reactive Pressure Coefficient" for fluid filled tubes, however it works for an annular area too if πr^2 is replaced by the cross-sectional area of the annulus in question. With this in mind the reactive pressure coefficients of fluid carrying objects in the phaco fluidic system will be called "L values" for want of any better symbol to use.

In our fluidics reactive pressure equation 26, the L value or ρ .*l*/area, is the coefficient of the "rate of change of flow" dI/dt. This means that we can think of ρ . *l*/ π r² as being a form of *Dynamic Resistance* or *Impedance* to flow which only comes into play, as a factor in fluidics, where there is some rate of change of fluid flow with time and therefore dI/dt is not zero.

Also we can conclude that with any *constant* flow situation eg dI/dt = 0, that the reactive pressure effects or *impedance* to fluid flow vanishes. For typical fluid carrying objects in phaco the "L" values are calculated for the reader and listed in the table 2. *Note that the L values are calculated values from the geometry of the fluid carrying objects*.

In addition if we take the kinetic energy density (energy per unit volume) from the Bernoulli equation $\frac{\rho v^2}{2}$, and substitute equation 1 in for flow rate we have $\frac{\rho I^2}{2(\pi r^2)^2}$.

We then convert this to a total energy, for a volume of fluid in a tube, length l and radius r: Therefore multiplying by $(l.\pi r^2)$ the volume of the fluid in the tube, we then have the kinetic energy Ek of the fluid flow:

Ek (constant fluid flow via a tube) =
$$\frac{\rho l}{\pi r^2} \cdot \frac{I^2}{2}$$

Equ.28

Equation 28 shows that the *kinetic energy* of the fluid within a tube of length l is proportional to the square of the flow rate I divided by 2. It is very interesting to note that the energy of the magnetic field stored in an inductor of value L with a constant flow rate I is⁵:

Ek (electrical current I, flow via an inductor) =
$$\frac{L}{2}$$

Equ.29

This again agrees exactly with the findings of equations 26 and 27 that the value $\rho . l/\pi r^2$ (the reactive pressure coefficient) in a fluidics system is the analogy of L or inductance in the electrical system. The usefulness of this finding relates to the fact that the mathematics and computer modelling for electrical systems and their dynamic behaviour is highly developed and can simply be pressed into service to find solutions to the fluidics system problems.

Shape of the time course to reach fluid flow equilibrium after flow begins:

Returning to the equation 22 and incorporating the Reactive Pressure Equation for Pr(t) equation 26 and using the symbol L to represent $\rho l/\pi r^2$ we now have:

$$Pb - \frac{L}{dt} = R.I$$
Equ.30

Re arranging the above:

$$Pb = \frac{L}{dt} + R.I.$$
Equ.31

Each object in the fluidics system has an L value on account of the fluid mass it contains. Like the resistance values which add together in series the L values add up to a total L value too. The equivalent flow circuit of this is shown in figure 6. The total Ri and Li values for a typical phaco irrigation system taken from table 2.



Now a *coiled symbol* represents the total L value of the fluid in the irrigation system Li, and the zig-zag shape the total fluid flow resistance of the irrigating system Ri.

Pb again is the applied bottle pressure. Switch S1 again represents the machine's pinch valve to stop and start the flow. We are interested in I(t), or the flow of fluid with time. Equation 31 is not in convenient form needs a solution we can graph. The solution to this type of differential equation for the flow with time I(t) is available from many texts^{5,6} and is a typical inverted exponential:

$$I(t) = \frac{Pb}{Ri} (1 - e^{-t.Ri/Li})$$

Equ.32

The plot of equation 32 gives the flow- time diagram Fig 7



Figure 7 was conveniently plotted by the Circuit Simulator. A hand plot from the equation 32 gives identical results. The irrigation apparatus total Li (total L value of irrigation system) from the table for the AMO yellow sleeve/needle is 5.66×10^8 and the flow resistance without the annulus being compressed (out of the eye) is 6.32×10^9 @ 30ml/min. The bottle is again set at 32.5cm to obtain a flow rate of 30ml/min. It is easy to see that after a time the flow rate will stabilise at Pb/Ri or $3156/6.32 \times 10^9 = 5 \times 10^{-7}$ or 30ml/min. By 500 milliseconds the flow is very stable and the reactive pressure has decayed away. After this time the flow is exactly as described by Ohms law, just as it is for any stable flow calculation. The scale on the y axis of the graph is in *nano* units eg $500n = 500 \times 10^{-9}$ cubic meters/sec. The reactive pressure Pr(t), not to scale, is plotted on the same graph to show how it decays away with time to zero when the fluid flow rate becomes stable. We will be looking at *reactive pressures* again in the section on lens fragment holding forces.

Leading up to stable flow rate, the flow is increasing in the profile of an inverted exponential form. The "time constant" (time to reach 63 % of its stable flow value) is Li/Ri seconds or 89 milli-seconds.

We now have a system to model how fluid flow, under a constant applied pressure gradient, builds up its flow with time from a standing start to a stable flow. It can be seen from figure 7 that it takes 300 to 400 milliseconds to get close to the stable flow rate of 30ml/min.

It can also be seen that a useful feature of the "circuit" model of figure 6 represented by equation 32 is that the pressures around a flow "circuit" always add to zero. In other words the applied pressure must always equal the sum of the reactive pressure and the pressure developed across the resistance by the flow I or the I.R value. The pressures around a closed circuit must always add to zero. This is another advantage of the *flow circuit concept* in the analysis of phaco fluidics. Kirchoff's second Law states exactly this in relation to closed electrical circuits. This rule also applies when other elements for example resistances, L values or compliant structures are added to the circuit as will be shown.

We had previously noted that the *linear* resistance to fluid flow in a tube is inversely proportional to the 4th power of the tube's radius and the same applies to *local* constrictions in the fluidic pathway with *square law* resistance. However looking at the L value relating to a fluid's inertia, we see that this is inversely proportional to the square or the second power of the tube's radius. This means that doubling the irrigation tube's internal radius, while significantly reducing the flow resistance of the tubing by 16 times, has much less of an effect on the fluidic inertia, reducing it by a factor of 4.

In considering equation 31 again, we see that although this was not in a good format to plot a graph, it does give great insight into the nature of fluidic inertia. It states that the reactive pressure P at any time t, is proportional to the "rate of change of flow" of the fluid, $\frac{dI}{dt}$, with the L value being the proportionality constant. In other words as the flow rate is increasing, with a positive slope at its steepest, the pressure P will be at its maximum positive value. This has the property that the flow is *delayed* with respect to the applied pressure under *dynamic* or alternating flow conditions. As can be seen in figure 7 the fixed applied pressure is delivered at t = 0 when S1 closes (or the pinch valve opens) but the flow doesn't build up until 400 milliseconds later. This has a profound influence on the timing of inflows, outflows and pressure changes in the eye during post occlusion surges as will be explained shortly. All of the above leads to the three basic rules or ABC's of fluidic inertia.

(3-4) THE ABC's OF FLUIDIC INERTIA:

- A) For alternating or rapidly changing fluid flows and pressures, the flow lags behind the applied pressure in time.
- B) For constant fluid flow rates the effects of fluidic inertia vanishes.
- C) For tubes carrying fluid, the inertia is inversely proportional to the square of the tube's radius and proportional to the length of the tube and the fluid density in the tube.

Summary of fluidic inertia:

From the above we can conclude that the effects of fluidic *inertia* disappear under constant flow or equilibrium conditions and therefore would not feature in constant flow solutions. Under rapidly changing flow conditions where the flow is stopping and starting the inertia is of great importance opposing any sudden changes of flow, and *delaying* the flow with respect to any applied pressure changes.

We are now ready to add the irrigation device to the eye to determine the nature of the fluid flow and pressure in the eye under dynamic circumstances. Before this can be done the fluidic characteristics of the eye must be defined. The eye is a compliant object so we evaluate the property of compliance first:

(3-5) DEFINITION OF COMPLIANCE:

Compliance (symbol Co in this book), of an elastic vessel, which could be a chamber or a plastic tube, is best described as the ratio of volume (Vo) change of the chamber to applied pressure change P within the chamber.

The basic relation for compliance is:

Change in Volume / Change in Pressure = Compliance:

$$Co = \frac{\Delta Vo}{\Delta P}$$
Equ.33

Note that the units of compliance are Volume/Pressure which is $Volume^2/Joules$, or m^6J^{-1} . For example a highly compliant chamber will have a large volume change for a given change in applied pressure, or will have a small change in internal pressure for a large change in introduced or removed volume of fluid.

Now if a very small fluid volume dVo is introduced or removed from a compliant chamber over a very small time dt, there will be a small change in pressure dP. We can write this in symbolic terms as:

$$\frac{dVo}{dt} = Co.\frac{dP}{dt}$$
Equ.34

Note that dVo/dt is now a "flow rate" as the units are Volume/sec. This may be re-written as:

$$I(t) = Co.\frac{dP}{dt}$$
Equ.35

Where I(t) is the flow at any time t. Equation 35 above provides great insight into the nature of the flow and pressure under dynamic or rapidly changing conditions of flow a compliant chamber. Notice how the flow at any time I(t) entering or leaving the chamber, is proportional to "the rate of change of pressure with time" or dP/dt, within the chamber.

In other words when the pressure within the compliant chamber is increasing, for example with a positive slope at its steepest then the flow I will be at its most positive value. The effect of this for *dynamic* alternating flows entering or leaving the chamber is to cause the flow variations to be phase advanced or be *early* with respect to the pressure variations. Or stated another way: Rapidly changing flow variations cause a pressure change within the compliant chamber that are late, or delayed in time with respect to the flow. This feature is of great importance in phaco fluidics. In terms of time delays between pressure and flow this is exactly the opposite property to that for the fluidic inertia.

Also from equation 35 we can see that if there is a constant flow rate (I = a constant value) entering *or* leaving the chamber, then the rate of change of pressure in the chamber dP/dt, is constant, in other words the pressure then increases or decreases linearly with time. Also if we fill continuously a compliant chamber at exactly the same rate we are emptying it, or in other words the compliant chamber is part of a In - Out fluid flow pathway, this makes the average flow I, equal to zero, then Co. dP/dt = 0 and the effects of compliance vanish. So in the same way the effect of fluidic inertia vanishes with constant flow situations, so do the effects of compliance vanish when the inflow and outflow to a compliant chamber are equal, or the flow is passing through the chamber at a constant rate, as it does in the eye with un-occluded flow during phaco cataract surgery.

Therefore, from one simple equation 35, we can deduce the three important properties of compliance, or the ABC's of Compliance.

(3-6) THE ABC's OF COMPLIANCE

- A) For alternating or rapidly changing fluid flows and pressures, the alternating compliant chamber pressure is delayed in time with respect to the flow entering or leaving the chamber.
- B) If a constant fluid current (constant flow rate) is added *or* removed from a compliant chamber, then the rate of change of pressure in the chamber during this time will be a linear increase or a linear fall respectively.
- C) When no net fluid is added or removed from a compliant chamber, in other words the outflow rate from the chamber equals the inflow rate then the effect of compliance vanishes to zero.

Now if we integrate (add up over time) equation 35 with time, to find the pressure in the chamber at any time t and ignore the integration constant (assuming the chamber has no initial pressure at zero time):

$$P(t) = \frac{1}{C_0} \cdot \int_0^t I \cdot dt \quad (EQUATION FOR COMPLIANT CHAMBER)$$

Equ.36

Equation 36 states the obvious, that if fluid is entering or leaving a compliant chamber over time you must integrate or add up the total flow over time to find the volume change and divide by the compliance to determine the new pressure in the chamber.

System analogies:

Compliance has a direct analogy with Capacitance in an electrical system. Capacitance is the ratio of amount of charge to the applied voltage (electrical pressure), rather than volume of fluid to applied pressure as in fluidics. Capacitors in electrical flow circuits are represented by two parallel plates, so this is a convenient symbol to use, for want of any better one, in fluidic flow diagrams to represent the compliance of the objects in our flow circuits.

In a "mechanical" system analogy of compliance "elasticity" is the property of springs. The stiffness of a spring is given as a spring constant **k**, which forms the proportionality constant between the applied force F and the displacement x, of the spring, given by Hooke's law: $F = \mathbf{k} \cdot \mathbf{x}$. The spring constant or "stiffness constant" **k**, is merely the inverse of the "compliance" of the spring, $1/\mathbf{k} = \text{spring compliance}$.

(3-7) COMPLIANCE OF FLUIDIC OBJECTS:

Compliance of tubing:

The compliance of tubing or any other elastic vessel for that matter is readily measured by filling the object under test with fluid (which for practical purposes can be regarded as incompressible, it is compressible but the error is small) and measuring the pressure change with added or removed volume. All of the compliances listed in table 1 have been measured from the disposable sets and machine vacuum sensors. It is possible to *calculate* a tube's compliance if you know the stiffness (or spring constant) of the type of plastic the tube is made from:

Equation 37 shows a formula for the compliance of fluid filled plastic tubing. This formula for compliance of plastic tubing Co, was derived by the author, as it was found that this formula is not readily available despite there being over 30 other parameters for plastic tubing that are published. The mathematical proof is attached in the data chapter.

$$Co = \frac{2.l.\pi r^2}{T.k}$$
Equ.37

Where l is the length of the tubing, r the internal radius, T the thickness of the tube wall and k the stiffness constant of the plastic from which the tube is constructed. The interesting thing to note is that for tubing of a total length l that Compliance is: Proportional to the square of the tubing internal radius r and proportional to the tubing length l linearly and inversely proportional to the thickness of the tubing's wall T and inversely proportional to the stiffness constant, or spring constant of the plastic. So stiffer plastic, thick walled tube, of the smallest possible internal radius would have the lower compliance. However in phaco fluidics the tube (for example aspiration tube) can't be too stiff or thick to handle easily and the peristaltic pump tubing itself needs to be able to be compressed by the pump rollers so it can't be too rigid.

The most interesting feature of the compliance of tubing though, is the fact that the compliance is spread along the length of the tube. The tube's inertia and resistance are also spread along the tubing length. This has unique effects on the fluidics of phaco which will be described in this book. Therefore with *plastic* tubing we can talk about compliance, resistance and inertia *per unit length* of tubing and we can call these CL, RL and LL values. Typical values of these are listed in table 2 for the aspiration tubing and irrigation system tubing.

Compliance of vacuum sensors:

The Vacuum Sensor in the phaco machine, communicating with the peristaltic pump tubing area, has to have some change in geometry for pressure or vacuum to be measured, or no signal could be generated. So this device adds to the compliance at the machine in the area of the peristaltic pump tubing near the machine and does influence the fluidic performance. On the other hand the vacuum level at the cassette in the venturi machine does not necessarily need to be monitored by a sensor if the properties of the venturi or electric vacuum device generating the vacuum are known. If a vacuum sensor is incorporated to monitor the cassettes' vacuum, it would confirm the venturi device or vacuum pump was producing the vacuum it was being instructed to. The compliance of the sensor in this instance is dwarfed by the large compliance of the air in the cassette which is approximately 3 orders of magnitude larger.

Compliance of air:

Unlike fluid, air has an obvious springy quality. Boyle's Law states that the product of pressure and volume for a gas is a constant at a constant temperature, or in other words P.V = K, where K is a constant. Therefore if we decrease the pressure around a bubble trapped in a fluid by $\frac{1}{2}$ its initial value, the volume of the bubble will expand to twice its initial volume value. This suggests that air has a *negative compliance*. If the pressure in the bubble decreases, then the volume of the bubble increases.

In the case of fluid filled tubing plastic tubing, or the eye, the opposite is true in that their internal volume decreases with a decrease in internal pressure. However when an air bubble is *trapped inside* a fluidics system it *increases the compliance*. This occurs because of the displacement of fluid volume by the bubble as the bubble expands when exposed to decreasing pressure (increasing vacuum) and shrinks when exposed to additional pressure. For example

consider a bubble which has a volume of 0.3 ml while sitting in the fluid of a fluid filled piece of plastic tubing sealed at each end. The pressure of the fluid is initially regarded as zero or atmospheric pressure. Then some fluid is aspirated out of the tubing with a syringe and needle until the vacuum in the lumen is 380mmHg (or 380mmHg vacuum). The bubble will then expand to twice its original size or 0.6ml. This reduces the volume for fluid to occupy by 0.3ml over that vacuum range and has a similar net effect to the tubing itself being compressed by atmospheric pressure and reducing the volume of the lumen of the tube which carries fluid. Therefore the amount of fluid which needs to be added back into the tube to get the internal tubing pressure back to zero and the bubble back to a volume of 0.3ml has increased by 0.3ml on account of the bubble being there.

The exact compliance of air depends on the pressure at which the compliance is being measured and on the volume of trapped air. For example if the initial bubble volume is larger there will be a larger volume displacement by the bubble for any vacuum level the bubble is exposed to. In the above example for a 0.3ml bubble we could say that the approximate increase in compliance due to the bubble was the amount its volume displacement 0.3ml, divided by the vacuum change of 380mmHg over which that displacement occurred. In mks units this is $0.3 \times 10^{-6} \text{ m}^3/50654 \text{ N/m}^2$ or $6 \times 10^{-12} \text{ m}^5/\text{N}$ (units equivalent to $\text{m}^6.\text{j}^{-1}$). This can be enough to decrease anterior chamber stability *depending on the location of the bubble*. Imagine this bubble near the probe end of the aspiration tubing, attempting to draw this 0.3 ml of fluid volume out of the eye via the phaco needle and probe, at occlusion break from 380mmHg vacuum and bear in mind the volume of the anterior chamber being only 0.2 to 0.3ml.

In general the compliance of a body of Air can be found by differentiating the Boyles Law to find the rate of change of volume with pressure, and substituting pressure and volume (P.V) in for the constant K. This yields simply the negative of the volume divided by the pressure, or -V/P.

If we consider a large volume of air in the cassette of the *venturi* machine eg 250ml, and assume a vacuum of 380mmHg (residual pressure of 380 mmHg) then the compliance , in MKS units, is very large at $4.9x10^{-9}$ m⁵/N, and this value dwarfs other compliances in a fluidic system by 3 orders of magnitude. Therefore the cassette air acts as a *vacuum reservoir* to help stabilise the vacuum value at the cassette. Small volumes of fluid entering the cassette have little effect on the vacuum there and this feature severely limits any utility for a "vacuum sensor" in a cassette to detect fluidic events such as occlusion make and occlusion break.

Having this large air compliance at the machine end of the aspiration tubing is not a problem in the *venturi* machine as the large compliance helps to damp the resonant properties of the aspiration system. The reasons for this will be explained later. On the other hand the compliance of the *vacuum sensor* in the *peristaltic* machine, along with the compliance of the peristaltic pump's tubing and the link tubing in the area will be shown to be very important in anterior chamber stability. The compliance of the peristaltic pump's tubing and vacuum sensor added together varies between peristaltic machine maker's disposables and their exact configuration. $4.5 \times 10^{-12} \text{ m}^6 \text{ j}^{-1}$ is a typical value.

Compliance of the Eye versus Compliance of the Machine's Vacuum Sensor and Disposables:

The eye's compliance is a significant controller of anterior chamber stability. The value, for a pig eye (converted to mks units) is in the order of 30×10^{-12} m⁶ J⁻¹. The compliance is a fairly stable value over the range of 5 to 50mmHg. Below 5mmHg the eye begins to "collapse" and the compliance increases significantly, Zacharias⁷. It will be shown how and why larger values of eye compliance improve anterior chamber stability and how large values of machine compliance, in the aspiration tubing and at the pump and vacuum sensor in the peristaltic machine, degrade anterior chamber stability.

It should be noted that local fluid currents within the eye do not alter the anterior chamber pressure or basic geometry, only fluid entering or leaving the eye will do that. The study of localised fluid currents within the eye would be the subject of another book.

Adding compliant structures to flow circuits:

Any compliant structure (excluding trapped air bubbles and cassette air) in the fluidics system is exposed on one surface to atmospheric pressure (zero reference value). Therefore in all our fluidic flow circuits the compliant chamber is represented by two plates (analogous to the symbol for an electrical capacitor) again joined into the flow circuits with lines that again represent lossless flow connections. One "plate" therefore will always be returned or be "connected" to the zero reference pressure point. It is also a simple matter to return one plate of the eye's compliance symbol (representing the external surface of the eye) to a positive pressure source above the zero reference pressure point. This is equivalent to adding positive orbital tissue pressure, or speculum pressure, on the outer wall of the eye for example.

(3-8) DYNAMIC IRRIGATION SYSTEM PATHWAY MODEL.

We now can analyse the flow situation of fig 8:



Fig 8

Figure 8 shows the irrigation system now feeding the eye, with no fluid leakage. The eye in this example starts at zero pressure and the irrigation valve opens (or S1 of figure 9 closes to allow flow) to start the flow. After some time the flow stops because the eye pressure will climb to the bottle pressure Pb.



Figure 9 now has an object with two plates representing the eye's compliance Ce which is $30x10^{-12}$. One plate is returned to the zero pressure reference point, the other is inside the "eye" to symbolise the compliance of the eye. It is obvious too for this flow circuit that the flow cannot keep running continuously because the flow, for a constant flow rate is "obstructed" by the compliant chamber much as a continuous electrical current flow is blocked by an electrical capacitor. The flow is a function of time and the compliant chamber "Charges with fluid volume" and the pressure within it after a time climbs to the bottle height. Along the way the pressure developing in the eye opposes the bottle pressure driving the irrigation inflow.

Using the principle that all of the pressures around the circuit add up to zero (Kirchoff's second law) and adding our equation for compliance 36 to our previous equation 31 we now have the equation to describe the flow circuit where the irrigation instrument is place within the eye:

$$Pb = \frac{L}{dt} + R.I. + \frac{1}{Co} \cdot \int_0^t I. dt$$
Eau.38

Now this equation again is not in convenient form, but it can be solved to give the answer to both pressure in the eye and flow in the irrigation pathway with time after the flow begins from a time t = 0 and until later when the flow stops because the eye (compliant chamber) has reached the bottle's pressure Pb.

It turns out that there are three possible forms of solution for equation 38. This is because there are three possible states of damping. Inspection of this fluidic system immediately shows that the system can be oscillatory. The fluid is given kinetic energy, it enters and inflates the compliant chamber (eye) and the kinetic energy is exchanged for elastic energy as the wall of the eye is stretched and this has the potential to expand and slightly over-expand the chamber. Fluid then can oscillate out of the chamber back into the tubing. This is very much analogous to a mass with kinetic energy oscillating on a spring with elastic energy and with resistive losses damping out the oscillations. Therefore the solution is either one of three formats:

Over damped, critically damped, or oscillatory fluid motion decaying away with time.

For the motion to be oscillatory; $Kreyszig^6$: the resistance must be less than twice the square root of the ratio of L/C.

In a mechanical system with a mass m on a spring, with a compliance C and a damping resistance R for the system to be oscillatory, $Kreyszig^{6:}R$ must be less than twice the square root of the ratio m/C.

Taking the values from table 2 for the typical irrigation system, with the yellow sleeve/needle annulus *in the eye*, $Ri = 7.36 \times 10^{9}$ (30ml/min value) $Li = 5.66 \times 10^{8}$ and $Ce = 30 \times 10^{-12}$. The behaviour is oscillatory, but it is getting close to critical damping so the oscillations are quickly damped out after the first half cycle.

The general equation solution for this under-damped case for Pe(t), and flow I (t), with time are, modified from Page Adams⁸:

$$Pe(t) = Pb\left[1 - e^{-\alpha . t} \sqrt{\frac{\omega^{2} + \alpha^{2}}{\omega}} . \sin(\omega t + \delta)\right]$$

$$Equ.39$$

$$\delta = tan^{-1}\left(\frac{\omega}{\alpha}\right)$$

$$\omega = \sqrt{\frac{1}{Li.Ce} - \frac{Ri^{2}}{4.Li^{2}}}$$

Equ.40

Where α is the damping constant which determines how quickly the oscillations decay away and ω is the angular frequency of the oscillations. δ is the phase angle between the pressure and flow.

As the reader can see the equations are becoming more involved. They don't make for easy reading and are requiring more symbols to explain them. Now the circuit simulator comes to rescue to plot a graphical solution so the reader can "see" what is happening with the pressures in the eye and flow with time. It can be readily seen from the equations however what is happening. The bracketed part of equation 39 is merely a function of time varying function relating to the frequency of the sinusoidal oscillation. The exponential part and its negative exponent, relates to the decay of heavily damped oscillations.

Figure 10 shows the result of running the flow circuit model of figure 8 with the graphical output, which is merely a plot of the equations 39 and 40 above with the Ri and Li total values of the irrigation system including the yellow sleeve needle annulus, and the compliance of the eye. The bottle height is 70cm or 51mmHg or 6798 N/m² in this example. The eye pressure is in blue and the inflow rate in brown.



At a time t = 0 the flow is switched on by S1 analogous to switching on the irrigation flow, to an eye with a zero initial pressure. The irrigation flow rate graph (brown) is super imposed with its peak value labelled and to see the time course on the X axis. The Y axis is the eye pressure Pe, values in K "Kilo" or 1000 Newtons per square meter. The flow rate now builds up and turns around again to approach a value of zero. The flow peaks around 135 milli-seconds after the irrigation pinch valve opens, to a value of 641nano cubic meters per second or 38.5ml/min.

In this instance we cannot calculate easily from the formula a "time constant" because the system is more complex. It can be seen from equation 40 that the peak rate of flow is proportional to the driving pressure, in this case Pb. This would be exactly true if the flow resistances that comprised Ri were a constant at any flow rate. The peak flow is less than proportional to the driving pressure due to the square law resistances which comprise part of Ri, so Ri increases with flow. The flow rate under static flow conditions or under dynamic conditions becomes proportional to the square root of the bottle pressure. In the case of the dynamic flow we can now inspect the approximate peak flow rates which occur.

A method therefore to allow for the changes in resistances with flow and to help increase the accuracy of the model is to select the flow resistance value which occurs at close to the *peak flow rates* of interest. This was done in the previous static flow circuit models where the resistances used were those at the static flow rates of interest.

One way to approach the assessment of the performance of the irrigation apparatus is simply to measure from the graph the time it takes to get the eye, from some initial lower pressure to near the bottle pressure. The dot on the graph shows this 90% point. The amount of time read from the graph of figure 10 is 425 milli- seconds.

The rate of change of eye pressure in the eye is therefore limited to a certain rate due to the combination of *resistance* and *inertia* in the irrigation pathway and the *compliance* of the eye. Recall from the ABC's of compliance that under dynamic conditions there will *delay the rise in pressure* in the compliant chamber (eye) with respect to the flow timing. This is evident in figure 10. In addition there is a delay in the flow starting after the pressure is applied at t = 0, and this is due to the fluidic inertia. These *combined delays* result in the lag in rise in eye pressure after the irrigation pressure is applied *or at any time the irrigation is attempting to raise the eye pressure from a lower value to a higher one (or the reverse)*.

In general with normal bottle height up to a meter, the peak inflow rate would rarely exceed 50 ml/min, and elevating the bottle will not produce proportional increases in the peak flow rates due to the square law resistances in the irrigation pathway. This tells us something about the flow rates at which we should be testing our irrigating instruments. We can now look at the mathematical model and circuit simulator result of figure 10, and compare this to a practical measurement and recording:



Figure 11 is the experimental measurement. In this experiment a phaco machine was set up with the standard disposables outlined in table 2 and the Yellow needle/sleeve fitted to the

probe. The probe needle and sleeve (without any "wound" compression) was placed in a test chamber with the same compliance as the eye $30x10^{-12}$ and also containing a low compliance piezo-resistive pressure sensor so as to measure the chamber pressure. The output of the sensor is recorded (graphed) on a Tek 464 storage oscilloscope, then photographed with a digital camera from the oscilloscope tube face.

The only function being used on the phaco machine in this experiment is the irrigation pinch valve and the pole to support the bottle. The irrigation pinch valve opens physically in time as indicated on the diagram. This generates a small negative pressure transient (artefact) in the chamber about 50 milliseconds later and after a time the flow increases into the compliant chamber (eye) and the pressure increases. The general pattern of pressure rise with time follows a nearly identical time course to that predicted by the flow diagram of figure 8 and the equation 39 and the plot of figure 10 complete with the damped oscillatory overshoot. However there are some very interesting and subtle differences which one could overlook if not careful:

The negative pressure artefact takes a while, in the order of 50 milliseconds to appear in the test chamber (eye). This is the time taken for this pressure impulse to propagate down the 2 meter tubing length from the pinch valve toward the chamber (eye). The fluid in the tubing leading to the pinch valve from the bottle is at the bottle pressure prior the valve opening. When the valve opens this fluid enters the tubing leading to the probe. The fluid in that part of the tubing and the eye at this moment start at a zero pressure. As the fluid enters the tubing absorbs the fluid and there is a delay before any fluid *emerges* at the probe and annulus and hole area at the opposite end of the tubing in the "eye". In addition to this there is a peculiar "kink" or *change in shape* of the pressure waveform as indicated on the diagram of figure 11. Also the initial rise in pressure in the chamber looks very *linear* and not slightly curved as the model in figure 10 suggested it would or should be.

To understand how the features seen in the actual recording come about we need to refer to figure 12:



Figure 12 is a new flow circuit diagram. This flow diagram now incorporates the compliance of the tubing leading to the pinch valve and the compliance of the tubing leading from the pinch

valve to the probe. The tubing model is now broken into a large number of sections each with a small amount of resistance, inertia and compliance to model these features which are distributed along the length of *real plastic tubing*.

The diagram shows the tubing broken into small sections of length x, each with a compliance of Cx, resistance of Rx and inertia of Lx. Only three of these are shown for clarity but one could imagine there being hundreds of small sections each with a small characteristic Lx, Cx and Rx. The totals of these simply add to the total values for the length of tubing under consideration. Lf and Rf is the inertia and resistance of the fittings near the bottle and drip chamber. The fluidic inertia of the irrigation sleeve and probe body are represented by the values Rs, Ls, Rp and Lp. Overall the totals of Ri and Li are just the same as those given in figure 9 however in this case the distributed compliance of the plastic irrigation tube is allowed for. Switch 1 in figure 12 represents the irrigation valve which closes at t just greater than zero.



The graphical result run on the circuit simulator and plotted in figure 13:

Now the practical experimental result of figure 11 is very closely replicated by the mathematical model, including the initial near linear rise in pressure, the kink in the waveform, and the delay from when the pinch valve opens at t just greater than zero to the pressure starting to increase in the eye. The kink itself is caused by pressure waves which travel in the irrigation tubing after a sudden change in applied pressure across the tubing and the linear initial pressure rise due to the initial constant flow situation induced by the tubing. More subtle kinks appear further along the waveform. These features will be studied in this book in detail in the section on *transmission line delay phenomena of compliant tubing below*.

While knowledge of the irrigation flow timing is useful after a pinch valve opens, it is not nearly as important as knowing what happens to the eye pressure when the flow is occluded during surgery at the phaco needle. The reason for this is that this behaviour is one of the

determinants of lens fragment holding forces during surgery. Again the dynamics are different to when the pinch valve opens:

Figure 14 shows the equivalent flow circuit in the situation where the pinch valve is open and the flow (30ml/min) is occluded at the phaco needle, depicted by S2 opening, and interrupting a flow rate of 30 ml/min.



Figure 15 is the graphical solution of figure 14. The bottle pressure is 51 mmHg and prior to occlusion the flow rate is 30 ml/min and the eye pressure is 23 mmHg during this stable flow situation. After occlusion at t = 1 second (green vertical line indicator on graph) the eye pressure starts to climb toward the bottle pressure with the typical overshoot of damped simple harmonic motion. This takes 260 milliseconds. The time is shorter than that indicated in figures

10, 11 and 13 because it is a smaller change of pressure to get from the anterior chamber pressure of 23 mmHg, to the bottle pressure of 51 mmHg. However the basic shape of the pressure increase profile, an inverted exponential format, is the same as seen in the basic model and mathematics describing figure 10. The inflow flow rate, shown (brown) is initially a stable 30 ml/min, equal to the aspiration flow, and falls away to zero after occlusion as the eye pressure approached the bottle pressure.

If the flow rate is higher prior to occlusion, the anterior chamber pressure will be lower in the un-occluded state for the same bottle height. Figure 16 shows the situation when the flow rate is 50 ml/min, and the anterior chamber pressure starts from a lower pressure of 3mmHg makes a longer climb toward the bottle pressure when the flow is occluded and this takes in the order of 300 milliseconds.



If the anterior chamber pressure is suddenly lowered by a very rapid fluid out-flow event, to some value below the bottle pressure, the irrigation system will take a certain time to return the eye to a new equilibrium pressure. This delay is shorter, the lower the flow resistance value and fluidic inertia of the irrigation system. The flow resistance total value in the irrigation system is dominated by the irrigating annulus or chopper.

In general the effects of the distributed compliance of the *irrigation* tubing does not have an enormous effect of the irrigation pathway dynamics and the simple flow resistance and inertia and eye compliance model of figure 9 and graph of figure 10 give a good approximation of the irrigation performance. However the subtle details in the experimental results of figure 11 are not explainable without incorporating the transmission line model to allow for the distributed feature of the compliance, resistance and fluid inertia along the irrigation tubing. The more subtle details found on practical experiment can be explained this way when the compliance of the irrigation tubing is factored into the model as seen in figure 13.

In the case of the *aspiration* system, where the pressure gradients and peak flow rates are much larger, the behaviour cannot be adequately or accurately explained at all without the incorporation of the aspiration tubing's distributed compliance and taking into account the transmission line behaviour of the aspiration tubing. So in summary we need this more complex tubing model for the aspiration system, but not the irrigation system.

Figure 17 shows the result using the simpler model of the irrigation system as described by equations 39 and 40 and the flow circuit of figure 9, which is a perfectly satisfactory model from a practical perspective. Again and using the Ri and Li values one has with chopper 2 and its improved handle. This can be compared with figure 10 for the yellow needle/sleeve. Due to the decrease of flow resistance the rate of eye pressure increase with time is more rapid for this chopper compared with the yellow/needle sleeve annulus. The chopper and its improved handle, provides the irrigation system with an overall Ri of 5.18×10^9 and an Li of 5.37×10^8 . The in-flow rate peaks to 47.6ml/min at around 140 milliseconds and it takes 300 milli-seconds for the yellow sleeve annulus arrangement. The *dynamic* performance of this chopper is superior to any of the other irrigating devices listed in table 2. Due to the lower resistance there is less damping with the improved chopper versus the annulus arrangement and the pressure overshoot seen in figure 17 is more obvious:



The irrigating instruments with the lower flow resistances therefore:

- 1) Maintain the anterior chamber at a higher pressure during stable flow situations.
- 2) Have a better response time to correct transient eye pressure reductions caused by rapid out-flow events, for example a post occlusion surge in a peristaltic machine.
- 3) Provide a greater rate of increase of anterior chamber pressure over time, from any given eye pressure, towards the bottle pressure. This sort of event occurs when a constant flow situation is occluded at the phaco needle.

(3-9) SUMMARY OF THE IRRIGATION SYSTEM PATHWAY:

Mathematical modelling and experiment shows that it takes around 300 to 400 milliseconds to bring the eye pressure from zero to the bottle pressure when the irrigation is initiated. Any rapid events or transient fluid outflows from the eye which significantly lower the anterior chamber pressure over brief periods, for example less than 250 milliseconds are not well compensated for by the irrigation pathway for typical phaco fluidics systems as it stands today.

The Eye's compliance hampers the irrigation, as the rise in eye pressure is delayed with respect to the inflow. On the other hand the eye's compliance has a stabilising effect on the eye pressure with fluid outflow or aspiration of fluid from the eye, as the fall in eye pressure is delayed with respect to the outflow. So eye compliance is a very interesting dual edged sword in those respects.

If the flow *resistance* in the irrigation pathway is increased, this then increases the lag time in the irrigation pathway. The bulk of the flow *resistance* in the irrigation pathway is in the irrigating instrument itself. On the other hand the bulk of the *inertia* in the irrigation pathway is in the total length of the irrigation tubing.

The fluidic inertia in the irrigation pathway can be decreased by using wider bore tubing (because the inertia is inversely proportional to the square of the tube's radius) however there are practical limits as to how bulky the tubing leading to the probe can be. Increasing the radius by a factor of 2 will reduce the inertial value of the fluid filled tubing at least to ¼ of what it was. In general, the industry standard irrigation tubing is 1/8" or 3.175mm internal diameter. AMO however has increased the bore on the irrigation tubing to 3/16" (or 4.75mm) over the 1.3 meter length leading from the bottle fitting to the pinch valve at the phaco machine. This helps a to reduce the overall irrigation *resistance* and *inertia* and is a move in the right direction.

With larger internal diameter irrigation tubing there is the possibility of more air bubble trapping as fluid can track along the inner tubing wall, bypassing trapped bubbles and not priming them out. These bubbles can turn up in the eye later during surgery.

(3-10) ASPIRATION SYSTEM PATHWAY DYNAMICS – Vacuum Machines.

We will now examine the aspiration system/pathway and ultimately fuse this with an overall solution with what we have just learnt about the irrigation pathway's limitations.

The aspiration pathway's behaviour and performance is completely different comparing the venturi and peristaltic machine. Comparing the two is like comparing apples and pears. The behaviour of the two is completely different at occlusion break and also different when an occlusion forms in terms of lens fragment holding forces.

Initially we will model the aspiration pathway as though the aspiration tubing were rigid, and the compliance of it (symbol Ca) is zero. Any system compliance will initially exist in the model only in the eye denoted as Ce and at the phaco machine's end of the tubing denoted Cm. This is the machine's compliance. This is the compliance of the peristaltic pump tubing leading to the first occluding roller and the compliance of the machine's vacuum sensor and tubing in that local area near the pump. In the case of the venturi/vacuum machine, the implications of the very high compliance of the air in the cassette of the venturi machine will also be explained.

As the model complexity increases to incorporate the distributed compliance of the aspiration tubing, it will be explained how the tubing is involved in a number of unique phenomenon at occlusion onset and occlusion break.



Aspiration dynamics Vacuum/Venturi Machines – Occlusion break:

Again we generate a flow circuit of figure 18 for the events which occur immediately after an occlusion break in the *venturi* machine. First we consider an eye with *no irrigation* but which has an initial anterior chamber pressure of 51mmHg. Prior to the occlusion break the switch or tap S2 is turned off. This represents a blocked phaco needle so the needle lumen and the vacuum there is not exposed to the anterior chamber. At a time t = 0 the switch S2 closes (occlusion breaks free) and fluid is pushed out of the eye toward the cassette. The vacuum in

the cassette is a constant 100 mmHg (surgeon's pedal in the same position) and the small amount of fluid entering the air filled cassette does not change the vacuum there to any significant degree as the cassette volume does not change significantly for a small inflow. The air in the cassette stabilises the cassette vacuum.

Using again the total with the AMO yellow needle we can now inspect the rates at which fluid is exiting the eye and note the effect on the eye's pressure over time. The total inertia of the aspiration system fluid is $La = 1.15x10^9$ and the total aspiration resistance $Ra = 2.37x10^{10}$ (45ml/min figure to select the value closest to the peak flow rates) and Ce, the eye's compliance is $30x10^{-12}$. Again we have a potentially resonant system, comprising Ce, La and Ra.

Inspection of the values shows that the system is *heavily damped*, over critical damping:

$$R > 2\sqrt{\frac{La}{Ce}}$$
 and in this case is *not oscillatory*, unlike the irrigation pathway

The equation for the this overdamped case for a discharging capacitor or compliant chamber with time Pe(t) is, modified from Page Adams^{8:}

$$Pe(t) = Pb - (Pb + Pv) \left[1 - \frac{1}{2\beta} e^{-\alpha t} \left[(\alpha + \beta) e^{\beta t} - (\alpha - \beta) e^{-\beta t} \right] \right]$$

Equ.41

$$\alpha = \frac{Ra}{2.La}$$
$$\beta = \sqrt{\frac{Ra^2}{4.La^2} - \frac{1}{La.Ce}}$$

The equation is formatted for the pressure change total (Pb + Pv) over which the event occurs and the fact that after a time the pressure approaches the vacuum value Pv. (The cited equation⁸ simply started at a maximum pressure in the compliant chamber(capacitor) and decayed to zero over time). With time then the value approaches a vacuum of -Pv, in this case a vacuum of 100mmHg below zero or our atmospheric pressure reference. Figure 19 shows the circuit simulator graphical result of equation 41 because, as the reader can see, it would be a painful proposition to plot it manually:



It can be seen that without any irrigation, the eye pressure approaches zero pressure (small red square on the blue/green line) at 320 milliseconds after occlusion break. This is with an initial eye pressure of 51 mmHg 6798 N/m² or 6.789K N/m² on the graph's y axis and a machine vacuum of -1330K N/m² or 100mmHg. Also by 2.4 seconds the eye has reached the machine's cassette vacuum level of 100mmHg and would be thoroughly collapsed. The outflow flow from the eye (magenta line) peaks around 146 milli-seconds and the peak outflow rate, superimposed on the graph is 44ml/min.

As can be seen from the equations and the nature of the graph of figure 19 that the solution due to the "over damping" contains no oscillatory or periodic terms such as sin or cos, only exponential functions with time. With enough time due to the $-\alpha$.t exponent, the value of the large bracketed portion of the equation approaches 1, so that the eye pressure Pe simply approaches the value of the vacuum in the cassette Pv or 100mmHg.

We are now in a position to combine the models created for the irrigation system and aspiration system into one:

(3-11) COMBINED IRRIGATION AND ASPIRATION SYSTEM- Vacuum Machine:

Combining the irrigation and aspiration system models generates the flow circuit depicted in figure 20:



We assume the eye is at the bottle pressure initially, as it is at over 400milli-seconds after the flow is occluded. Again when the occlusion breaks at t = 0, S2 closes and the flow begins. The bottle pressure is 51mmHg, and so is the eye pressure prior to occlusion break (ignoring any leakage over the time intervals under consideration) and the cassette vacuum again is100mmHg.

The solution to the equation which describes this overall "flow network" With Pb, Ri, Li, Ce, Ra, La and the cassette vacuum Pv is very large and daunting. This is because there are now two damped resonant systems with different damping properties coupled together. However we can relax because the circuit simulator software comes to rescue and the result is represented graphically see figure 21.



The graph of figure 21 now contains the inflow rate (brown), the outflow rate (magenta) and eye pressure (blue/green) after occlusion break. This shows the relative timing. The Y axis is the eye pressure value for the blue/green graph. The inflow and outflow graphs (brown and magenta) are superimposed to show the relative timing on the x axis.

The Ri value is $7.36x10^9$ which corresponds to the yellow sleeve/needle in the eye and the Ra = $2.37x10^{10}$ (45ml/min value), Li = $5.66x10^8$.La = $1.15x10^9$ Ce is $30x10^{-12}$. Pb = 51mmHg, Pv = 100mmHg.

We can see now how the inflow *lags* the outflow in the time after occlusion break. After a while the inflow and outflow are equal (ignoring wound leakage not added to this flow circuit).

As can be seen for the *venturi/vacuum* machine complete flow network, the eye pressure drops, after the occlusion breaks free, and returns to the equilibrium pressure when the flow rate stabilises after about 400 to 500 milliseconds. The slight eye pressure *undershoot* is due to the *resonance* (mild oscillation) in the *Irrigation system* as described above comprising the inertia of the irrigation fluid and the compliance of the eye. This pattern of eye pressure after occlusion break in the *venturi* machine has been observed and documented by other investigators, Wilbrandt⁹. The description here is now the formal explanation for the occlusion break eye pressure profile in the venturi machine and how this occurs. The mild gradual *undershoot of eye pressure* to 14.2 mmHg then recovery to 15.2mmHg is caused by simple harmonic motion (under-damping) in the *Irrigation system* and is unrelated to the venturi machine itself or the aspiration system properties which are well over critically damped. If we repeat this example using the same phaco needle but the improved chopper 2 and handle, with a lower irrigation resistance, making Ri 5.18x10⁹ and Li also little lower at 5.37x10⁸ then the result is shown in figure 22.



Due to less damping in the irrigation system the eye pressure "undershoot" is more obvious. In this case, also due to the lower flow resistance in the irrigation device, the *stable flow equilibrium* eye pressure is higher than that seen in figure 21 at 23.9mmHg vs 15.2mmHg.

This is also the predicted pressure by the simple eye pressure equation 19 for the venturi machine when the same values of Ra, Ri and Rt are used with a vacuum of 100mmHg and a bottle height of 70cm (51 mmHg).

It can be seen that stable anterior chamber pressure and anterior chamber stability does not occur until about 600 to 700 milliseconds after occlusion break. Even though the value of eye pressure undershoot, with the improved or lower resistance irrigation device, is larger than with a higher irrigation resistance device, the absolute pressure drop, which is the important feature, is lower to only to 19.8 mmHg versus 14.2 mmHg with the two irrigation devices respectively.

A point of note here is that typically the venturi machine has no significant "Post Occlusion Surge" per se. After occlusion break there is as *relatively slow return of eye pressure* to a new stable flow equilibrium as can be seen from fig 21 and 22. This is because there is a *heavy state of damping in the venturi aspiration system*.

Anterior chamber instability in the venturi machine stems from the fact that the un-occluded flow rates are often on the high side. Therefore the anterior chamber pressures are low during un-occluded flow. At occlusion the eye pressure then climbs to the bottle pressure say 51mmHg and after occlusion break returns to a relatively low value or 15.2 mmHg with a flow rate of 38.8 ml/min with the disposables system outlined above and 100 mmHg vacuum. Wound leakage takes the eye pressure value lower. Therefore anterior chamber instability seen in the venturi machine is better understood as high flow rate related occlusion instability or *flow occlusion instability*, rather than *post occlusion surge*. In general the flow rates in venturi machines are often on the high side and not known to the surgeon, even at modest vacuums. In contrast the peristaltic machine, which aside from a post occlusion surge, which is unrelated to the machines peristaltic pump motion, has very stable un-occluded flow on account of the peristaltic pump regulating the flow rate.

It may at first thought seem to be a simple matter of adding flow resistance to the aspiration system on the venturi machine, with flow restrictive devices such as longer lengths of aspiration tubing or aspiration tubing with a smaller internal diameter to allow the use of high vacuums while still having sensible flow rates. The compromise is that lower vacuum levels, if they are required during the cataract surgery, then result in very low flow rates unless the flow restrictive device is removed. This limits the surgeon control over the vacuum to a fixed high level. Low flow rates reduce the clearance of fine anterior chamber lens debris (smoke) and encouraging reduced probe and needle cooling and encouraging wound burn. More will be said about this in the section on advantages and disadvantages of peristaltic and venturi or electric vacuum pump phaco machines.

The venturi machine's aspiration system compliance and the air at the cassette is not a major issue. To establish a vacuum in the cassette, air must be removed from the cassette. The venturi device extracts air from the cassette, reducing the pressure as there are then fewer air molecules in the same volume of the rigid cassette. The air remaining within is equivalent to expanded air with a lower pressure. Then either fluid or air can be driven via the tubing toward the cassette. This will increase the pressure by decreasing the volume of the cassette in the case of fluid entry, or increasing the pressure (decreasing the vacuum) by adding more air. As the cassette's

volume is generally large, 100ml or more, a small volume of introduced fluid, at times after occlusion break, will not alter the pressure (vacuum value) greatly. The vacuum in the largely air filled cassette is dependent on the position of the surgeons pedal and the maximum vacuum settings on the machine. Because a large volume of fluid would be required to neutralise the vacuum (expanded air) in the cassette and compress the remaining air to a volume whereby atmospheric pressure was re-established, then the cassette can be thought of as a *vacuum reservoir chamber*. Small amounts of fluid (a few millilitres) entering the chamber will not significantly alter the vacuum there. (This is completely unlike the peristaltic machine where the stored vacuum is quickly neutralised by the post occlusion surge flow which can be detected at the vacuum sensor). A vacuum sensor, if present in the vacuum/venturi machine's cassette is therefore unable to recognise or detect the onset of occlusion break, as the cassette vacuum remains stable with the introduction of a small amount of fluid into the cassette.

The compliance of the aspiration tubing leading to the probe in the venturi machine does become an issue if high vacuum (small bore) aspiration tubing or flow restrictive devices are used in the aspiration system which then allow high vacuum levels eg 200 or 250mmHg or greater to be used. The properties and influence of aspiration tubing will be discussed shortly.

(3-12) ASPIRATION SYSTEM PATHWAY DYNAMICS - PERISTALTIC MACHINES:

Equivalent flow circuits are again used to describe the dynamics within the *peristaltic* aspiration system. The flow circuit is depicted in figure 23:



Figure 23 depicts two on off switches or taps S2 and S3. Ce is the compliance of the eye 30×10^{-12} and Cm, the compliance of the machine's peristaltic pump tubing and vacuum sensor area 4.5×10^{-12} . In this case because the peak aspiration flow rates are very high (around 100ml/min) the 90ml/ min Ra value of 3.08×10^{-10} is used to give the most accurate modelling result possible. Ri is 7.36×10^{9} , Li = 5.66×10^{8} and La = 1.15×10^{9} . In this case Pv is the maximum

vacuum prior to occlusion break stored in Cm, and we set this at 500 mmHg (66650 N/m²). In this instance again we ignore the distributed compliance of the aspiration tubing for now and imagine the aspiration tubing is rigid, having only fluidic *inertia* and flow *resistance* but no compliance.

When occlusion occurs we can regard S2 as open and the anterior chamber not in communication with the aspiration system. The eye also climbs to the bottle pressure of 51mmHg. Again we disable or disconnect the irrigation system to first observe and describe the behaviour of the aspiration system. After occlusion, the peristaltic pump is still turning and removing fluid from the aspiration line. The result is that the vacuum in the aspiration system builds up over the rise time. This vacuum is being monitored by the machine's vacuum sensor, and climbs over time, toward the value programmed by the surgeon, in this example 500mmHg. When this state is detected the pump is instructed to stop. In this case we now regard switch S3 as open. A pump roller simply leaves a blind end in part of the pump's tubing which contributes to part of the value of the machine's compliance Cm.

Cm is now *charged* to the maximum vacuum value Pv = 500mmHg and the system is set up for a post occlusion surge. At occlusion break at t = 0 (S2 closes), fluid is rapidly extracted from the eye and enters the aspiration tubing from the eye, and the stored vacuum in the aspiration system decreases below the set maximum, and therefore the pump is instructed to start. If the pump could start with a negligible delay, then in fact the post occlusion surge is amplified somewhat as will be shown. In practice the pump has a significant electromechanical delay of 100 to 200 milliseconds.

The vacuum, at the vacuum sensor falls to a lower vacuum (a more positive pressure value) and the pump is instructed to start, albeit with the inherent delay. In this initial instance we will leave the pump off to observe the effects. As the fluid enters the aspiration system the compliant structures at the machine expand back to their previously uncompressed geometry.

Looking at the equivalent flow circuit of figure 23, we now have an interesting situation where the circuit contains two compliances, Ce and Cm, which are in series in a circuit with both La and Ra. Regardless of the zero pressure reference point, being located between these two compliances, this *resonant system* behaves as though there is one compliance, equal to Ce and Cm in series. The formula for adding capacitors (or compliances) in series ,Horowitz ⁵ to find the total compliance Ct, is,

$$Ct = \frac{Ce.Cm}{Ce + Cm} .$$

Inspection of the values of Ct, La and Ra show the system is *oscillatory* (under damped)

Equ.42
$$Ra < 2\sqrt{\frac{La}{Ct}}$$
. So we expected oscillatory behaviour.

Therefore the effect of the machines compliance acting in series with the eye's compliance reduces the overall aspiration system compliance Ct such that a state of under damping occurs.

The equation solutions presented here are formatted to give the pressure across Ce, which is the eye pressure at any time or pressure in the eye as a function of time Pe(t). The solutions are presented describing this event for the pressure in the eye with time Pe(t) and the outflow flow of fluid from the eye with time I(t), modified from Page Adams⁸ are:

$$Pe(t) = Pb - (Pb+Pv) \cdot \frac{Cm}{(Ce + Cm)} \left[1 - e^{-\alpha \cdot t} \cdot \sqrt{\omega^2 + \alpha^2} \cdot Sin(\omega t + \delta) \right]$$

$$Equ.43$$

I (t) = (Pb + Pv) Ct.
$$e^{-\alpha t} .(\frac{\omega^2 + \alpha^2}{\omega})$$
.Sin(ωt)

Equ.44

$$\alpha = \frac{\text{Ra}}{2\text{La}}$$

$$\delta = \tan^{-1}\left(\frac{\omega}{\alpha}\right)$$

$$\omega = \sqrt{\frac{1}{\text{La.Ct}} - \frac{\text{Ra}^2}{4\text{La}^2}}$$

$$\text{Ct} = \frac{\text{Cm.Ce}}{\text{Cm+Ce}}$$

Equation 44 is very similar to equation 41 for the venturi machine, except that the result is periodic or oscillatory and the eye pressure fluctuation due to the occlusion break, and post occlusion surge which follows is scaled by the ratios of the compliance of the machine Cm and the sum of the machine and eye compliance Cm and Ce. It is within these solutions that the mysteries of post occlusion surge are explained:

Firstly, looking at the equation for eye pressure Pe(t) we see that the function in the brackets is merely a function of time varying in an oscillatory manner. We also see that the dynamic magnitude of the surge proportional to:

 $(Pb + Pv) \ge Cm / (Cm + Ce)$. This is a very interesting result:

The magnitude of the surge is in fact proportional not to the maximum vacuum alone, but the sum of the vacuum and the bottle pressure. As most surges of any practical significance occur with a maximum vacuum of over 200 or 250mmHg, it has generally been the Surgeon's experience that the surge is "proportional to the maximum vacuum". This is because, as a numerical value the bottle pressure is the smaller value and is swamped by the bigger figure.

The surge flow rate I(t) and its peak value depend on the total compliance Ct and the sum of the bottle and vacuum pressure values.

Also we can see from equation 43 that the magnitude of the surge is proportional to Cm/(Cm+Ce). This means that all other things being equal, phaco machines with higher compliance disposables and vacuum sensors (eg large Cm) will generate bigger surges for the same maximum vacuum setting, and this has been widely known. However this equation shows and explains the critical importance of the eye's compliance, as the eye's compliance value is typically 6 to 7 times larger than Cm. *Eyes with a higher compliance will have a lower post occlusion surge*, all other things being equal. This explains information presented by other experimenters, Zacharias⁷ that the shape or profile of the measured post occlusion surge depends on the compliance of the chamber in which the surge is measured.

Plotting out equations 43 and 44 would be a tedious proposition. Again the circuit simulator and graphical solution is deployed. This is a handy shortcut to "see" the result, however is should be remembered that the answer to key questions in fact lies in the equation solutions above.



Figure 24 shows the fluid outflow and eye pressure Pe(t) plot of equation 43 after occlusion break at t = zero when the occlusion breaks free. The outflow (magenta line) is superimposed. As can be seen there is rapid reduction in eye pressure over a brief period, of about 150 milliseconds which takes the eye pressure to zero. With no irrigation the net end pressure point is $-2796 \text{ N/m}^2 \text{ or } -21 \text{ mmHg}.$

Note that there is a rapid fluid flow extraction from the anterior chamber which peaks at 70 milliseconds after occlusion break with a peak outflow rate of $1.70u (1.70x10^{-6} \text{ m}^3/\text{s})$ or 102ml/min. Referring to the graph figure 10 for irrigation flow, it can be seen that the irrigation system would be expected to be too "slow" to deal with this rapid transient fluid extraction from the eye, and indeed it is too slow which is why the eye pressure dips just after the peak surge outflow.

We are now in a position to combine the peristaltic irrigation, and aspiration systems, and observe the dynamic behaviour. As will be seen it is fairly closely explained by the combination of figure 10 for the *irrigation dynamic properties* and figure 24 of the *aspiration dynamic properties* if they were merely overlaid upon each other.

(3-13) COMBINED IRRIGATION AND ASPIRATION SYSTEMS – PERISTALTIC MACHINES.



We again refer to an equivalent flow circuit figure 25:

Figure 25 depicts the combined fluidics flow circuit for the peristaltic machine. When S3 is closed the pump generates a flow Ip which is set to 30ml/min in this example. When S3 is open this is equivalent to the pump being stopped. When S2 is open this is equivalent to an occlusion of flow at the phaco needle tip. The occlusion suddenly breaking free is represented by S2 closing and flow being re-established at a time t = 0.

After occlusion (S2 open) the aspiration system vacuum climbs to the panel programmed maximum value (in this example 500mmHg) then the pump stops therefore S2 and S3 are both open at this point. After a time in the order of 300 milliseconds (ignoring wound leakage) the eye pressure will have climbed to the bottle pressure Pb. Now the system is set up for a *post occlusion surge* which occurs immediately after the occlusion breaks free. Due to the complexity of the mathematical solution involving two linked sub systems with inertia resistance and compliance, only the graphical solution of the circuit is presented.

Figure 26 shows the inflow rate (brown), the outflow rate (magenta) superimposed on the eye pressure graph (blue/green) with time. This enables a view of the relative timing of all of the important events involved in a post occlusion surge in the peristaltic machine.



When the vacuum at the vacuum sensor near the peristaltic pump falls below the maximum value the pump is instructed to start(switched on again) however the pump only produces motion and observable effects after a delay of around 200 milliseconds due to the electromechanical latency of the pump. In this first instance to see the surge on its own without the influence of the pump re-starting and affecting the profile of the tail end of the surge, the pump starting again has been inhibited. Later we will start the pump at different times to see the effect on the eye pressure.

The interesting results of figure 26 are that the inflow lags the outflow in time. This has been recorded by other experimenters using flow sensors on the irrigation tube and aspiration tube near the probe: Zacharias⁷. The eye's pressure drop during the surge lags the outflow (typical of a compliant chamber). The inflow delay relates to fluidic inertia in the irrigation pathway and the eye's compliance.

It can be seen that a post occlusion surge from 500mmHg will take the anterior chamber pressure to zero or close to near 2.6mmHg in a typical peristaltic setup with a bottle height of 70 cm. If the bottle is elevated, the post occlusion surge magnitude *increases* by a small proportion , however the average chamber pressure increases directly with the bottle pressure , making the eye pressure value, a more positive number, at the surge peak, even though the magnitude of the pressure surge transient increases.

Another interesting feature to be gleaned from figure 26 is that the peak inflow rates are much lower at 46 ml/min, under half in fact of the peak outflow peak rate of 101ml/min. Also the outflow peaks early at 72 milliseconds and the inflow much later and more broadly at 280 milliseconds. The actual volume of each though, integrated over time, the same. The peak inflow is a wider pattern, and the outflow narrower.



Figure 27 shows the eye pressure again during a post occlusion surge with the two flow graphs removed but the pressure at the machine's vacuum sensor on the same graph not to scale. The vacuum in this example starts at 500mmHg prior to the surge and due to the fact the irrigation is disabled in this example arrives at the bottle pressure later. The relative timing is quite clear.

Now we can involve the peristaltic pump and start it up initially at 700 milli-seconds after the surge starts to observe the effect, figure 28:



The red line on the graph indicates when the pump mechanically starts, which is always some delay, about 100 to 200 milliseconds after it is electrically instructed to start. Also after the pump does mechanically start motion, it takes a while to build up the flow, due to the inertia of the fluid in the entire fluidics pathway and the time taken to establish a vacuum, so the eye

chamber pressure drop is delayed and is much more of a *gradual drop* than seen with the surge flow related pressure drop. The constant flow rate of 30ml/min drops the eye pressure *slowly* down to 23.3mmHg and equilibrium is established as per equation 13.

In general as soon as the vacuum in the vacuum sensor area falls below the maximum value, which it does within in the first 20 milliseconds after the surge begins in the eye (this small delay will be explained in the section on transmission line/tube delay) the pump is electrically instructed to start.

The pump motor has a delay in starting due to its physical mass and electrical inertia (inductance) and there is a delay to establish flow. So typically a pump will start its mechanical motion at about 100 to 200 milli-seconds after the surge begins. This is shown in Figure 29:



It can be seen that as a result of the pump starting again during the pressure drop of the post occlusion surge, that the pressure reduction in the eye caused by pump flow then *blends in with the "tail end" of the post occlusion surge*. This is the typical measured post occlusion surge format seen in peristaltic phaco machines. The starting of the pump has no significant effect on the surge peak amplitude and is unrelated to the peak eye pressure dip caused by the early post occlusion surge outflow which has been shown peaks around about 70 milliseconds after occlusion break. The early surge flow is generated secondary to elastic energy in the compliant parts, Cm, at the machine and is not related to pump motion or pump motor control. If the pump could start more quickly and have less delay, then this in fact aggravates the post occlusion surge.



Figure 30 demonstrates the situation where the pump, if it could, starts motion 50 milliseconds after the surge starts. The eye pressure is now taken to -3.8mmHg and the peak is a little more delayed at 240 milliseconds rather than 200 milliseconds.

Some occlusions may break free prior to maximum vacuum and the pump actually stopping, in these cases the surge is also a little aggravated by pump flow. Nothing is to be gained by starting the pump in the forward direction earlier than it characteristically starts in peristaltic machines, typically around 100 to 200 milliseconds after occlusion breaks. On the other hand, if the early part of the surge was reliably detected pump reversal could potentially help the surge, however the pump reaction time and the fluidic effect is too slow and the surge has already peaked before the pump in existing machines could help an actual detected surge. Remarks regarding superior pump control with microprocessors and smart software improving surge control are clearly meaningless, as it is the fluidic's system physics which determines the surge magnitude. The pump can be reversed when a surge is *anticipated* (not detected) to lower the aspiration system vacuum prior to occlusion break and this will be discussed in the section on surge management techniques.

It is worth noting at this point from figure 28 how the pressure drop within the eye, when the peristaltic pump re starts, is very gradual compared to the rapid fall over 200 milliseconds generated by the surge flow. The irrigation flow is better able to compensate for gradual pump related aspiration flow changes. It can be seen though that there is still a small timing difference between the inflow and the outflow after the pump starts up. With a maximum vacuum of 250 mmHg the peak pressure drop, during a post occlusion surge, in the eye will be about 27 mmHg. The "experimenter" or the manufacturer can adjust the machine's flow rate so that the pressure drop will also be about 27mmHg due to constant flow, typically around 30ml/min after the pump starts up. Then if the peristaltic pump starts around the time usual time of 200 milliseconds there is "no significant overshoot". See Figure 31:



However this situation, which has been presented in the past to Ophthalmologists at meetings, to venture to suggest that there is no surge in a particular machine is totally misleading. The experiment setup merely disguises the importance of the rapid surge pressure drop and rapid surge flow which is not adequately compensated for by the fluidic delays on the irrigation pathway. *The surge itself should not be measured as an overshoot* value because this incorporates the pump's flow rate and irrigation resistance product, which is in fact usually zero during the first significant part of the surge to its apex. The surge should be specified as value to which the eye pressure is reduced below the bottle pressure, usually over the first 200 milliseconds, after occlusion break begins with a specific irrigation apparatus, phaco needle and disposable set and phaco machine because the vacuum sensor is compliant. Any remarks regarding flow rates and post occlusion surges therefore are not very relevant, given that the pump is not usually running during the initial significant part of the surge.

Marketing remarks regarding superior software, smart pumps, intelligent microprocessors and so forth resulting in improved surges are not very meaningful. The surge performance and eye pressure over time during the first 200 milliseconds of the surge is well described by equation 43 and does not relate to the pump's motor or control of the motor, merely the compliance of the pump's tubing, the vacuum sensor compliance, the eye's compliance, the bottle pressure and the aspiration vacuum prior to the surge, along with the geometry of the fluidic components with their respective resistances and inertial or L values. The compliance of the aspiration tubing also features and the effect of this is to be explained shortly.

Surges can be "anticipated" and the aspiration system vacuum lowered. This will be discussed in the section in (4-4) below.



Now if we look at figure 32 to illustrate what happens after the surge has occurred and with a constant flow rate of 30ml/min generated by the pump. The result shown again with the yellow needle/sleeve with the 30ml/min value used for the irrigation resistance Ri of $7.36x10^9$ and the 30ml/min value for the aspiration resistance Ra of $2.15x10^{10}$. The vacuum that is generated by the pump to support the flow rate is the interesting feature (green). This is shown in the figure not to scale with the post occlusion eye pressure Pe(t) (blue/green) to scale. The vacuum Pv at the vacuum sensor near the pump stabilises to the value associated with the flow rate of 30ml/min. This is $-7640N/m^2$ or a vacuum of 57.3 mmHg simply as predicted by the vacuum equation 12 for peristaltic machines with a Pb of 51mmHg and the value of Ri, Ra and the flow rate. This means to avoid pump interruption, and reduced flow rates then the vacuum on the machine's panel needs to be set higher than this. If we now change the yellow needle sleeve system for a 2.2 micro coaxial system we get the result shown in figure 33:



In this instance both the Ri and Ra values are now increased due to the smaller calibre annulus and the smaller internal diameter of the micro coaxial needle. The Ri is now $8.36x10^9$ or about 13% higher than with the yellow needle/sleeve and the Ra is $2.89x10^{10}$ or 34% larger than the AMO yellow micro-flow needle. The smaller calibre and higher resistance phaco needle results in some reduction of the post occlusion surge amplitude. This is because increasing the aspiration resistance Ra helps to damp down the surge flow transient.

The increase in Ri on the other hand, degrades the irrigation performance and increases the delays in the irrigation system's response to transient eye pressure reductions. Also the anterior chamber pressure will be lower under any given flow rate and bottle height settings in the un-occluded flow state.

Another consequence of the overall increased Ra using smaller internal diameter phaco needles is that now a vacuum of 12097 N/m^2 or 90.7mmHg is generated at by the pump at the vacuum sensor at the flow rate of 30ml/min. This means that the maximum value on the machine's panel cannot be set below this or the pump will be intermittently interrupted and the flow rate will drop. This moves away from a situation where it becomes possible to run *low vacuum* during phaco while still maintaining a reasonable flow rate and having adequate clearance of lens debris from the anterior chamber and satisfactory phaco crystal cooling. With the surgical technique of initial lens sculpting, prior to cracking, often low vacuums are preferred. However with lens chopping techniques low vacuums may not be often used.

The forces encouraging contact between the lens and the oscillating phaco needle tip are provided by the zonules supporting the lens body, not the fluidics system during sculpting.



Finally figure 34 shows what happens when a large flow resistance of $1.5x10^{11}$ is added to the aspiration system again with the yellow needle/sleeve arrangement. Again the eye pressure Pe (blue/green) and the machine vacuum, at the vacuum sensor (green). In this case the Y axis is

the vacuum Pv to scale, and the eye pressure graph is not to scale. This shows how the post occlusion surge is now completely suppressed and the consequence is that now a very large vacuum of $43,219 \text{ N/m}^2$ or 324 mmHg is generated by the modest flow rate of 30ml/min in the peristaltic machine. This means that you could not set the maximum vacuum on the machine below a value of 324 mmHg without getting interruption of the pump. In the case of a *venturi machine*, with this additional resistance added, then if the vacuum were below 324 mmHg the flow would start to fall off. At 100mmHg vacuum the flow rate would be only around 9ml/min.

CHAPTER 4.

(4-1) EXTERNAL PRESSURE ON THE OUTER WALL OF THE EYE:

If the orbital tissues are tense, or there is speculum pressure then positive pressure applied to the exterior of the eye, we will call it P(orb). This pressure subtracts from the internal applied pressure from the fluidics pathway, specifically the pressure provided by the irrigation source.

This net pressure P(net) = Pe - P(orb) is now the "effective pressure" or "net pressure" which acts on the eye's compliance such that the eye's anterior chamber has a lower volume. Returning to the basic relation for compliance of the eye and using P(net):

Volume (of the anterior chamber) = Ce.P(net). Although a pressure measurement with respect to the atmosphere (zero) would show Pe to be some value, however this is not the *net* pressure applied to the eye as a compliant structure. This means that the anterior chamber will physically begin to collapse, not at zero anterior chamber pressure, but when the anterior chamber is still at the positive value equal to P(orb). Therefore:

Any pressure on the exterior of the eye simply can be regarded as subtracting from the bottle pressure, or subtracting from any eye pressure value present during any flow situation. *Therefore the volume of the anterior chamber experienced by the surgeon will be lower for any given anterior chamber pressure when there is external eye pressure.*

Over the years a shallow anterior chamber for a normal irrigation system pressure has been described as "vitreous pressure" when this really represents pressure on the external globe wall. Typically this is pressure from the speculum or tense orbital tissues after local anaesthetic or associated orbital haemorrhage. It is possible to have real vitreous pressure which crowds out the anterior chamber volume in a case of angle closure glaucoma & malignant glaucoma for example and possibly misdirected "irrigation flow" from excessively high bottle pressure and in cases of vitreous and or choroidal haemorrhage causing volume expansion in the posterior segment.

Now that the basics of phaco fluidics have been examined, we can refine the final model to incorporate the features introduced by the aspiration tubing's distributed compliance. The results of this analysis are quite remarkable. It will be shown how the aspiration tubing confers some unique and remarkable properties to phaco fluidics, especially over the first 50 milliseconds of the initial events such as the post occlusion surge.

(4-2) COMPLIANT TUBING AND TRANSMISSION LINE DELAY PHENOMENA:

We firstly must consider "what flexible tubing is" in terms of its resistance compliance and inertia. We previously looked at this briefly for the irrigation tubing where the effects are minor.

Flexible tubing has a flow resistance, a compliance and inertia of the fluid it contains. In the case of tubing, these three properties are spread linearly along the tubing length.



Figure 35 shows a model for tubing which was alluded to previously in the section on the irrigation system. It indicates that a length of flexible tubing can be regarded as being composed of a number of small elements. These elements are resistance Rx, per very small length dx, and Inertia Lx of the fluid contained per small length dx, and compliance of the flexible material comprising the tube Cx per small length dx. The measured *compliance* per unit length CL for the typical modern 1/16 inch internal diameter *aspiration tubing* under consideration is around $0.34x10^{-12}$ per meter, and *the inertia* per unit length is LL. For the aspiration tubing under consideration the LL value is $5x10^8$ per meter See table 2.

If the compliance per unit length *CL*, multiplied by the small distance dx gives the compliance *Cx*, of the small section of tube under consideration:

$$Cx = CL.dx$$

Equ.45

If we consider that the aspiration tubing initially has a high vacuum within its lumen, for example 500 mmHg (which is a lumen positive pressure of 260mmHg), then under these conditions the tubing wall material has been compressed by 500mHg derived from atmospheric pressure. When the occlusion breaks the pressure at the probe end of the aspiration tubing is now abruptly the eye pressure Pb, or 51mmHg with a bottle height of 70cm.

When the occlusion breaks free a small volume of fluid, dVo, enters a small section of tubing length dx close to the probe. The section of tubing there, has a compliance Cx, over a small distance dx, expands (charges with a small fluid volume). This local change in pressure propagates to the adjacent section of aspiration tubing, and so on, up the aspiration tubing toward the machine. The velocity Ve, of this propagating pressure change is given by the small distance dx divided by a small increment in time:

$$Ve = \frac{dx}{dt}$$

*Equ.*46

The initial small section of aspiration tubing length dx near the probe, with compliance Cx, has charged over a pressure range of 260mmHg to 760 + 51mmHg, eg over a pressure difference of 551mmHg (73448 N/m²) which we have called (Pb + Pv) previously using simply the bottle pressure and vacuum value as absolute numbers to avoid references to atmospheric pressure.

The small volume (or "charge") of fluid, dVo, acquired over the pressure change is:

$$dVo = Cx.(Pb + Pv)$$

Equ.47

In some small time dt, the flow I is dVo/dt and will be represented as Ic, as it is a *constant flow*, therefore from 47 above:

$$Ic = \frac{Cx. (Pb + Pv)}{dt}$$

*Equ.*48

Equ.49

and now substituting from 45:

$$Ic = CL. (Pb + Pv) \frac{dx}{dt}$$

Now substituting from 46 into 49 for dx/dt we have an equation for the initial constant flow of fluid into elastic tubing where there is a pressure step or pressure change, in this case equal to the bottle pressure and the vacuum suddenly applied to the tubing. This happens at occlusion break which typically occurs after the eye pressure has climbed to the bottle pressure (ignoring leakage):

$$Ic = Ve.CL.(Pb + Pv)$$
Equ.50

Equation 50 above describes the interesting fact that as the aspiration tube is expanding initially at the beginning of an abrupt pressure change (at an occlusion break for example) and that the fluid flow is drawn into it at a *constant flow rate* which depends on the initial pressure gradient (in this case the bottle pressure plus the vacuum level), the compliance per unit length of the tubing and the characteristic velocity Ve of the fluid filled tubing. In these examples we ignore the elasticity of the fluid itself as it is dwarfed by the elasticity of the tubing walls.

The characteristic velocity Ve for an analogous electrical "delay line" can be calculated from the formula of 51 below and is equally as appropriate to fluidics because both systems distributed resistance, compliance (capacitance) and inertia (inductance):

$$Ve = \sqrt{\frac{1}{CL.LL}}$$

Equ.51

The foregoing now gives the final result for the constant flow Ic which the aspiration tubing attempts to draw, via the phaco probe body and phaco needle from the anterior chamber initially at least, after occlusion break:

$$Ic = (Pb+Pv) \cdot \sqrt{\frac{CL}{LL}}$$
Equ.52

Taking the values of CL and LL, for the typical aspiration tubing from table 2 the velocity is 76.7 m/sec. This represents a pressure wave, propagating along the aspiration tubing at the fluid filled tubing's "characteristic velocity" toward the phaco machine at 276 km/hr immediately after occlusion break. As the aspiration tube is 2m long, this represents a delay, symbol **td**, of 26 milliseconds from the time that the surge begins in the anterior chamber to when there is some expansion of the tubing at the phaco machine end and vacuum sensor area. This model agrees perfectly with practical measurements:



Figure 36 shows the simultaneous recording of a post occlusion surge from 500mmHg vacuum and a bottle pressure of 51mmHg. This recording was acquired with a pressure transducer within a compliant test chamber similar to the eye, and from the electrical signal derived from the author's phaco machine's vacuum sensor.

The recordings were captured on a dual channel TEK 464 storage oscilloscope and photographed directly from the screen with a digital camera. The mathematical analysis agrees with the experimental recording. The shape of the pressure drop in the chamber is a little more rounded off at its peak than that shown in the previous figures calculated from the mathematical models because the compliance of the test chamber/eye increases at low pressure levels as the eye starts to collapse where as the compliance in the model stays a fixed value.

The vacuum sensor recording cuts off at zero, because the output of this particular sensor is only for vacuum, not positive pressure. *Close inspection of the timing of the machine vacuum and eye pressure waveforms at the start of the surge*(left hand red line) *shows that there is an approximate 20 to 25 millisecond delay between the surge beginning in the chamber (eye), and the vacuum falling at the machine's vacuum sensor* (timing gap between the red lines) In addition the fall in chamber pressure is approximately linear over about twice this time period td, eg about 52 milliseconds. This is explained as follows:

Substituting in the numbers for an initial vacuum of 500mmHg and bottle pressure 51mmHg total (73448 N/m²) and the LL and CL values from table 2 for the aspiration tubing and using equation 52 we find the flow rate for the tubing Ic is an initial constant of $1.92x10^{-6}$ m³/s or 115ml/min. It is interesting that equation 52 has a similar format to Ohms law, and one can see that during this phase of constant flow Ic, that the tubing behaves as a constant resistance of value of $\sqrt{\frac{LL}{CL}}$. This can be called the "Surge Resistance" or "Surge Impedance" of the tubing.

This is the "resistance to flow" it presents to a step pressure gradient over the initial time equal to 2td, or in this case 52 milliseconds, or the round trip delay time for a pressure wave.

From the tubing figures from table 2 *the surge resistance* of the aspiration tubing therefore has a value of 3.83×10^{10} . This is only a stable resistance value over a time period of 2td. If one imagined an infinitely long aspiration line/tube, the flow drawn into it would remain constant indefinitely as the pressure wave propagated up it away from the probe. In practice what happens is that the pressure wave reaching the machine after the delay time td, is *reflected* back toward the probe and takes another time period of td to return to the probe and eye. The *constant flow* conditions after occlusion break therefore initially remain stable for a delay time of 2td, or 52 milliseconds, when the reflected wave returns to the probe.

The reflected wave is out of phase(has been inverted) after the reflection and is then yet again reflected at the probe tubing interface and returned to the machine where reflections continue until they decay away.

Now returning to point B of the ABC's of compliant chambers; if constant flow, or current, is extracted from a compliant chamber, like the eye, then there must be a constant or linear fall in pressure over that time.

The calculations suggest that the tubing is attempting to draw a constant flow from the eye, over the first 52 milliseconds after occlusion break of $1.92x10^{-6}$ cubic meters per second or 115 ml/min. However the tubing is attempting to draw this flow via the resistance of the probe body (small at $9.62x10^8$) and resistance of the phaco needle which is significant, for the yellow micro-flow needle at least around $1.68x10^{10}$, which is the 90 ml/min flow rate resistance value. The *surge resistance* of the aspiration tubing calculated is $3.83x10^{10}$ so there is a total effective flow resistance of $5.60x10^{10}$.

The initial driving pressure at occlusion break is (Pb + Pv) or 73448N/m². The initial flow rate (pressure/ total effective resistance) is therefore 73448/5.60 $x10^{10} = 1.31x10^{-6}$ or 78.7 ml/min flow rate extracted from the eye over the first 52 milliseconds after occlusion break.

Also this initial event, at the beginning of a post occlusion surge, is *nearly completely independent of the irrigation pathway properties*, because over this short initial time frame, the irrigation system for practical purposes is nearly completely inactive due to its inherent delays. Now returning to the basic relation for the eye as a compliant chamber we can say that:

Flow rate into or out of the chamber = $\frac{\text{Ce. } \Delta \text{ Pe}}{\Delta t}$

Where Δ Pe is the change in eye pressure over the time, Δ t, which is 52 milliseconds. Then the eye pressure drop in the eye over this constant flow time will be:

$$\Delta \text{ Pe (drop)} = \text{Flow rate } x \Delta t / \text{Ce} \\ = (1.31x10^{-6} \text{ m}^3/\text{s} \times 0.052 \text{ s}) / 30x10^{-12} \\ = 2271 \text{ N/m}^2 \text{ or } 17 \text{ mmHg.}$$

Therefore a 17 mmHg pressure drop is expected by mathematical analysis & calculation to occur as a close to linear fall in eye pressure over the first 52 milliseconds after occlusion break. This agrees closely to the practical recording of figure 36 where the waveform kink is seen to occur about 17 mmHg below the 51mmHg initial eye pressure value.

To double check this reasoning and calculation, we need to create the equivalent flow circuit and run it in the circuit simulator and add the tubing model:



Figure 37 shows the equivalent flow model of the entire phaco fluidics setup for the peristaltic machine, to show the events which occur at occlusion break at t = 0. The advantage of the model is that as well as being able to visualise the anterior chamber pressure, the inflow and outflow from the tubing can also be graphed. This allows to inspection of the temporal relationships which a high degree of accuracy.



Figure 38 depicts the result of the flow circuit depicted in figure 37. Displayed on this graph initially, to avoid too much clutter, is the eye pressure Pe (blue/green) and machine vacuum Pv

(green) over time after the occlusion breaks free at t = 0. Pv is not to scale but shows the timing of the vacuum at the machine.

Taken from the graph of figure 38, the linear fall in eye pressure occurs over a period of around 52 milliseconds, as predicted by the equations and as found by practical experiment as shown in figure 36. Taking the result directly from the graph of figure 38, the drop in eye pressure over this 52 mill-seconds is 6798 - 4664 or 2134N/m² or 16 mmHg. The equations above predicted 17 mmHg drop over that period and this is in very close agreement. This also closely agrees with the experimental recording of figure 36. Also, as found by experiment and the mathematics, there is a delay in the order of 26 milliseconds between the time that the pressure falls in the eye (at the moment the occlusion clears) and a drop in vacuum at the machine's vacuum sensor.

To demonstrate the subtleties of these events caused by the tubing's transmission line behaviour we can use the circuit simulator to evaluate the entire fluid filled aspiration system, with the aspiration tubing. This time however we artificially hold the pressure in the eye Pe, and the vacuum at the machine Pv constant. This would in fact be the situation if the eye's irrigation system were perfect with zero resistance and zero fluidic inertia, and if we were using a venturi machine where the vacuum Pv in the cassette stayed constant at occlusion break. The flow circuit model for this scenario is depicted in figure 39:



When occlusion breaks, the tubing starts to draw fluid (flow) from the constant pressure source Pe. The flow begins when S2 closes, at just greater than zero. The values for resistance per unit length, inertia per unit length and compliance per unit length of the aspiration tubing are taken from table 2. The total aspiration resistance is $2.15x10^{10}$ (30 ml/ min value) and the vacuum Pv, is set at 3952 N/m² or 29.6 mmHg and the eye pressure is 51mmHg or 6798mmHg.The total diving pressure is therefore 80.6mmHg or 10744 N/m².

This makes the flow rate $10744/2.15x10^{10} = 5x10^{-7}$ or 30 ml/min when stable flow is established. We are not interested in this stable flow state but in the initial time period immediately after the flow begins.



Figure 40 shows the fluid inflow to the aspiration tube at the probe(magenta), and the fluid outflow from the tube(brown) at the machine end beginning when S2 closes, or occlusion breaks and the flow begins at t = 0.

The inflow to the tubing connected to the probe and needle is fairly constant over the first 52 milliseconds at 230nano cubic meters/sec or 14.8ml/min, starting a little higher and falling a little lower over the 52 milliseconds (this tilt downwards is due to the high resistive flow loss in the tubing. With low resistive losses, as say typically seen in an electrical transmission line, this tilt would not be obvious).

Using the surge resistance value calculated above for the tubing of $3.83x10^{10}$, and adding the flow resistance of the phaco needle $7.78x10^9$ (30ml/min figure)and probe body $9.62x10^8$ the total initial total resistance therefore is $4.70x10^{10}$. Then with the applied *pressure step* of $51\text{mmHg} + 29.6 \text{ mmHg} (10744 \text{ N/m}^2)$ then the initial flow rate would be expected to be $10744/4.70x10^{10} = 228n$ or 13.7ml/min which is close to the 230n cubic meters per second found on the graph from figure 40. Therefore, ignoring the tilt due to the tubing's high resistive losses, the equations predicting the initial surge resistance and the surge flow rate and characteristic velocity agree with the circuit simulators output and this agrees with the delay timing seen on practical experiment with the fluidics setup and experimental result of Figure 36.

Notice from figure 40, that the outflow from the aspiration tubing, at the machine end is 26 milliseconds late compared with the start of the surge or the initial inflow to the aspiration tubing. This is the cause of the delay between the post occlusion surge beginning in the eye and being detectable at the machine's vacuum sensor. Also notice that after 52 milliseconds, the inflow to the aspiration tube suddenly jumps upwards. This corresponds to the positive pressure wave which travelled up the tubing being reflected and inverted on reflection and travelling back down the tubing and arriving at the probe a time of 2**td** later, or 52 milliseconds.

Now we can inspect a graph with the aspiration tubes flow profile combined with the post occlusion surge eye pressure profile and machine vacuum. This is depicted in Figure 41 which is the graphical solution of fig 37 but this time including the fluid inflow to and out of the aspiration tubing:



Figure 41 would have been very difficult to understand without evaluating the aspiration tubing under a fixed pressure step as was done in fig 40 because now the stair step inflow and outflow patterns of the aspiration tubing itself are now part of the *outflow pattern* of the aspiration system, peaking at 70 millisecond, as seen in figures 26 through 30 where the tube's compliance was ignored.

In figure 41 the Y axis is the flow rate in nano cubic meters/sec. The eye pressure graph Pe, and the Vacuum pressure graph Pv are overlaid to see the relative timing. The eye pressure dips down to zero, typical of a post occlusion surge from a 500mmHg maximum vacuum. The peristaltic pump starts a 200mSec so the tail of the surge has its usual format. The kink in the eye pressure waveform seen in this comprehensive model, and in the experimental recordings corresponds to the first pressure wave reflection from the machine after occlusion break. As seen in the graphical solution, the beginning of the fall in eye pressure is a *linear drop* over a time frame of 52 milliseconds.

(4-3) HOLDEN VELOCITY EQUATION FOR FLUID FILLED PLASTIC TUBING.

The "delay line" property of the aspiration tubing is independent of the applied pressures and relates to the characteristic pressure propagation velocity for the fluid filled plastic tube in question and the length of the tube and its values of compliance per unit length and L value per unit length of the particular tube. Using the authors equation 37 for the compliance of plastic tubing and the equation for the reactive pressure coefficient and equation 51 above for the characteristic velocity for pressure waves propagating along plastic (compliant tubing) it can be shown that the characteristic velocity **Ve** for fluid filled plastic tubing is:

$$\mathbf{Ve} = \sqrt{\frac{\text{T.k}}{2.\rho}}$$
 (Holden Velocity equation for fluid filled plastic tubes)
Equ.53

Where p is the density of the fluid (1000) and T the thickness of the tube wall and k the stiffness constant of the plastic the tube is made from. The characteristic velocity is independent of the diameter of the aspiration tubing. This means a tube with a thicker wall or stiffer plastic has a higher characteristic velocity. This reduces the delay between the pressure drop of the post occlusion surge occurring in the eye, and this being detectable at the machine's vacuum sensor.

Equation 53 is very similar to the standard physics equation for the velocity of pressure waves in media where the product T.k/2 of the tubing represents the *elasticity*, or stiffness of the media. The elasticity in this case dominated by the relatively soft plastic tubing rather than the elasticity of the relatively incompressible fluid.

The actual time delay, before a pressure event occurring in the eye, and therefore being detectable at the machines vacuum sensor is equal to the length of the aspiration tubing divided by characteristic velocity of the tubing. With the characteristic velocity of 76.7 m/sec for the 2 meter long aspiration tube listed in the table it therefore takes 26 milliseconds for the pressure transient generated at occlusion break in the eye to travel along the 2m length leading to the machine's vacuum sensor.

The typical 26 millisecond delay, between the occlusion break in the eye and the pressure beginning to increase, or at least be detectable, at the machine's vacuum sensor area (in the peristaltic machine only) has implications in the early real time detection of post occlusion surges by the machine's vacuum sensor and electronics. On the flip side of this coin, a slower velocity and a longer transmission delay assists in lens fragment holding forces at occlusion. This will be explained in the section on lens fragment holding forces.

The reason why the consequences of the step-wise patterns of aspiration tube flow are difficult to see on recordings of pressure in the eye, or vacuum at the vacuum sensor during the surge is quite simple. The stepwise initial constant outflow from the eye, or the inflow to the aspiration tube, results initially only in the linear decrement in the eye pressure during the first 52 milliseconds after occlusion break. Then the reflected pressure wave and sudden step in flow rate causes only the small kink in the eye pressure waveform at that point and this has been missed

by investigators in the past. After this time the reflected waves have lower amplitudes and the effect of the eye's compliance is to simply filter them (analogous to a filter capacitor) out of the eye's pressure recording.

The compliance at the machine's vacuum sensor and pump tubing area also acts as a "filter" in the peristaltic machine and the initial outflow from the aspiration tube, which is fairly constant over the time of 26 milliseconds to 78 milliseconds (see figure 40) simply results in a near linear vacuum decrement (increase in pressure) over that time. This can be seen on the model figure 41 and the experimental recording figure 36 the change in waveform shape is barely visible. The initial vacuum decrement at the sensor blends in with the falling vacuum level as more fluid volume arrives at the machine end if the tubing. In essence the reflections at the eye and probe on one end of the tubing, and the machine vacuum sensor area on the other end of the tubing occur because the impedance of those structures is different to the aspiration tubing. Therefore the energy is partially reflected and partially transmitted at those interfaces. The pressure waves travel up and down the tubing between the probe and the eye, and the fluid flow adopts a stair-step pattern at any time that the aspiration system and its tubing, is subjected to an *abrupt pressure change*. This occurs during either occlusion of flow or at an occlusion break.

It is interesting to note, that printed recordings of post occlusion surges presented by the authors previously cited, Zacharias⁷ in *non-compliant chambers* as an eye substitute, did show significant *pressure steps*, corresponding to aspiration tubing's reflected pressure waves, but these were not remarked upon by the authors. The pressure steps in the chamber representing the eye, were in fact easily visible because there was no significant compliance to filter them out. In addition because the compliance of the chamber representing the eye Ce was low, the amplitude of the post occlusion surge was higher because the ratio Cm / (Cm + Ce) was higher.

It is also interesting to note that, ignoring the flow steps, the *average* flow increase to equilibrium ,when a pressure gradient is applied across the aspiration tubing, as shown in figure 40, is an inverted exponential as it would be if there were no tubing compliance and only fluidic inertia and flow resistance.

Application of Transmission line theory to Phaco Fluidics:

When we apply this *transmission line tube model* to phaco fluidics it provides exceptional insight it the behaviour of the machine's fluidics not only for the first 52 milliseconds after occlusion break but for the dynamic properties which occur at *occlusion*, which are even more interesting. A common statement relating to *venturi* machines is that they have the advantage of "instantaneous vacuum" being present with occlusion at the needle tip. Investigating this has lead the author to some remarkable and interesting results discussed further in the section on *Lens Fragment Holding Forces in Phaco*.

(4-4) POST OCCLUSION SURGE MANAGEMENT IN PERISTALTIC MACHINES.

The three major phaco machine manufacturers, Alcon, AMO and B&L have all recognised the importance of keeping the aspiration system compliance as low as possible. This includes the compliance of the aspiration tubing, pump tubing (or soft parts) and the vacuum sensor.

However there is one positive spin off from increased compliance in the aspiration system in that the additional compliance in that area helps "smooth" or filter out pressure pulsations in the aspiration system caused when the peristaltic pump rollers strike and begin to compress the tubing.

Peristaltic pump pulsations can be reduced by stretching the pump tubing around the rollers and narrowing the internal volume of the tubing and therefore reducing the volume displaced when the rollers initially compress the tubing. The pump must be run at a higher speed to compensate for this. Smaller diameter pump heads and shorter smaller internal diameter pump tubing helps. In all these cases the pump rotational velocity must be higher for the same flow rate. In 2006 AMO changed their Sovereign tubing to a type which more resembles the low compliance aspiration tubing used by Alcon and B&L in its physical properties. However the author has noticed that pump roller pulsations are more readily visible in the anterior chamber now however they do not cause any problems. Also AMO increased the bore of the Irrigation tubing which assists in lowering the flow resistance and reducing the inertia of the irrigation fluid which is a good move.

AMO also introduced CASE as a surge management technique as an upgrade for their Sovereign machine. This is a system that essentially anticipates when a post occlusion surge is likely to occur because a high occlusion vacuum has been attained. The peristaltic pump is then reversed to lower the aspiration vacuum below the maximum programmed vacuum, perhaps to a level set at ½ that. Therefore when the occlusion breaks free later on, the magnitude of the surge behaves as a typical surge from the lower vacuum level so therefore the surge is lower. It is a system which prevents the surgeon having programmed high max vacuums for more than the brief time before the pump is rolled back. So the *average* occlusion vacuums the surgeon has are lower. However the surgeon has access to higher vacuums if albeit briefly. Due to the unpredictable timing and nature of an occlusion break, the break may possibly occur prior to pump rollback at the maximum vacuum level. This will produce a surge proportional to that maximum level and not the lower level.

A true peristaltic machine based surge neutralisation technique would involve the detection of actual surges at the vacuum sensor and a methodology to reduce the surge effect. Due to the inherent transmission delay in the aspiration tubing there will always be a 26 millisecond, or thereabouts, delay before electronic detection of occlusion break at the vacuum sensor. The peak outflow occurs at about 70 milliseconds after occlusion break. This leaves only 44 milliseconds to provide a correction (such as neutralising the aspiration system vacuum) and perhaps only 10 to 20 milliseconds in reality as the correction needs to be made prior to the surge outflow peak occurring at 70 milliseconds. Current peristaltic pumps cannot act this quickly to reduce the vacuum. It would require a specialised fluidic system which does not exist in current phaco machines.

Other non- machine based surge reduction methods include adding resistive apertures (smaller internal diameter phaco needle or flow restrictive devices) to the aspiration system. While these do successfully reduce the peak surge flow as shown in figure 34 they have the problem that increasing the total flow resistance in the fluidic pathway means that for any given flow rate generated by the pump, that the associated vacuum with be higher. To fully suppress the surge peak in the peristaltic machine, a flow resistance in the order of 1×10^{11} needs to be added to the aspiration system. When that is done, the vacuum level that is generated by the pump at a flow rate of 30 ml/min, rather than being the usual 57mmHg is now an enormous 324mmHg. Therefore the panel maximum vacuum setting on the machine cannot be set at a lower value than this or there will be constant interruption of the pump rotation, and a seriously reduced flow rate. Low flow rates, give less cooling and are associated with wound burn and reduces the clearance of fine lens debris from the anterior chamber.

(4-5) FLOW OCCLUSION CHAMBER INSTABILITY MANAGEMENT VENTURI MACHINES:

The fundamental cause of chamber instability in venturi machines is high flow rate related anterior chamber pressure loss and the eye pressure returning to the bottle pressure with occlusion.

Actual flow rates are not known generally to the surgeon unless time is taken to measure them with a specific disposable set, irrigation device and phaco needle, or unless the manufacturer provides the data for a specific setup. In general, flow rates leading to or over 50 to 60ml/min cause a pressure loss along the irrigation pathway resistances which about equal the bottle pressure 51 mmHg (70 cm bottle). Therefore at these flow rates, anterior chamber collapse at that bottle height is very likely. This problem limits the useable maximum vacuum level on the machine.

In general the approach to solving the above problems has been to increase the aspiration pathway resistance by lengthening the aspiration tube and coiling it up. Another method is to use aspiration tubing with a smaller internal diameter and another method involves placing a filtered small bore (resistive) aperture in the aspiration tube. This helps to encourage turbulence at higher flow rates and therefore increased flow resistance under high flow rate conditions. Small apertures also undergo turbulent flow and generally have square law resistance properties rather than linear resistance typical of laminar flow. The entry to a small aperture or hole is such that the flow resistance is proportional to the flow rate rather than being independent of the flow rate as it is in tubing with established laminar flow. Small internal diameter phaco needles also act as square law flow resistances.

The problem remains however that the addition of flow resistance to the aspiration pathway alters the dynamics to the extent that always a higher vacuum(either required in venturi, or generated as a result of flow rate in peristaltic) is associated for a given flow rate. In both cases this limits the use of low vacuum levels without reduction in flow rates and therefore more

wound heating occurs. This could be avoided by removing the flow restrictor device to allow low vacuum levels during different stages in phaco cataract surgery.

(5-1) LENS FRAGMENT HOLDING FORCES - lens fragment chatter.

This is one of the most interesting areas in phaco fluidics and rarely discussed.

Again and with the physical methods used so far in this book, the best way to understand this is to start with a simple scenario first.

Consider fluid flow travelling along at a constant rate in a rigid tube, under an applied pressure P on the left hand side of the tube, pushing the fluid to the right side of the tube where the pressure is zero at its far end.

Now consider what happens when the left hand side of the tube is suddenly blocked with a solid physical obstruction. The applied pressure P is now isolated from the fluid flow by the obstruction and this pressure is now applied to the obstruction itself and not the fluid.

On the side of the obstruction, in contact with the fluid, an interesting event occurs. The fluid, prior to the obstruction, had kinetic energy of the flow. The flow was abruptly stopped and this kinetic energy now is converted into an abrupt negative pressure transient applied to the obstruction, over the time period, whatever that may be as the fluid flow abruptly decelerates to zero.

One way to analyse this is to use F = ma again where acceleration (or deceleration where the velocity is negative) is defined as $\Delta v / \Delta t$. Where v is the velocity t is time and Δ is some small time increment. Therefore we can say that:

F. $\Delta t = -m \Delta v$ Equ.54

Or in other words the Force Δ time product equals the change in momentum or m. Δ v. This force time product in Physics is known as an *impulse*.

In the case of the fluid flow example above, the fluid velocity falls to zero after some time, Δ t. Now if we go to a practical example: Consider a 2 meter length of *rigid metal aspiration tubing*, with a 1/16" inside diameter. At a flow rate of 30 ml/min, the flow velocity is 0.25 meters per second. The mass of fluid in the 2 meter length of tubing is 3.96×10^{-3} kilograms. We will "assume" initially that the flow stops abruptly within a millisecond after it is interrupted by the solid occlusion. Using the equation above the impulse of force over the 1 millisecond time frame on the fluid side of the occlusion is very close to 1 Newton. The cross sectional area of the tube is 1.98×10^{-6} m² so the negative pressure impulse (vacuum) applied to the fluid side of the occlusion in the tube is $1/1.98 \times 10^{-6}$ or 505050 Newtons per square meter or a massive 3788 mmHg. If the deceleration occurs over a longer time frame, for example 10 milliseconds, the vacuum impulse would be 378 mmHg. The reason why this vacuum impulse is so large is that the tubing and the fluid are relatively incompressible so the flow stops rapidly. If there was an air bubble in the tubing near the occlusion the rate of deceleration of the fluid would be cushioned and the vacuum impulse generated at occlusion would decrease as the bubble expanded like a spring. Or the deceleration is reduced if the tubing is flexible.

If, as it is in practice, the aspiration tubing itself is flexible in phaco fluidics, then the vacuum impulse depends on the elastic properties of the aspiration tubing. We have already learnt that the tubing itself has transmission line delay phenomena due to distributed inertia and compliance. This in fact is the property which influences the vacuum impulse and its time profile at occlusion. The vacuum impulse is generated on the phaco needle lumen side of the lens fragment which is occluding the phaco needle tip. At the same time this is happening, the eye pressure is climbing toward the bottle pressure, from whatever the eye pressure value was prior to occlusion. We have previously learnt as shown in figures 10 through 13 that this process takes around 400 milliseconds.

Once the needle is occluded with a fragment, the forces on either side of the fragment (the pressure in the anterior chamber and the vacuum in the phaco needle lumen) are responsible for holding the fragment on the needle tip. After a time these pressures stabilise at the sum the bottle pressure Pb and the machine vacuum Pv multiplied by the cross sectional area of the phaco needle lumen facing the fragment. Therefore the lens fragment holding forces are reduced with a small phaco needle lumen (unless its mouth is flared) and increase with increasing bottle pressure and machine vacuum levels.

The *dynamic* vacuum build up situation immediately after occlusion is not what one might intuitively expect. Also in the *peristaltic* machine, as well as the *initial* fragment holding forces there is a further build up of occlusion vacuum to a value programmed by the surgeon. In general though the vacuum used in the un-occluded state in *venturi* machines are higher than those that occur in peristaltic machines because the aspiration flow resistances of the phaco needle and tubing, historically at least, have been higher. With the use of high vacuum tubing and flow resistive (restrictive) devices the *venturi* machine users may use even higher unoccluded vacuums. The flow rates in venturi machine setups are generally higher than peristaltic and it is the kinetic energy of this flow which is converted to impulse vacuum energy at an abrupt occlusion.

The *transient or dynamic vacuum levels*, achieved immediately after occlusion, on the phaco needle lumen side of the fragment are higher than the vacuum at the machine. This is due to the deceleration of the moving fluid and the *impulse vacuum* generated by that. The impulse vacuum can be regarded again as a form of *reactive pressure* which is associated with deceleration of the fluid in the aspiration system at occlusion.

Once again to help understand this we employ an equivalent flow circuit to analyse the behaviour of a phaco fluidics system at *occlusion make* (the time that occlusion forms).



Figure 42 shows a flow diagram for a *venturi* machine with a constant vacuum of 57 mmHg at the machine cassette and a 70 cm bottle (51 mmHg). When S 2 closes at t = 0 zero this corresponds to occlusion break. When S2 opens, this corresponds to a total flow occlusion at t = 1 second.

The values of Rn, Ln, Rp, Lp are the respective values of the phaco needle resistance and inertia and the probe body resistance and inertia leading to the needle. The tubing has the RL, LL and CL values are for the typical aspiration tubing from the table. The irrigation pathway has Ri and Li values for the yellow needle and sleeve.

We run the flow circuit of figure 42 in the simulator. A plot of the pressure in the eye Pe, and the vacuum in phaco needle lumen near the occlusion is plotted in figure 43.



In the un-occluded state the pressure in the phaco needle mouth or entrance area is very close to the anterior chamber pressure of 23mmHg. The Y axis of the graph is the machine vacuum (beige colour) and the eye pressure (blue) in K or 1000 N/m^2 . The irrigation flow (brown) is included not to scale but to show the relative timing.

When the occlusion forms abruptly at t = 1 second there is a transient vacuum impulse immediately after occlusion, to about 2.7 to 3 times the machines vacuum level, to 157 mmHg and this is sustained for **2td** or 52 milliseconds. Then the reflected pressure wave arrives, phase reversed (eg a positive pressure pulse) taking the pressure on the phaco needle lumen side of the lens fragment to a value nearly equal to that of the pressure on the anterior chamber side of the fragment. This produces a momentary loss of lens fragment holding force and the fragment could bounce off the needle tip (release risk time) an then be re-acquired resulting on oscillation or chatter of the fragment on the tip.

After this the pressure in the needle lumen reverses again and continues until the oscillations within the aspiration tubing die down and the phaco needle lumen vacuum settles on the vacuum level at the machine. While these processes are occurring, the pressure in the eye (blue) ignoring wound leakage, is climbing progressively toward the bottle pressure, the time frame there depending largely on the irrigation flow resistance as previously outlined.

If the vacuum level at the machine is increased to 100mmHg, and the flow rate increases to 44ml/min, the result is shown in Figure 44 and the impulse vacuum peaks to over 255mmHg.



If the aspiration tube is lengthened from 2.0 meters to 3.0 meters, all other things being equal, the result is shown in figure 45. This confers some advantage as the reflected pressure wave arrives later and with a lower amplitude.



It should be noted that the oscillations and reflected pressure waves in aspiration tubing only occur when either the occlusion break or the occlusion make are abrupt. For example if the needle is slowly occluded over a long time frame of 50 to 100 milliseconds the pressure changes are very gradual and the impulse vacuums very small and the abrupt pressure steps are reduced. The same applies to a slowly breaking occlusion.



Figure 46 shows the result with a *peristaltic* machine. In this example the un-occluded vacuum is 57 mmHg and the maximum vacuum is 300mmHg. A similar process occurs initially, except that after a time the vacuum level begins to increase due the pump removing fluid from the aspiration pathway after occlusion. The vacuum goes off scale on the graph and settles on 300mmHg or -39.990K N/m² which is off the graph below, then the pump is instructed to stop because the measured vacuum equal the value programmed by the surgeon on the panel.

At occlusion break t = 0, a typical post occlusion surge is seen peaking around 200 milliseconds and lowering the anterior chamber pressure to 9.6 mmHg.

Summary of lens fragment holding forces in Phaco:

Abrupt occlusions at the phaco needle result in *reactive* vacuum levels, or *impulse vacuums* which are higher than the un-occluded vacuum level at the machine prior to occlusion. This is caused by the kinetic energy of the un-occluded flow of fluid, in the aspiration system, being converted to pressure or vacuum energy at occlusion. The impulse vacuum corresponds to the forces required to decelerate the moving body of fluid in the aspiration system. This is one example of the conservation of energy as outlined by the Bernoulli equation.

In general the vacuum levels in the un-occluded state have been higher in the venturi machine, and the flow rates higher, than in the peristaltic machine fluidics systems. This has in the past given the users of venturi machines the benefit brisk and powerful lens fragment holding forces and good follow-ability of lens fragments into the phaco needle mouth. This enhances the efficiency of ultrasound emulsification of lens fragments as there is improved contact between

the phaco needle tip and the lens fragment. This has made it difficult for some venturi users to switch to peristaltic machines. If the flow rate in the peristaltic machine is increased then the apparent lens fragment holding power will increase at occlusion and the follow-ability will improve.

For any given flow rate the un-occluded vacuum level in the peristaltic machine increases with additional flow resistance in the aspiration pathway. For example smaller internal diameter phaco needles or flow restriction devices. This decreases the numerical value of the difference between un-occluded vacuum level and the occlusion vacuum level in the peristaltic machine and therefore shortens the rise time all other things being equal. Flow restrictive devices are however are a dual edged sword, in that their incorporation into the aspiration system in either type of machine means that low vacuum levels cannot be used without attenuation of the flow rates.

In the venturi machine the flow rates are attenuated due to the larger total system flow resistance Rt .On the other hand in the peristaltic machine the flow rates can become attenuated when the vacuum measured at the machine's vacuum sensor is higher than the maximum value set on the machines panel and the pump rotation is interrupted.

Aside from the oscillations of vacuum seen with abrupt *occlusion make*, due to the aspiration tubing's transmission line delay properties, the user of either type of phaco machine can regard the lens fragment holding forces, at least 500 milliseconds after occlusion, as being proportional to the sum of the bottle pressure and machine vacuum settings, and proportional to the roots sectional area of the mouth of the phaco needle. Flaring the entrance to the needle therefore, where possible is a helpful feature.

The *fluidics energy* in phaco-emulsification cataract surgery is responsible for *pushing* the cataract out of the eye and the *ultrasound energy* is responsible for preventing permanent or long lived occlusions of flow due to hard lens fragments, as this inhibits the fluidics. If the lens or cataract is soft enough, no ultrasound energy is required for fluidics energy to remove it. The lens fragment holding forces generated by the fluidics system are responsible for the interaction of the fragment and the phaco needle. The ultrasound energy is primarily a repulsive force (longitudinal ultrasound). Efficient lens fragmentation with ultrasound energy therefore relies on a balance of energies between the fluidics and the ultrasound. With the correct balance of energies there is neither anterior chamber over pressurisation or dynamic anterior chamber instability nor is there excessive heating of the phaco needle, or any other significant toxic effects from the ultrasound energy. Under these circumstances the ultrasonic and fluidic stressors on the eye during surgery are minimised and the surgery is safer.

(5-2) SUMMARY OF FLOW CIRCUIT MODELLING - applications to Phaco.

We now have a new powerful tool in Ophthalmology to improve our phaco machinery and instrumentation. This will lead to reduced complications from cataract surgery and improved safety for patients. This system enables:

- # Verify the performance of existing disposable sets and fluidic surgical instruments such as irrigating choppers, sleeve annuli and phaco needles.
- # Understand the influence of machine settings, bottle heights and maximum vacuum settings on anterior chamber stability.
- # Understand the impact of wound leakage and orbital or speculum pressure on the chamber stability.
- # Understand the lens fragment holding forces in phaco.
- # Understand the differences between peristaltic and venturi machines, or any newer machines with altered fluidic properties.
- # Test the design of any new fluidic instrument by measurement of their geometry and calculating their properties of Resistance (using Poiseuille's or Darcy- Wiesbach) and fluidic Inertia (using the reactive pressure coefficient) and compliance (directly measured from the object, if any) as has been done for the items in table 2. Then these parameters are inserted into flow circuit and run in a circuit simulator to obtain graphical results. These include dynamic flow rates, which are more difficult to obtain by practical experiment than are dynamic pressure levels, without influencing the system being measured.
- # Alter the design of objects in the fluidic pathway for objective improvements in performance. Any improvement assists chamber stability and therefore reduces the chances of tissue damage. This improves surgical safety, patient safety and therefore would be expected to improve visual outcomes that follow those.
- # Create new surgical fluidic devices which help to solve chamber instability with the knowledge gained from the fundamental equations for resistance, compliance and inertia of the objects in a phaco fluidic system.

(6-1) PRACTICAL ADVICE FOR SAFE FLUIDIC SETTINGS.

Finally after the fluidics analysis outlined in this book, one might ask what simple and practical pearls have been gained which can be immediately introduced to the operating room?

Peristaltic Machines:

From the analysis we learnt that the average flow rates are stable, set by the surgeon on the panel and conservative flow rates of 30ml/min are safe with a 70cm bottle. Higher flow rates are accommodated with higher bottle levels, but be cautious of the fact that a 1 meter bottle height generates around 70mmHg IOP. High IOP's & deep chambers stress the zonules and ciliary body and your view can be impaired, increasing the risk of capsule rupture due to a poor view and steeper downward angulation of the probe needle. In general flow rates above 50ml/min should be avoided as it necessitates greater bottle heights to support this, however some surgeons are trying these for rapid removal of fragments and lens nucleus rotation techniques but the process might in the long run prove to be more risky for inadvertent tissue damage especially with more junior surgeons.

Flow rates below 20ml/min are better avoided (in phaco mode) as there is reduced phaco crystal cooling and the potential for wound burn.

The flow rate at occlusion break and the 200 milliseconds thereafter (post occlusion surge flow) can be problematic at high vacuum levels. In general a maximum vacuum level of 500 mmHg will generate a peak outflow which will drop the anterior chamber pressure transiently by about 50mmHg and a vacuum of 250 mmHg will drop it by about 25 mmHg in a typical peristaltic machine in phaco mode. Assuming a bottle height of 70 to 80 cm, one should be cautious of raising the maximum panel vacuum level above 250 to 300mmHg to avoid significant post occlusion surges.

Vacuum Machines:

The flow rates are vacuum dependent and setting of higher vacuums will result in higher flow rates. The anterior chamber pressure drops with the flow rate because the bottle's pressure energy is dissipated by the flow resistance in the irrigation pathway. Therefore the IOP will swing between the bottle pressure and a lower value roughly proportional to the magnitude of the flow rate. A vacuum level is always reached where the flow rate is such that the entire bottle pressure is dissipated to zero and the eye pressure approaches zero. Typically with a bottle height of 70 cm, this flow rate would be around 60ml/min, and the vacuum that generated that flow rate could be anywhere from 180mmHg to 400mmHg depending on the flow resistance in the aspiration tubing and phaco needle system. ("high vacuum tubing" has a higher flow resistance).

For both types of machines create snug wounds to avoid leakage and avoid speculum pressure where possible. A folded gauze swab under the superior edge of the speculum, between it and the lateral orbital rim area can help reduce pressure on the globe by lifting the speculum arm upwards away from the globe.

Quick Test:

A simple method is presented for the surgeon to quickly check if the machine vacuum level (vacuum machine) or the flow rate (peristaltic machine) are within "safe limits"

Most people have a 30 cm ruler (1 foot) or can guess the length of it fairly closely. A 30 cm water column has a pressure of around 22mmHg, which is a suitable minimum level IOP during phaco cataract surgery. This allows for any pressure lowering effects of wound leakage or chamber shallowing effects of pressure on the exterior of the globe.

Therefore if the phaco machine (either type of machine) is set up with the surgeon's preferred bottle height and machine settings and the phaco probe has its test chamber (condom) on as it does after the machine is primed, then the surgeon fully depresses the machine pedal to ensure maximum flow rate. The probe is then held one ruler height (30 cm) above the eye. This subtracts 22mmHg bottle pressure from the test chamber. If the chamber does not collapse under these circumstances, then there is at least 22 mmHg IOP at whatever flow rate is being generated in combination with whatever bottle height is set. If the chamber is seen to collapse, then the flow rate is set too high on the Peristaltic machine, or the vacuum level is set too high on the Vacuum machine and need to be reduced.

In the case of the peristaltic machine, with the probe a ruler height above the eye another test can be performed. The aspiration tube can be kinked over (occluded) near the probe, and the pedal depressed, then the aspiration system vacuum will climb to the maximum value set on the panel. With the tube suddenly un-kinked (occlusion break) the chamber can also be observed. If it is seen to transiently collapse (buckle inwards with a indented curvature) then the post occlusion surge is lowering the IOP below 22mmHg and the panel maximum vacuum is set too high and needs to be reduced.

The ruler height above the eye method also takes into account the actual distance between the bottle and the eye as often the patient's head is in a non standard position and the bottle height read off the machines panel is not accurate.

This is shown in Figure 47 and is called THE RULER TEST.


(7-1) **BIBLIOGRAPHY**

- Physics for biology and pre medical students.
 D. M. Burns & S.G.G. MacDonald.
 2nd Edn. Addison-Wesley Publishers, London.
- Transport Phenomena.
 R. Byron Bird, Warren E. Stewart, Edwin N. Lightfoot.
 2nd Edn. John Wiley & Sons Inc.
- Handbook of Hydraulic Resistance.
 I. E. Idelchik.
 3rd Edn. Begell House Inc.
- Akahoshi. Outcomes in Ophthalmic Surgery. Vol 1, Number 4, Fall 2005.
- 5 The Art of Electronics.
 Paul Horowitz, Winfield Hill
 2nd Edn. Cambridge University Press.
- 6 Advanced Engineering Mathematics. Erwin Kreyszig.
 7th Edn. John Wiley & Sons, Inc.
- 7 Volume-based characterisation of post occlusion surge. Zacharias & Zacharias. Journal of Cataract and Refractive Surgery. Oct 2005, Volume 31, Number 10.
- 8 Principles of Electricity.
 Page Adams.
 Chapman Hall, LTD. First printing 1931.
 1943 Edn (Rare book- Photo copy relevant section available from Dr. Holden)
- 9 Comparative Analysis of the fluidics of the AMO prestige, Alcon Legacy, and Stortz Premier Phacoemulsification systems. Hans R Wilbrandt. Journal of Cataract and Refractive Surgery. Vol 23, June 1997.

	RESISTANCE	INERTIA	COMPLIANCE
(7-2)TABLE 2.	J.s.m ⁻⁶	$J.s^2.m^{-6}$	$m^{6} J^{-1}$
PHACO FLUIDICS			
SYSTEM OBJECTS	Calculated(linear)#	Calculated Reactive	
(Fluid carrying objects)	or measured at 30 and 60 or	pressure coefficient-Holden	Measured Values
	90 ml/min.		
SILICONE IRRIGATION TUBING (3.5m) 1/8" ID	$1.40 \times 10^{2} \#$ BI - 4×10 ⁸ /m	$4.42x10^{\circ}$ $I_{L} = 1.26x10^{8}/m$	$\frac{17.5 \times 10^{-12}}{CL - 5 \times 10^{-12} / m}$
PROBE BODY	$9.60 \times 10^8 @ 30$	6.79×10^7 (Tubular part)	
IRRIGATION SIDE			
120mm x 1.5mm ID. 4 63 /3 55m x 20mm	$1.12x10^9 @ 60$	2.88×10^6 (Annular part)	
IRRIGATING CHOPPER HANDLE	9.56 $x 10^8 \#$	7.55×10^7	
L=150mm, 1.59mm ID			
Improved Chopper Handle	$1.07 \ x 10^8 \#$	2.52×10^7	
IRRIGATING DEVICES:			
AMO Yellow sleeve & needle:	3.48 x10 ⁹ @ 30 Free	1.38×10^{7}	
11mm-1.40 I.D sleeve 0.9-1.0mm O.D needle	4.52 x10 ⁹ @ 30 Eye		
(I.D = internal diameter)	4.09 x10 ⁹ @ 60 Free		
(O.D = outer diameter)	5.32 x10 ⁹ @ 60 Eye		
	a aa w109⊖ aa E		
AMO Blue sleeve & plain 19g	$2.89 \times 10^{-6} @ 30 Free$	7	
needle:	$3.40 \times 10^{10} \oplus 50 \text{ Eye}$	$1.05 x 10^7$	
1.10-1.25mm O.D needle.	450×10^{9} 60 Free		
	4.50 X 10 @ 00 Lyc		
R&I palo Rivo sloovo & Miero.	3.19 x10 ⁹ @ 30 Free		
flow Plus Needle:	3.66 x10 ⁹ @ 30 Eye		
11mm-1.57 I.D sleeve 1.17mm	3.60 x10 ⁹ @ 60 Free	1.29×10^{7}	
O.D grooved needle	4.14 x10 ⁹ @ 60 Eye		
8	4.25×10^9 @ 30 Free		
ALCON 2.2 pink sleeve coaxial.	5.52×10^{9} @ 30 Eve		
9.5mm-1.25mm I.D sleeve	5.45×10^9 @ 60 Free		
otonini Oto neodio.	7.08 x10 ⁹ @ 60 Eye	$1.31 x 10^7$	
	-		
Standard CHOPPER L= 30mm LD=0.8mm O D=1.00 to 1.25mm	2.98×10^9	$5.79 x 10^7$	
1.D=0.01111,0.D=1.00 to 1.201111	$4.33 \times 10^{9} @ 30$		
	6.04 X 1 U ² @ 60		
IMPROVED CHOPPER DESIGN	$3.19 \times 10^9 @ 30$	3.00×10^{7}	
	$3.75 \times 10^{9} @ 60$	5.00 % 10	
Fittings: Joiners/bottle spike/chamber	$3.56 \times 10^8 \text{ #}$	4.00×10^{7}	
outlet.	4.80 $x10^8$ @ 30		
	7.70 $x10^8 @ 60$		
IRRIGATION SYSTEM			
EXAMPLE TOTALS in Eye:	AMO blue sleeve/nlain 190	B&L Blue sleeve and	ALCON 2.2 cogrigly
AMO Vollow slama readla	needle:	$Ri = 6.50 \times 10^9$ @ 30	$Ri = 8.36 \times 10^9$ @ 30
$Ri = 6.32 \times 10^9$ @ 30(out of eye)	Ri = 6.30×10^9 @ 30 Ri = 7.70 × 10^9 @ 60	$Ri = 7.43 \times 10^9 @ 60$	$Ri = 1.04 \times 10^{10} @ 60$
$Ri = 7.36 \times 10^9 @ 30$	$Li = 5.63 \times 10^8$	$L_1 = 5.65 \times 10^{\circ}$	$L_1 = 5.66 \ x 10^{\circ}$
$\kappa_1 = 8.61 \times 10^{-5} \ (0.60)$ Li = 5.66 \times 10^8			
	Standard Chonner and handles	Immound Changer and Land I	
	$Ri = 7.17 x 10^9 @ 30$	Ri = 5.18×10^9 @30	
	$Ri = 9.17x10^9$ @ 60	$Ri = 6.03 \times 10^9 @ 60$	
	$L_1 = 6.1 / x_10^{\circ}$	$Li = 5.37 \ x 10^8$	

PIG EYE Ce:			30 x10 ⁻¹²
ASPIRATION TUBING POLY or PVC, 1.59mm ID, L=2m	$\frac{1.28 \ x 10^{10} \#}{\text{R}L} = 6.4 \ x 10^{9} / \text{m}$	$\frac{1.01 \ x 10^9}{LL = 5 \ x 10^8 / m}$	$0.68 x 10^{-12}$ CL = 0.34 x 10^{-12}/m
PHACO NEEDLES: AMO YELLOW PHACO NEEDLE 20g MICRO-FLOW L=20mm, I.D 0.635mm	7.78 $x10^9$ @ 30 1.00 $x10^{10}$ @ 45 1.23 $x10^{10}$ @ 60 1.68 $x10^{10}$ @ 90	6.31 x10 ⁷	
Plain AMO 19g Phaco Needle L = 20mm, I.D = 0.83mm	2.88 x10 ⁹ @ 30 3.98 x10 ⁹ @ 60 5.08 x10 ⁹ @ 90	3.69 x10 ⁷	
B&L Microflow Plus needle, L= 19mm ID = 0.59mm.	1.01 $x10^{10}$ @ 30 1.52 $x10^{10}$ @ 60 2.03 $x10^{10}$ @ 90	$6.95 x 10^7$	
Alcon (2.2) pink needle. L= 30mm ID= 0.55mm	1.51 $x10^{10}$ 30 1.97 $x10^{10}$ 60 2.43 $x10^{10}$ 90	$1.26 x 10^8$	
PROBE BODY ASPIRATION SIDE L=150mm,ID=1.59mm	9.62 x10 ⁸ #	7.5 x10 ⁷	
PERISTALTIC PUMP TUBING + VACUUM SENSOR AREA, Total Cm :			$4.5 x 10^{-12}$
ASPIRATION SYSTEM TOTALS With:			
AMO Yellow needle	AMO 19g plain needle	B&L micro-flow plus needle	Alcon 2.2 coaxial pink.
$\mathbf{Ra} = 2.15 \ x 10^{10} @ \ 30$ = 2.37 \ x 10^{10} @ \ 45 = 2.60 \ x 10^{10} @ \ 60 = 3.06 \ x 10^{10} @ \ 90	$\mathbf{Ra} = 1.66 \ x \ 10^{10} \ @ \ 30$ $= 1.77 \ x \ 10^{10} \ @ \ 60$ $= 1.88 \ x \ 10^{10} \ @ \ 90$	$\mathbf{Ra} = 2.39 \ x 10^{10} \ @ \ 30$ $= 2.89 \ x 10^{10} \ @ \ 60$ $= 3.40 \ x 10^{10} \ @ \ 90$	$\mathbf{Ra} = 2.89 \ x10^{10} \ @ \ 30$ = 3.35 \ x10^{10} \ @ \ 60 = 3.80 \ x10^{10} \ @ \ 90
$La = 1.15 x 10^9$	$La = 1.12 x 10^9$	$La = 1.15 \ x 10^9$	$La = 1.21 \ x 10^9$
Rt = 2.89 $x 10^{10}$ @ 30 in eye = 3.46 $x 10^{10}$ @ 60 in eye	$\mathbf{Rt} = 2.29 \ x 10^{10} @ \ 30 \text{ in eye}$ $= 2.55 \ x 10^{10} @ \ 60 \text{ in eye}$	$\mathbf{Rt} = 3.04 \ \mathbf{x10^{10}} @ \ 30 \text{ in eye}$ $= 3.63 \ \mathbf{x10^{10}} @ \ 60 \text{ in eye}$	$\mathbf{Rt} = 3.73 \ x 10^{10} @ \ 30 \text{ in eye}$ $= 4.39 \ x 10^{10} @ \ 60 \text{ in eye}$

(7-3) COMPLIANCE of TUBING: HOW TO CALCULATE: H. Holden 2007.



1) FIRST DESCRIBE THE MECHANICAL PROPERTIES OF THE TUBING MATERIAL

CONSIDER a length of tubing material, cross section 1 unit



The force F above is the "Force per unit area" therefore the total force FT, acting on a larger section of plastic is the product of the total area and K. x.

FT = AREA. K. x. (1)

Round section of material (long tube)



With an increase in lumen pressure $P_2 > P_1$, then the pressure increase is $P_2 - P_1$ or ΔP .

R increases with applied pressure to R2 or to a value of $(R + \Delta R)$ where ΔR is an additional length to the lumen radius.

The total force applied to the lumen surface area is:

$$FT = \Delta P.2.\pi.(R + \Delta R).l \qquad (2)$$

We can now equate this with equation (1). The total elongation around the circumference increases until the elastic forces in the tube wall balance with the force from the lumen pressure. The elongation x around the wall is:

$$x = 2.\pi.(\mathbf{R} + \Delta \mathbf{R}) - 2.\pi. \mathbf{R}$$

The AREA is T.l and the spring constant of the plastic is K, therefore:

$$\Delta P.2.\pi.(R + \Delta R).l = (2.\pi.(R + \Delta R) - 2.\pi. R). K.T.l$$

Solving for ΔP :

$$\Delta P = \frac{\Delta R.K.T}{(R+\Delta R)}$$
(3)

The volume increase ΔV over the applied pressure and expansion is:

$$\Delta V = l.\pi (R + \Delta R)^2 - l.\pi R^2 \quad (4)$$

Compliance Co = $\frac{\Delta V}{\Delta P}$, therefore from (3) and (4):

$$Co = \frac{(l.\pi(R + \Delta R)^2 - l.\pi R^2)(R + \Delta R)}{\Delta R.K.T}$$

Therefore:

$$C_0 = \frac{2.l.\pi.R^2}{K.T} + \frac{3.l.\pi.R.\Delta R}{K.T} + \frac{l.\pi.(\Delta R)^2}{K.T}$$

Typical aspiration tubing used in Phaco Fluidics has a small increase in radius over a large range of pressure (500mmHg). Typically (R + Δ R) / R = 1.006 over that pressure or vacuum range, or Δ R being small at 0.6% so that the terms involving Δ R can be ignored, therefore:

$$Co = \frac{2.l.\pi.R^2}{K.T}$$
 (5) Holden Compliance equation for fluid filled plastic tubing.

This makes the compliance per unit length CL:

$$CL = \frac{2.\pi . R^2}{K.T} \qquad (6)$$

(Equation 6 is useful in computing the tubing's characteristic velocity)

Assumptions about tubing compliance: The tubing ends are ignored and it is assumed that the fluid filling the tubing for practical purposes is incompressible. The tubing behaves linearly with respect to its elasticity, and this property is the same whether or not the tubing lumen is under either vacuum or pressure with respect to atmospheric pressure.

Therefore fluid filled plastic tubing compliance is:

- # Directly proportional to the length of the tubing
- # Proportional to the square of the tubing's internal radius

Inversely proportional to the thickness of the tube wall and the stiffness constant of the plastic.

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NOTES:

The purpose of this book is to help fellow Ophthalmologists negotiate their way in the field of Phacoemulsification machine fluidics. The desired goal is safer and more efficient cataract surgery. With a better understanding of the equipment we use, we can tailor the machine settings to suit individual eyes and identify the causes for anterior chamber instability when it occurs. Also as a group we will be better able to critically judge the value of any marketing claims related to Phaco-emulsification equipment. There is the risk that if we don't keep up with the technology that it will become "black box" and we will be less able to critically analyse claims of improved performance in the machinery. Also as incremental improvements occur to our machine technology we also need to be sure they are significant enough and are cost effective to be valuable, as ultimately the machine and technology costs are handed to the patient. We owe this consideration to our patients.

Therefore this book is dedicated to my Wife Jeanette and my three daughters Bailey, Kelsey and Danielle who supported me over the time to create it and also dedicated to the thousands of cataract patients I have had the privilege to perform surgery on over the last 15 years.

This book is available at no cost as a downloadable pdf.

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ABOUT THE AUTHOR:

Dr Holden was born in New Zealand in 1958 and educated in Auckland. From a very young age he became interested in Electronics Engineering and Medicine and initially became an Electronics Engineer, specialising in Video and Television Engineering. The career change to medicine came in the early 1980's and over time, a special interest in the Physiology of human vision. This led to a career in Ophthalmology. Initially the focus was laser and refractive surgery, and Dr Holden moved to Australia to train in that sub-specialty with the first College approved Refractive Surgery Fellowship in 1997. Over time though, the focus shifted to cataract surgery in elderly patients with visually disabling cataracts. Currently Dr Holden has a busy general Ophthalmic practice in Maroochydore, Queensland.

Given the dependence of modern cataract surgery on the machinery the natural progression was a thorough understanding of the phaco machine technology. Ultimately to facilitate the design and construction of Australia's first prototype Peristaltic phaco machine. This machine was deployed in a TGA registered trial. Subsequently another type of machine, vacuum based, was also designed and built. The goal was to create simple, economical and miniature phaco machines for developing nations where resources are tight. Dr Holden also became aware that the current Ophthalmic literature was lacking in any solid scientific model to explain the fluidic events which both influence the surgical safety and determine the stability of the anterior chamber of the eye during surgery. In addition it was noted that were many phaco machine marketing claims which appeared to have little scientific merit. Therefore this book was created to put the record straight on phaco fluidics and provide a solid basic science grounding for this subject. It represents the results of 8 years of research into this topic and it has required the integration of a number of the different basic sciences to achieve because there had been no current modelling system available for the unique features of phaco fluidics. The ultimate goal being improved surgical safety for the cataract patient.