The Influence of Aneurysm Configuration on Intra-Aneurysmal Pressure and Flow

A. SORTEBERG, D. FARHOUDI

Department of Neurosurgery, Rikshospitalet, The National Hospital, 0027 Oslo, Norway

Key words: cerebral aneurysm, flow, pressure, thermodilution

Summary

Aneurysm rupture may occur in conjunction with swift pressure increases like heavy lifting. We therefore investigated the transmission of pressure waves as well as local flow rates in various types of experimental aneurysms and their parent arteries. 0.014-inch guidewires containing a combined pressure and thermistor sensor were inserted into both the dome and the parent artery of idealized silicone aneurysms mounted in a pulsatile flow phantom. Intravascular pressure and thermodilution responses to injections of room-temperature normal saline at a rate of 5cc.s⁻¹ over two seconds were recorded simultaneously at both sites. Flow was evaluated semi-quantitatively applying the thermodilution principle. We found that the propagation of pressure was attenuated at the dome of the aneurysms compared to their respective parent arteries. This difference was more distinct in side-wall aneurysms than in bifurcational aneurysms.

The attenuation of traveling pressure in the aneurysm was most effective at low systemic pressures. The intraaneurysmal flow rate was unique, always lower than in the respective parent arteries and highly dependent on the configuration of the aneurysm. We observed considerably higher flow rates in bifurcational aneurysms compared to side-wall aneurysms. Bifurcational aneurysms were hence characterized by a relatively high pressure transmission and high flow rates which may represent a stimulus for enlargement of untreated aneurysms and promote coil compaction in endovascularly treated lesions.

Introduction

The genesis, progression, and eventual rupture of cerebral aneurysms is still not entirely understood. In a clinical setting, rupture of cerebral aneurysms is often associated with situations of sudden and swift increases in pressure; i.e. heavy lifts or Valsalva manoeuvres. Once rupture has occurred, aneurysms are increasingly repaired by the endovascular route^{1,2,3}. The endovascular accessibility of an aneurysm may, however, be limited, both in terms of anatomical properties of the vascular tree or the aneurysm itself as well as in terms of hemodynamic factors with special emphasis on intraaneurysmal flow. High flow and turnover in an aneurysm may complicate the insertion and stability of microcatheters and coils, and influence the long-term perspective of aneurysm occlusion^{4,5,6}. Flow and pressure are likely to be defined by the anatomical configuration of an aneurysm and its relation to its parent artery^{5,7}. Exact knowledge of this anatomical-hemodynamic relationship could be an additional tool in identifying aneurysms where problems with endovascular treatment could be anticipated. It could also be helpful in procedural planning with regard to supplementary endovascular devices necessary for instance for balloon support or endovascular stents. More important, however, would be insights into the propagation of acute pressure waves within the vascular tree, bearing in mind their above mentioned linkage to aneurysm rupture.

Employing a vascular phantom, we studied the pressure and thermodilution responses to

defined injections of cool normal saline within geometrically different silicone aneurysms and their parent arteries at different systemic pressure levels and pulse rates. These injections of normal saline provided significant and rapid increases in pressure and in addition allowed semi-quantitative evaluation of the local flow rate using the thermodilution principle.

Methods

The present study used a custom made system comprising a molded silicone replica of the aortic arch and brachycephalic vessels (figure 1,8). Fitted to segments representing the internal carotid arteries and vertebral arteries were connections leading to a basin into which silicone molded aneurysms of different shapes could be placed and connected to the “systemic” arterial and venous circulation. The chamber into which the aneurysms were mounted at defined coupling sites could be filled and perfused with physiologic solution using a circuit that was not connected to the vascular component of the phantom (MicroVention, Inc., Aliso Viejo, CA 92656). With the aneurysm in place, the flow phantom was connected to an electromagnetic flow pump (LMI Milton Roy, Acton, MA 01720, USA) that allowed pulsatile flow with simulated heart rates varying between 40/min and 100/min. Using an adjustable clamp mounted on the main inflow connection to the phantom, the systemic pressure could be varied.

Figure 1 The flow phantom comprising a molded silicone replica of the aortic arch and brachycephalic vessels. Fitted to segments representing the internal carotid arteries and vertebral arteries were connections leading to a basin into which silicone molded aneurysms of different shapes could be placed.

The amplitude of the pulsatile excursions could be set by the liquid-level in an accumulator chamber and was kept constant throughout the experiment in every aneurysm investigated. The model was perfused with normal saline. A constant temperature of 37°C was achieved with a heater unit and was controlled by a thermostat/heater control unit built into the reservoir of the model.

A 6 French guiding catheter was positioned in the “artery” connected to the aneurysm parent artery. A 0.014” guide-wire mounted pressure-temperature sensor (PressureWire™ Sensor, RADI Medical Systems, Uppsala, Sweden, (figure 2) was pressure calibrated at the barometric room pressure and the temperature was calibrated to zero at room temperature. This pressure-temperature sensor was advanced through the guiding catheter and placed adjacent to the wall of the aneurysm on the zenith of the dome (figure 3), i.e. the most common site of aneurysm rupture. The flexible tip past the sensor assured a stable position of the sensor. After having been calibrated in an identical manner, another identical pressure-temperature sensor was placed in the parent artery approximately 3 cm proximal to the neck of the aneurysm. We chose that relatively remote position of the parent artery sensor to minimize disturbances to inflow into the aneurysms. Using a commercially available angiographic injector (Angiomat 3000, Liebel-Flarsheim/Company, Cincinnati, Ohio, USA), normal saline at room temperature was injected through the guiding catheter (tip positioned 6-8 cm proximal to the aneurysm) at a rate of 5cc.s^{-1} over two seconds. Injections of this type induce abrupt and reproducible increases in intravascular pressure⁹. By means of the principle of thermodilution, this technique also allows qualitative assessment of flow¹⁰. The observed decrease in temperature obtained by the injection of room-temperated saline into the warm phantom circulation is proportional to the volume flow-rate at the site of measurement. In the present study, the normal saline within the perfusion phantom was maintained at 37°C which then maximally could cool down to room temperature (20-22°C) during an injection. As the temperature sensor was calibrated to zero at room temperature, the temperature hence fell from approximately 17°C to anywhere between 17°C and 0°C – depending on the local volume flow. The intravascular pressures and

thermodilution curves were recorded simultaneously using the RADIANalyzer™ (RADI Medical Systems, Uppsala, Sweden). These recordings were then transferred into a computer for storage and off-line analysis using the RadiView 1.00 2xT Software (RADI Medical Systems, Uppsala, Sweden). After each injection with normal saline at room temperature the temperature in the flow model (system temperature) was allowed to return to 37°C prior to the next injection.

Injectations were performed at various combinations of pulse rates (40/min, 60/min, 80/min and 100/min) and “systemic” pressures (50 mmHg, 100 mmHg, and 140 mmHg). For every possible combination of pulse rate and pressure two injections were performed, resulting in a total of 24 injections for every aneurysm. Although the pulse rate and pressure were adjusted, flow in the parent artery was kept constant by setting the stroke-volume in the electromagnetic pump in such a manner that flow remained constant at 200 ml.min⁻¹. The Reynolds number in the parent artery was then 239. To control the correct setting of stroke – volume, flow was measured using an intra-operative flow probe that was clamped around the parent artery proximal to the aneurysm. Flow was monitored throughout the experiments using a CardioMed CM-2008 flowmonitor (CardioMed A/S, Oslo, Norway).

Six aneurysms were studied, three bifurcational aneurysms and three side-wall aneurysms. The geometrical properties of these idealized aneurysms are listed in table 1. In bifurcational aneurysms, the afferent parent arteries were 4 mm wide and 50 mm long, with the efferent parent arteries being 3.5 mm wide and 50 mm long. Constancy of the branching angle was provided by the fixed coupling site within the basin. The parent arteries of the side-wall aneurysms were straight, 4 mm wide and 100 mm long.

Data analysis

The two injections at each pulse rate/pressure level were averaged.

Pressure was analyzed with respect to comparing the pressures in the parent artery versus the pressure at the dome of the aneurysm (figure 4). The ratio of the pressure in the parent artery by the pressure in the aneurysm was calculated automatically online by the four channel RADI analyzer and denoted FFR. In case

Figure 2 The 0.014-inch pressure-temperature sensor on top of a regular match. In the left extension of the sensor unit is the flexible tip which assures a stable intravascular position of the sensor.

of identical pressures in the parent artery and at the dome of the aneurysm FFR becomes 1.0. The difference of the FFR at baseline and the FFR during the pressure increases caused by the injections of saline was denoted ΔFFR . If there are identical pressure increases in the parent artery and in the aneurysm, the ΔFFR becomes zero.

Flow was analyzed using the following thermodilution parameters (figure 5): the maximum drop in temperature caused by the saline injections (ΔT), which is proportional to the volume flow at the site of measurement (the larger ΔT is, the higher volume flow)¹⁰; the dissociation of pressure and flow (which exists in all oscillating systems and is closely linked to the impedance) denoted “shift ϵ ”, i.e. the time

Figure 3 Position of the pressure-temperature sensor at the zenith of the aneurysm dome (arrow) in a side-wall aneurysm and a bifurcational aneurysm.

gap between the observed increase in pressure and the start of decrease in temperature; the total length of temperature change was denoted dilution interval, which is comparable to the wash-out or “turn-over” at the site of measurement. For statistical analysis we used SPSS 10.0 for Windows®. Results were considered significant if $p < 0.05$.

Results

Pressure

The baseline FFR (Pressure in the parent artery divided by pressure in the aneurysm) was not significantly different from “1”, i.e. the pressure was similar at both sites. The abrupt increase in pressure created by the injections of saline was always smaller in the aneurysm than in the parent artery (figure 4,5). This difference increased the lower the systemic pressure was (ΔFFR at 50 mmHg = -0.050, $p < 0.001$, ΔFFR at 100 mmHg = -0.034, $p < 0.001$, ΔFFR at 140 mmHg = -0.027, $p < 0.001$), but was independent of the pulse rate. Furthermore, every aneurysm had a unique ΔFFR . There was a marked difference in ΔFFR between bifurcational aneurysms (median -0.028) and side-wall aneurysms (median -0.043), (Wilcoxon signed ranks test, $Z = -3.497$, $p < 0.001$), which means that the pressure increase in the dome of side-wall aneurysms was more attenuated than in bifurcational aneurysms compared to the observed pressure increases in the respective parent arteries. We could not delineate a clear relationship of the ΔFFR to the neck size, aspect ratio or volume of the aneurysms.

The maximum drop in temperature DT (local flow rate)

Local volume flow, as expressed by ΔT , was in every instance lower in the aneurysm than in the corresponding parent arteries (mean ΔT in the aneurysms 9.40 vs ΔT in the parent arteries 14.63, $t = -14.463$, $p < 0.0001$ paired samples test, figure 6). In the aneurysms, DT was highly dependent on the configuration of the aneurysm. The observed flow rate was unique for each aneurysm. There was considerably higher flow in bifurcational aneurysms compared to side-wall aneurysms ($Z = 4.195$, $p < 0.0001$, 2-tailed Wilcoxon signed ranks test, figure 6). To obtain a similar high inflow into a side-wall aneurysm, the aspect ratio had to be decreased to a degree that would obviate coiling without balloon

or stent support (side-wall aneurysm #2). Figure 7 illustrates a typical recording in a side-wall aneurysm versus a bifurcational aneurysm. None of the other geometrical aneurysm characteristics correlated significantly with ΔT , considering all aneurysms as one group. When dividing into subgroups, however, ΔT in side-wall aneurysms was highly significantly linked to the neck size ($r = 0.963$, $p < 0.0001$, Pearson's correlation) and inversely to the aspect ratio ($r = -0.960$, $p < 0.0001$, Pearson's correlation); i.e. intraaneurysmal flow increases with increasing size of the aneurysm neck. We could not find a similar relation in bifurcational aneurysms. Local flow was altered negligibly only by variations in pressure and pulse rate and was non-corresponding within the aneurysm and their parent arteries.

Pressure - flow shift ϵ

The pressure-flow shift ϵ was negative in all instances within the aneurysms, i.e. flow was leading pressure (figure 8). Furthermore, the shift ϵ was significantly higher in all parent arteries than within the aneurysms ($t = -22.697$, $p < 0.0001$, paired samples test, figure 8) reaching even positive values during some conditions (pressure leads flow). There was no significant difference in the shift ϵ between bifurcational and side-wall aneurysms. In the parent arteries of bifurcational aneurysms, however, the shift ϵ was often positive, whereas it was negative or close to zero in the parent arteries of side-wall aneurysms (0.0563 vs -0.0279, $Z = -3.065$, $p = 0.002$ Wilcoxon signed ranks test). The shift ϵ showed no specific variations with changing pulse rates or different pressure levels. In side-wall aneurysms, the shift ϵ was linked to the neck size ($r = 0.859$, $p < 0.0001$, Pearson's correlation) and inversely to the aspect ratio ($r = -0.694$, $p < 0.0001$, Pearson's correlation); i.e. the dissociation of pressure and flow increased with decreasing neck size. No statistically significant correlation with geometrical features was found in bifurcational aneurysms.

The dilution interval

The dilution interval was longer in all aneurysms than in their respective parent arteries ($p < 0.001$, t-test for paired samples, figure 9). Moreover, the length of the dilution intervals was different in the various aneurysms ($\chi^2 = 130.5$ $p < 0.001$, Kruskal-Wallis test). The dilution interval in side-wall aneurysms and their

Figure 4 Pressure recording within the aneurysm (red) and in the proximal parent artery (blue). The green line is the computed ratio of the pressure in the parent artery to the pressure in the aneurysm. The rapid increase in pressures is caused by the injection of saline at a rate of 5cc.s^{-1} over two seconds. ΔFFR is the difference of the FFR at baseline and the FFR during the power injection of saline. Note the attenuation of the pressure increase in the aneurysm.

parent arteries was longer than in bifurcational aneurysms and their parent arteries (aneurysms: $Z=6.805$ $p<0.001$; parent arteries: $Z=9.954$ $p<0.001$, Wilcoxon two sample test, figure 9). No strong links of the dilution interval and geometrical properties were found in bifurcational aneurysms. In side-wall aneurysms, the dilution interval correlated strongly with the aspect ratio ($r=0.750$, $p<0.0001$, Pearson's correlation) and inversely with the neck size ($r=-.956$, $p<0.0001$, Pearson's correlation); i.e. the smaller the neck of an aneurysm, the longer the aneurysmal turnover took. The length of the dilution interval did not vary significantly with changes in systemic pressure or pulse rate.

Discussion

The notion of aneurysmal morphology representing a major determinant of intraaneurysmal flow is not novel. Ferguson¹¹ realized the importance of hemodynamic factors in the initiation, growth and rupture of aneurysms and was the first to report turbulent flow in cere-

bral saccular aneurysms¹². Later, aneurysmal flow was described as disturbed, rather than turbulent, with the blood entering during systole at the downstream ostium, forming a counterclockwise vortex wave, and leaving the aneurysm at the upstream ostium during systole^{13,14}. Some studies focused on the size of aneurysms, which is the only geometric variable clinically applied with a critical value when deciding on prophylactic treatment of unruptured aneurysms. Steiger and Reulen¹⁵ measured flow with Laser Doppler and observed disturbed flow and vortex trails in zones of deceleration without aneurysmal size influencing the basic flow pattern. Liou et Al.¹⁶ identified a "middle range of aneurysm sizes" where the vortex is stronger and the fluctuation level is high at the dome, whereas smaller or larger aneurysms lacked the vortex inside the aneurysm model. That "middle range of aneurysm size" in their model, however, is difficult to translate into human aneurysm sizes as they applied aneurysms measuring 80×60 mm and 56×50 mm¹⁶. Ujiie et Al.¹⁴, in their rabbit model, measured velocity profiles with a 20 MHz Doppler probe and

Figure 5 Pressure (above) and temperature (below) recording during an injection of cool saline at a rate of 5cc.s^{-1} over two seconds. Simultaneous recording in the aneurysm (red) and its parent artery (blue). The injection causes an abrupt increase in pressure along with a drop in temperature (ΔT) which is proportional to the local volume flow. The difference between the observed increase in pressure and drop in temperature represents the “pressure-flow shift ϵ ”. The total time of change in temperature is denoted “dilution interval” and corresponds to the turnover. Note the smaller drop in temperature in the aneurysm as an expression for lower flow within the aneurysm than in its parent vessel.

found intraaneurysmal flow to be size dependent with a to-and-fro pattern for aneurysms smaller than 2.5 mm, circulating flow for aneurysms between 2.5 and 5 mm, and circulating flow with a superimposed rotating wave for aneurysms larger than 5 mm. The latter pattern of intraaneurysmal flow could be created in all aneurysms by narrowing the size of the neck¹⁴. All the aforementioned studies evaluated flow using Doppler velocimetry. Velocities can be transformed into regional flow under certain assumptions, as for example knowledge of the cross-sectional area investigated or the insonation angle. In the present study, we semi quantitatively evaluated volume flow applying the thermodilution technique that was introduced in 1954 by Fegler¹⁷ and nowadays is the

most widely used mode to determine cardiac output in humans. We found specific flow rates for every single aneurysm. This observation of unique aneurysm flow profiles concurs with the findings of Cebal et Al.¹⁸. However, there was no clear correlation of flow to the size of the aneurysm, either in terms of depth or volume. This finding is not in contrast with those of Ujiie et Al.¹⁴ and Liou et Al.¹⁶ because their technology (velocity measurements) was sensitive to the mode and direction of flow rather than the quantity of volume flow. Even though the decision to treat an unruptured aneurysm most often is based on some defined “critical size” it appears to be unlikely that size alone is such an important factor for intraaneurysmal hemodynamics. Kyriacou and Humphrey¹⁹ advocate the

Figure 6 The maximum drop in temperature ΔT caused by the cold saline injections in the aneurysms (above) and their parent arteries (below) at varying systemic pressures and pulse rates. The drop in temperature is proportional to the local volume flow. Black lines: bifurcational aneurysms and their parent arteries. Grey lines: Side-wall aneurysms and their parent arteries. Flow is unique for every aneurysm but quite similar in the parent arteries. Flow is higher in bifurcational aneurysms. In a side-wall aneurysm with very wide neck flow is also high. Changes in systemic pressure and pulse rate affect flow insignificantly only.

important role of aneurysm shape, vessel wall properties and loading conditions as they consider all stresses and strains in aneurysms to be multiaxial. In this sense, the strong correlation of the aspect ratio to flow we found was valid only in side-wall aneurysms; i.e. it was strongly dependent on the relation of the aneurysm to its parent artery. The increase in volume flow into an aneurysm with increasing ostium width was also described by Burleson et Al.²⁰ In bifurcational aneurysms, we found flow to be almost as high as in the parent artery. This high inflow and rapid turn-over along with a presumably distinct flow distribution within bifurcational aneurysms versus side-wall aneurysms

is likely to prevail over the effect of other geometrical properties. Concerning the flow pattern, Steiger et Al.²¹ found highly unstable flow patterns within bifurcational aneurysms, especially when the aneurysm was angled straight in relation to its parent artery as it was the case in our bifurcation aneurysms. They stated that this fluctuating flow exerts significant mechanical stress which can contribute to growth and eventual rupture of the aneurysm. This idea is further supported by the recent study of Cebal et Al.¹⁸ who found unstable flow patterns, small impingement regions, and small jet sizes in ruptured cerebral aneurysms employing their patient-specific computational hemodynamics

Figure 7 Pressure and flow registration during an injection of cool saline at a rate of 5cc.s^{-1} over two seconds in side-wall aneurysm #3 (above) and bifurcational aneurysm #1 (below). The systemic pressure is 100 mmHg and the pulse rate is 80 . min⁻¹. Red: measurements inside the aneurysm, blue: measurement inside the parent artery. The aneurysms have identical neck-sizes of 4 mm and identical aspect ratios of 1.75. Despite that, the intraaneurysmal flow is very different, with the flow in the bifurcational aneurysm being more than twice of that in the side-wall aneurysm. The aneurysmal turnover is much slower in the side-wall aneurysm, as expressed in the dilution interval. Due to the long dilution interval, the upper trace has been more compressed than the lower one (see time-axis).

model. In contrast, most unruptured aneurysms revealed simple stable flow patterns with large impingement zones and jet sizes¹⁸. Such stable predictable flow patterns are most often seen in side-wall aneurysms²². Likewise, assuming that coil compaction becomes more likely with increasing hemodynamic stress, the likelihood of incomplete coil occlusion after endovascular treatment should be larger in bifurcational aneurysms than in side-wall aneurysms (except those with an extremely wide neck). Graves et Al.²² found the force of the aneurysmal inflow to be the determining factor in both the placement and stability of coils. Gobin et Al.¹³ considered hemodynamic stress within the aneur-

ysm to be proportional to the aneurysmal filling and wash-out. This corresponds to our findings of larger drops in temperature to the cold injectate and shorter dilution intervals, i.e. larger stress in bifurcational aneurysms.

In general, flow within an aneurysm is enhanced when flow in the system becomes higher¹³. In the present study, flow was maintained constant at 200 ml.min⁻¹ though the pulse rate and systemic pressure were varied. In response to these variations we observed insignificant changes in intraaneurysmal flow only. Gobin et Al.¹³ also found aneurysmal hemodynamic stress to be independent of the frequency but dependent on the flow rate and the pulsatility of flow.

Figure 8 The dissociation of pressure and flow (shift ϵ) in the aneurysms (upper) and their parent arteries (lower) at varying systemic pressures and pulse rates. Black lines: bifurcational aneurysms and their parent arteries. Grey lines: Side-wall aneurysms and their parent arteries. The shift ϵ is always negative in the aneurysms, i.e. flow was leading pressure. In the parent arteries of bifurcational aneurysms the shift ϵ is partly positive, whereas it is negative or close to zero in the parent arteries of side-wall aneurysms. The shift ϵ shows no specific variations with changing pulse rates or different pressure levels.

The propagation of pressure and flow waves in pulsatile systems is pressure and flow dependent, amongst other factors, on the viscosity of the propagation medium, the elasticity of the vascular tree and the reflection coefficient resulting from branching. We used normal saline to perfuse the model. Newtonian liquids, like saline, will fit the Womersley equation for pressure-flow relations in pulsatile flow. Although the viscosity of normal saline is about merely one third that of blood which principally is a non-Newtonian liquid under steady flow conditions, blood would actually also behave close to Newtonian in oscillating blood flow and at a hematocrit of up to 40 per cent and at frequencies above 1 Hz²³.

In agreement with that, Cebal et Al.²⁴ analyzed the technique and specificity of image-based patient-specific analysis of aneurysm hemodynamics and concluded that Newtonian models were adequate in characterizing intraaneurysmal hemodynamics. The presently observed pressure and flow values can therefore be assumed to be quite similar to those one could have obtained if the model were perfused by blood. Sorteberg et Al.⁸ hence measured similar pressures and flow in a canine aneurysm before and after coiling when the aneurysm was perfused by blood and when it was perfused with normal saline.

Our results might also have been influenced

Figure 9 The length of the dilution interval represented by the total time of change in temperature occurring after injection of cool saline in the aneurysms (above) and their parent arteries (below) at varying systemic pressures and pulse rates. Black lines: bifurcational aneurysms and their parent arteries. Grey lines: Side-wall aneurysms and their parent arteries. The dilution interval is longer in all aneurysms than in their respective parent arteries. It is also different in the various aneurysms and parent vessels, with the dilution interval in side-wall aneurysms and their parent arteries being longer than in bifurcational aneurysms and their parent arteries, i.e. the turnover is quicker in bifurcational aneurysms and their parent arteries.

by the elasticity of the silicone model. Although silicone may represent a fair approximation of the elasticity of the arterial wall, differences in compliance can be assumed. In a pulsatile system, elasticity of the wall will always produce waves that travel retrograde and which in part may neutralize shear stresses. Unless flow is measured in vivo inside human cerebral aneurysms every model will represent a mere approximation of the true complex elasticity in the system, which in reality will also be influenced by the opposing force of intracranial pressure. According to McDonald²⁵, even though specific values along the flow velocity

curve will be erroneous, the pattern of flow will be similar to the real one over a wide range of elasticity, even in a model with rigid tubes. Image based computational fluid dynamics (CFD), for example, assumes rigid vessel walls and the reflection of waves from other vessels is not accounted for^{4,18,24}. It is unlikely that our model introduces larger errors than the CFD technology into the flow-pressure relationship due to elasticity and wave reflection.

The phenomenon of wave reflection is most probably responsible for the present finding of a positive shift in the artery carrying a bifurcational aneurysm versus a corresponding neg-

ative shift in the parent artery of side-wall aneurysms. Every branching of the vascular tree increases the total cross-sectional vascular area and thereby enhances retrograde wave reflection. In that situation, pressure was leading flow. It is also interesting that the dissociation of pressure to flow (shift ϵ) was less within all aneurysms than their parent arteries, i.e. the effects from oscillatory flow might have been less pronounced inside the aneurysms.

We observed similar pressures within the aneurysms and their parent arteries, which is in consensus with the investigations of others^{9,13,26,27,28}. Likewise, Novak et Al.²⁷ also tried to link intraaneurysmal pressures to the size of the aneurysm. Burleson et Al.²⁰, in a computer modeling study, showed that there is a maximum pressure area at the distal extent of the ostium and that the distribution of pressure varied with the configuration of the aneurysm. We measured pressure at the dome of the aneurysm, where Burleson et Al.²⁰ found the lowest pressures. Still, our pressures were of the same magnitude as in the parent artery. Assuming the calculations of Burleson et Al.²⁰ are correct, pressure at the downstream ostium should then exceed parent artery pressures. Unlike others, we obtained direct, simultaneous measurements of instantaneous pressures in the parent vessel and the aneurysm "non-invasively" i.e. without puncturing the vessel wall and thereby possibly introducing confounding factors. Furthermore, this is, to our knowledge, the only study focusing on sudden significant pressure increases (created by the saline injections). In other experiments, the systemic pressure was manipulated pharmacologically and was hence regulated much more slowly²⁸. Very swift pressure increases might be critical, as many patients report their aneurysm bleed to have occurred in situations of transiently elevated pressure (heavy lifts, abdominal press). These abrupt pressure increases were not transmitted with full strength into the aneurysm. The attenuation of that enormous pressure wave was most effective when systemic pressure was low and/or the aneurysmal turnover was low. In aneurysms, the pressure-volume curve is N-shaped, i.e. there is a steep linear increase in volume with increasing pressure until a pressure threshold is reached where small pressure elevations cause a disproportional high increase in volume ("jump in volume")²⁹. Parallel to this, one could expect

the transmission of pressure to follow the same pattern, with high attenuation in the left, steep part of the pressure-volume curve and virtually lacking attenuation close to the pressure threshold. We also found the attenuation of pressure waves to be less in bifurcational versus side-wall aneurysms. Bearing in mind the pressure-volume curve, bifurcational aneurysms hence might carry a higher risk of enlargement than side-wall aneurysms. According to the International Study of Unruptured Intracranial Aneurysms, ISUIA³⁰, larger aneurysms are more likely to rupture. This study³⁰ also found an increased risk for rupture in basilar tip and posterior communicating artery aneurysms which resemble the configuration of our bifurcational aneurysms. The strong transmission of pressure waves at bifurcations could also be a contributing factor to compaction in coiled aneurysms. In a clinical setting, lower blood pressures could contribute to prevent aneurysm rupture or reduce the risk of re-rupture of aneurysms. However, in acutely ruptured aneurysms where the intracranial pressure most often is increased, systemic blood pressure is often allowed to remain elevated in order to secure adequate cerebral perfusion. In that situation, lowering the arterial blood pressure could lead to inadequate cerebral perfusion pressure unless the raised intracranial pressure is treated with drainage of cerebrospinal fluid.

Although the transmission of pressure and flow through unruptured as well as treated aneurysms is complex, our findings suggest particular caution when handling bifurcational aneurysms.

Conclusions

Acute pressure waves and the flow rate are attenuated in aneurysms compared to their parent arteries. The attenuation of traveling pressure in the aneurysm was most effective at low systemic pressures. Bifurcational aneurysms are characterized by less dampening of abrupt pressure increases and considerably higher flow rates than side-wall aneurysms. In side-wall aneurysms intraaneurysmal flow is proportional to the size of the neck. The high stress in bifurcational aneurysms may represent a stimulus for enlargement of untreated aneurysms and promote coil compaction in endovascularly treated lesions

References

- 1 Brilstra EH, Rinkel GJE et Al: Treatment of intracranial aneurysms by embolization with coils: a systematic review. *Stroke* 30: 470-476, 1999.
- 2 Graves VB, Strother CM et Al: Early treatment of ruptured aneurysms with Guglielmi detachable coils: effect on subsequent bleeding. *Neurosurgery* 37: 640-648, 1995.
- 3 Molyneux A, Kerr R et Al: International Subarachnoid Aneurysm Trial (ISAT) Collaborative Group: International subarachnoid aneurysm trial (ISAT) of neurosurgical clipping versus endovascular coiling in 2145 patients with ruptured intracranial aneurysms: a randomized trial. *The Lancet* 360: 1267-1274, 2002.
- 4 Byun HS, Rhee K: CFD modelling of blood flow following coil embolization of aneurysms. *Medical Engineering & Physics* 26: 755-761, 2004.
- 5 Kiyosue H, Tanoue S et Al: Anatomic features predictive of complete aneurysm occlusion can be determined with three-dimensional digital subtraction angiography. *Am J Neuroradiol* 23: 1206-1213, 2002.
- 6 Zubillaga AF, Guglielmi G et Al: Endovascular occlusion of intracranial aneurysms with electrically detachable coils: correlation of aneurysm neck size and treatment results. *Am J Neuroradiol* 15: 815-820, 1994.
- 7 Barrocas AM, Derdeyn C et Al: Histologic and hemodynamic effects of endosaccular platinum coils for intracranial aneurysms. *J Long-term Eff Med Impl* 14(3): 225-242, 2004.
- 8 Sorteberg A, Sorteberg W et Al: Hemodynamic versus hydrodynamic effects of Guglielmi detachable coils on intraaneurysmal pressure and flow at varying pulse rate and systemic pressure. *Am J Neuroradiol* 25: 1049-1057, 2004.
- 9 Sorteberg A, Sorteberg W et Al: Effect of Guglielmi detachable coil placement on intraaneurysmal pressure: experimental study in canines. *Am J Neuroradiol* 22: 1750-1756, 2001.
- 10 Sorteberg A, Sorteberg W et Al: Effect of Guglielmi detachable coils on intraaneurysmal flow: experimental study in canines. *Am J Neuroradiol* 23: 288-294, 2002.
- 11 Ferguson GG: Physical factors in the initiation, growth, and rupture of human intracranial saccular aneurysm. *J Neurosurg* 37: 666-677, 1972.
- 12 Ferguson GG: Turbulence in human intracranial saccular aneurysms. *J Neurosurg* 33: 485-497, 1970.
- 13 Gobin YP, Counord JL et Al: In vitro study of haemodynamics in a giant saccular aneurysm model: influence of flow dynamics in the parent vessel and effects of coil embolisation. *Neuroradiology* 36: 530-536, 1994.
- 14 Ujiie H, Tachibana H et Al: Effects of size and shape (aspect ratio) on the hemodynamics of saccular aneurysms: a possible index for surgical treatment of intracranial aneurysms. *Neurosurgery* 45: 119-129, 1999.
- 15 Steiger HJ, Reulen HJ: Low frequency flow fluctuations in saccular aneurysms. *Acta Neurochir* 83: 131-137, 1986.
- 16 Liou TM, Chang WC, Liao CC: Experimental study of steady and pulsatile flows in cerebral aneurysm model of various sizes at branching site. *ASME J Biomech Engineering* 119: 325-332, 1997.
- 17 Fegler G: Measurement of cardiac output in anesthetized animals by a thermodilution method. *Quart J Exp Physiol* 39: 153-164, 1954.
- 18 Cebal JR, Castro MA et Al: Characterization of cerebral aneurysms for assessing risk of rupture by using patient-specific computational hemodynamics models. *Am J Neuroradiol* 26: 2550-2559, 2005.
- 19 Kyriacou SK, Humphrey JD: Influence of size, shape and properties on the mechanics of axisymmetric saccular aneurysms. *J Biomechanics* 29(8): 1015-1022, 1996.
- 20 Burleson AC, Strother CM, Turitto VT: Computer modelling of intracranial saccular and lateral aneurysms for the study of their hemodynamics. *Neurosurgery* 37(4): 774-784, 1995.
- 21 Steiger HJ, Liepsch DW et Al: Hemodynamic stress in terminal saccular aneurysms: A laser-Doppler study. *Heart Vessels* 4: 162-169, 1988.
- 22 Graves VB, Strother CM et Al: Flow dynamics of lateral carotid artery aneurysms and their effects on coils and balloons: An experimental study in dogs. *Am J Neuroradiol* 13: 189-196, 1992.
- 23 Mc Donald DA: The viscous properties of blood in Blood flow in arteries. London, England: 2nd ed., Edward Arnold, 1974, pp 55-70.
- 24 Cebal JR, Castro MA et Al: Efficient pipeline for image-based patient-specific analysis of cerebral aneurysm hemodynamics: technique and sensitivity. *IEEE Transactions on medical imaging* 24(4): 457-467, 2005.
- 25 Mc Donald DA: Pulsatile flow in an elastic tube in Blood flow in arteries. London, England: 2nd ed., Edward Arnold, 1974, pp 283-308.
- 26 Boecher-Schwarz HG, Ringel K et Al: Ex vivo study of the physical effect of coils on pressure and flow dynamics in experimental aneurysms. *Am J Neuroradiol* 21: 1532-1536, 2000.
- 27 Novak P, Gilkstein R, Mohr G: Pulsation-pressure relationship in experimental aneurysms: observation of aneurysmal hysteresis. *Neurol Res* 18: 377-382, 1996.
- 28 Sekhar LN, Scialassi RJ et Al: Intra-aneurysmal pressure measurements in experimental saccular aneurysms in dogs. *Stroke* 19: 352-356, 1988.
- 29 Austin GM, Schievink W, Williams R: Controlled pressure-volume factors in the enlargement of intracranial aneurysms. *Neurosurgery* 24(5): 722-730, 1989.
- 30 Wiebers DO, Whisnant JP et Al: International Study of Unruptured Intracranial Aneurysms Investigators. Unruptured intracranial aneurysms: natural history, clinical outcome, and risks of surgical and endovascular treatment. *Lancet* 362: 103-110, 2003.

Angelika Sorteberg MD, PhD
 Dept. of Neurosurgery, Rikshospitalet,
 The National Hospital,
 0027 Oslo, Norway
 E-mail: angelika.sorteberg@rikshospitalet.no