

Arterial Stenting: From Hemodynamics and Wall Mechanics to Restenosis - Experimental in vivo and Theoretical Approaches

Frank Nicoud¹ - H el ene Vernhet-Kovacsik² - Antonia P erez-Martin² - Michel Dauzat²
- Ramiro Moreno³

¹ CNRS UMR5149 I3M, Universit  Montpellier II, France

² EA 2992 - Laboratoire de Physiologie Cardiovasculaire, U.F.R. de M decine de Montpellier-N mes, Site de N mes, France

³ Radiology department, CHU of Rangueil, Toulouse, France

To date, angioplasty is the first choice treatment for arterial stenosis and short occlusion. Stenting improves both initial and long term results of angioplasty but does not avoid restenosis [1,2]. The native artery diameter appears to be a major factor of restenosis whose rate is dramatically higher in small diameter arteries. For instance, the actual estimated rate of restenosis after deployment of metallic stents in coronary arteries is about 30% [1-3]. Impressive results in lowering in-stent restenosis rate have been gained with drug-eluting stents during the last 2 years [4, 5]. However, a residual rate of restenosis has been reported, with predominant location at the stent edges [6]. The respective role of stent induced hemodynamical changes, decreased drug efficiency at the stent edges, and wall injury during stent impaction, remains to be assessed.

The so-called *compliance mismatch* between the native artery and synthetic vascular substitutes is often reported as the main possible contributing mechanical factor of graft failure, especially in small caliber arteries [7-11]. Abrupt compliance change and concomitant non-laminar flow patterns may contribute to the early accumulation of thrombus and later neointimal thickening [12, 13]. Studying the intimal response to stenting in animal models, Barth et al. [14] and Sutton et al.[15] found a larger neointima thickness at the distal extremity of the stent than at its proximal or middle parts. Therefore, the predominantly distal location of intimal hyperplasia after stenting could be explained by a mechanical theory involving the compliance mismatch. In order to evaluate this hypothesis, it would be essential to assess the presence and degree of compliance mismatch between the native and stented parts of small caliber arteries. This paper reports some experimental and numerical studies developed

recently to gain insight about the relationships between stenting, mechanical / hemodynamical changes and restenosis. Experimental *in vivo* results are presented first.

Does arterial stenting induce compliance-mismatch?

Vernhet et al performed an *in vivo* experimental study [16] to evaluate changes in mechanics after endovascular stenting in small diameters arteries. Using a surgical right femoral approach, self-expanding stents (Wallstent®, Schneider, Switzerland) were placed in the infra-renal aorta of five New-Zealand white rabbits. Blood pressure changes (ΔP) were monitored in the aorta. Blood flow velocity was measured with a 20MHz pulsed Doppler probe and the Pulsatility Index (PI) was calculated. Aortic diameter and diameter changes were measured with a 20MHz probe in echo-tracking mode. This study demonstrated that arterial wall compliance and distensibility decreased at the stent level, and were then significantly lower than upstream and downstream. There was no significant change in PI. Therefore, endovascular stenting produces a significant decrease in arterial wall compliance and distensibility of the rabbit abdominal aorta.

Are there differences between stents?

The number of stent designs available on the market has dramatically increased during the last years in order both to improve feasibility and to decrease immediate and late restenosis. Presently, stents are characterized by their components, size and geometry, and their method of implantation. *In vitro* evaluation of the mechanical characteristics of stents has been extensively reported [10-13]. In 1999, Rolland et al [17] published and described impressive results regarding hemodynamics and arterial wall mechanics in large diameter arteries after stenting (swine iliac arteries). They demonstrated marked but varying, or even opposite, changes depending on the stent design. Vernhet et al. [18] performed a study in small caliber arteries (with a diameter in the range of human coronary arteries) to compare three different stent designs: the balloon expandable, rigid Palmaz stent (Johnson and Johnson, Warren, NJ), the self-expandable, flexible Wallstent (Schneider, Bülach, Switzerland), and the balloon expandable covered Jostent (Jomed, Rangendingen, Germany), using the same experimental method [16]. Whatever the stent design, there was a marked compliance mismatch between the stented and non stented aorta. Unlike Rolland et al. results, no significant diameter compliance or distensibility coefficient difference was found between the 3 tested types of stents at all the studied levels. There are, indeed, marked differences between Rolland et al.

and Vernhet et al. studies. Vernhet et al. investigated much smaller arteries than Rolland et al., and used a different technique, with a higher ultrasound frequency (20 instead of 10 MHz), ensuring a 0.01mm spatial resolution, and echo-tracking instead of ultrasound time of flight, avoiding any mechanical constraint on the artery, whereas Rolland et al. used a silicone clip supporting two transducers positioned face to face around the vessel.

Validation of the non invasive evaluation of compliance and distensibility.

The evaluation of arterial wall mechanical characteristics is based on the measurement of diameter and systolo-diastolic diameter changes. This is most commonly achieved with high-frequency ultrasound and the wall-tracking technique to measure the displacement of the near and far arterial walls during the cardiac cycle [19]. Alternatively, the ultrasonic transit-time method (the so-called sonomicrometry method) is used mainly in animal experiments, because it requires placing ultrasonic transducers on each side of the artery [17]. Moreover, the surgical approach itself induces acute and chronic changes in wall mechanics as reported in the literature [20]. The surgical approach is responsible for post-operative fibrosis, so that repeated measurements may be difficult, even in experimental studies on animals. Therefore, some authors rely on non invasive trans-parietal measurement with B-mode ultrasound, although this technique yields a lower spatial resolution. B-mode ultrasound appears a promising technique for the evaluation of endovascular stenting which, unlike vascular surgery, does not require surgical exposure of the vessel. Vernhet et al. validated a new computerized system processing real-time B-mode sonographic sequences for the automatic measurement of vessel diameter and diameter changes over time, and the calculation of arterial compliance and distensibility [21]. Therefore, longitudinal studies or repeated measurements in the same animal can be carried on.

Impact of high pressure inflation during stenting

The role of the luminal diameter obtained immediately after stenting in the occurrence of restenosis after coronary stenting has been emphasized [22]. On the other hand, some authors have postulated that stent overdilatation causes deep vascular wall injuries and actually stimulates intimal hyperplasia and restenosis [23]. As changes in wall shear stress induce endothelial dysfunction [24], it seems important to investigate the mechanical factors that could trigger such changes.

Vernhet et al. evaluated the wall mechanics changes induced by stent overdilatation in a model of small caliber artery: the infra-renal rabbit aorta [25]. A significant decrease of diameter compliance, distensibility coefficient, and aortic diameter was found between the native artery and the aorta downstream from the stent after stent overdilatation. Spasm caused by overdilatation of the artery may be responsible for this decrease in diameter downstream from the stent. However, spasms are common but transitory effects of overdilatation in human clinical practice. The hypothesis of a hemodynamic effect related to stent overdilatation is questionable. By worsening the compliance mismatch, stent overdilatation may enhance the reflection of the pulse wave upstream from the stent, and result in a decrease in pulsatility downstream from the stent, but this mechanism is probably not predominant, since the stent length is relatively small compared to the arterial pulse wavelength. Nevertheless, the main cause of compliance mismatch is the dramatic decrease of compliance found at the stent level itself. Therefore, the slight and maybe transitory decrease in compliance observed downstream from the stent could not void or even reduce the compliance mismatch.

Long term effect of stent placement on wall mechanics

By impairing arterial wall mechanics, stents are responsible for disturbed flow. The biomechanical hypothesis is that flow disturbances are associated with changes in wall shear stress and endothelial function stimulation, ultimately leading to intimal hyperplasia. Vernhet et al. evaluated long term wall mechanics changes and related histo-pathologic changes induced by stent placement in small diameter arteries [26]. They found that endovascular stenting of the rabbit aorta impairs wall mechanics and that performing 30% stent overdilatation does not worsen this impairment but induces greater intra-stent intimal hyperplasia. A marked decrease in wall distensibility was observed immediately after stent deployment at the stented arterial segment and was a chronic, long-lasting phenomenon. Long term mechanical adaptative changes of the host artery were found upstream from the stent. The increase in distensibility upstream from the stent could result from flow stagnation induced by compliance mismatch at the junction between the host artery and the extremity of the stented artery.

Correlation between compliance mismatch and intimal hyperplasia

Wall mechanics changes may not be sufficient to explain a greater intimal thickening after stent oversizing (overdilatation) since no additional decrease in compliance and distensibility was found [26]. Schwartz et al argued that the degree of direct vascular injury could be responsible for the magnitude of the increase in intimal thickness, because the amount of intimal hyperplasia was higher when the injury score was higher [23]. Another hypothesis is that higher inflation pressure causes markedly larger surface contact areas and contact stresses between endothelium and stent-struts, that greatly stimulates the endothelial function [27].

Effect of compliance mismatch on hemodynamics, shear stress and endothelial dysfunction

Due to the lower shear rate observed between struts and the presence of stagnation zones between stent struts [27, 28], the particles residence times would be higher in the stagnation zone, stimulating endothelial function. Proliferation of arterial smooth cells in the media and neointima may be an adaptative remodeling response to compliance mismatch, leading to restenosis at the stent extremities.

Stenting was reported to induce shear stress changes in *in vitro* as well as *in vivo* studies. Moreover, Wentzel [29], in a clinical study with intra-vascular sonography, have demonstrated that the thickness of intimal hyperplasia inversely correlates with the intensity of shear stress. Carrier et al. [30] demonstrated that higher shear stress induced decreased intimal hyperplasia, may be through reduction of inflammation.

Does compliance mismatch exist in stented human pathological arteries?

Compliance mismatch has been previously demonstrated in atheromatous animal models [31]. Human atheroma differs from the minimally calcified atheroma induced by diet and intima abrasion in New Zealand rabbits. Subsequent differences in wall mechanics may be expected. Vernhet et al. evaluated 15 patients 3 months or more after placement of a Carotid Wallstent® (Schneider, Switzerland) in the extracranial carotid artery for the treatment of significant carotid stenosis [32]. B-mode ultrasound examination was performed with a 7.5MHz probe on the stented carotid artery and on the contra-lateral internal and common carotid arteries. Carotid diameter and systolic diameter changes were measured using a dedicated image processing system (IÔ V3.1, IÔDP, Paris), while pulse blood pressure was measured. The evaluation could be completed in 8/15 patients. Compliance was significantly lower at the proximal, mid and distal stent level, than upstream, downstream or on the contra-lateral internal and common carotid artery. Stenting pathological human carotid arteries induced a

compliance mismatch between the native carotid artery and the stented segment. Although the underlying diseases (atherosclerosis, radiotherapy, dysplasia) may, to some extent, reduce arterial wall compliance and distensibility [33], the dramatic additional decrease in compliance and distensibility after stenting always resulted in a marked compliance mismatch.

Does stenting produce significant pressure wave reflection?

The general one-dimensional equations describing the pulsatile blood flow (mass and momentum conservation) in compliant arteries are well known since the 70's [34]. Although these equations are non-linear, it has been shown that a better overall accuracy is reached by using a linear formulation. The main reason for this unexpected result is that the linear equations can be solved in the Fourier space so that the viscoelastic wall behaviour can be accounted for easily by letting compliance be dependant on the frequency of the velocity/pressure fluctuations [35]. Considering a sector whose diameter and compliance do not depend on the axial position, the classical wave (Helmoltz) equation can then be derived, showing that the pressure fluctuations result from the superposition of two waves travelling in opposite directions. In the purpose of modelling the wave reflection induced by an endovascular stent placed in an elastic artery, three successive homogeneous segments can be considered [36], each having its own set of constant area and compliance. A convenient formula to assess the amount of wave reflection induced by the stent (and associated compliance mismatch) can then be derived analytically. The modulus of the reflection coefficient is found to be approximately equal to

$$|R| \approx \sqrt{\pi} f \sqrt{\rho} \frac{R_a^2 C_a - R_s^2 C_s}{R_a \sqrt{C_a}} \frac{L_s}{R_s^2},$$

where f is the frequency of the pulsation, ρ is the blood density, C is the compliance, R is the radius of the (stented) artery and L_s is the length of the stent. Indices “s” and “a” stand for the stented and the non-stented artery respectively. From this relation, the amount of wave reflection is proportional to the geometrical / mechanical mismatch $(R_a^2 C_a - R_s^2 C_s) / R_a R_s \sqrt{C_a}$ induced by the stenting, the stent shape factor L_s / R_s as well as the flow conditions. It increases with both the frequency of the wave and the length of the stent. Moreover, a stent satisfying the relation $R_s = R_a \sqrt{C_a / C_s}$ would produce no wave reflection. Clearly, such a prosthesis is not realistic since experimental measurements have

shown that C_a / C_s is as high as 5. However, since the compliance of the stented artery C_s is always smaller than the compliance C_a of the host artery, it follows from this simple model that overdilation ($R_s > R_a$) tends to reduce the amount of reflected waves. Using typical values for the geometrical / mechanical characteristics of the artery and the stent as well as for the pulsation frequency and blood density, the above relation for the reflection coefficient shows that only a small part (a few percents) of the incident pressure waves is reflected by the stent. From a physical point of view, this result is related to the fact that the stent-to-wavelength ratio is always small in clinical applications, meaning that endovascular prosthesis, although less compliant than the host artery, cannot modify the wave propagation drastically.

Does stenting modify wall shear stress (WSS) significantly?

Measuring the space-time evolutions of the wall shear stress under *in vivo* conditions is very challenging due to technological difficulties. Using numerical techniques to obtain quantitative data regarding the hemodynamic changes after stenting thus appears as a natural option. In these approaches, the Navier-Stokes equations which describe the evolutions of the blood velocity and pressure fields are discretized over space and solved by using appropriate finite elements based algorithms. The changes in geometry over the cardiac cycle are usually accounted for thanks to an Arbitrary Lagrangian Eulerian formulation while a projection method is most often used to impose the divergence free constraint on the velocity field (a direct consequence of the blood incompressibility) [37].

Two types of stent modelling have been considered so far. Most studies have addressed the effects of stenting at a micro-scale level, including the geometrical description of the struts shape and spacing [38,39]. These studies provide information relevant to the hemodynamic changes immediately after stenting, before the wires have been integrated with the surrounding tissue. Noticeably, they suggest that the flow may reattach downstream of the wires when the strut spacing is greater than about six wire diameters. On the contrary, other studies have dealt with the global effects of the compliance mismatch, neglecting the details of the prosthesis structure. The prosthesis is then modelled as a uniform elastic tube with its own compliance. The results of such studies are more relevant to long-standing stenting, after the wires have been integrated with the surrounding tissue. The reason for considering this point of view is that intimal hyperplasia is certainly a long lasting phenomenon compared to

the integration of the stent wires by the vessel: for example, a nearly continuous endothelial and pseudo-endothelial cell layer can be observed on the luminal surface of the stented rabbit aorta only one week after stenting while intimal hyperplasia is only completed after six 6 to 12 weeks for this model [40]. However the wall motion must be accounted for when the stent is seen as a compliant element, which makes the numerical analysis more difficult.

The motion of the (stented) vessel boundary results from the coupling between the fluid and wall mechanics and the local radius is mostly related to the pressure field. Such a coupling is difficult to handle since the density of blood and tissues are of the same order and because the rheology of the vessels is far from well understood. Under the linear elasticity assumption, a perturbative approach can be used [41] in order to replace the coupled fluid-vessel problem by a cascade of two simpler weakly coupled problems. The first problem provides the exact solution into a rigid vessel, the second one approximates the blood flow modifications due to the compliant wall. If we are mostly interested in the response of the fluid mechanics to wall motion perturbations, the fluid-wall coupling problem can be avoided by prescribing the wall motion *a priori*. Indeed, in the case where the stent does not introduce large pressure wave reflections and under the linear elastic assumption, the wall motion is mostly given by a single propagative wave and can be described analytically [42]. Under typical flow and mechanical conditions obtained from animal experiments, the numerical results show that the amplitude of the WSS variation over the cardiac cycle is alternatively increased (by 35 % at the middle of the stent, by 50 % in the proximal transition region) and decreased (by 20 % in the distal transition region) by long-standing stenting. This supports the idea that stenting can induce endothelial dysfunction via hemodynamic perturbations. Numerical simulations performed for over-dilated stents also indicate that the negative effects of long-standing stenting on hemodynamics decrease when the radius of the prosthesis is slightly increased [42,43].

Perspectives

Experimental *in vivo* studies of the relationships between compliance mismatch and shear stress changes along a stented artery are under progress. Because the measurements uncertainties are large due to technological difficulties, systematic comparisons between measurements and mathematical / numerical models are warranted in order to draw definite conclusions.

Despite new pharmacological progress such as drug-eluting stents, the restenosis rate did not reach zero, especially in peripheral arteries, for example femoral arteries, in selected populations such as diabetics, and in very small calibre arteries. Optimizing stent design is still warranted and should be based upon an improved comprehension of stent-artery interactions. Low-order physically meaningful numerical / analytical models for stented arteries would be essential in the development of appropriate optimization loop in order to decrease the number of *in vivo* testing. Such model should incorporate both small-scale (at the struts level) and large-scale (e.g.: global equivalent compliance) information about the prosthesis.

References

1. Fischman DL, Leon MB, Baim DS, et al. A randomized comparison of coronary stent placement and balloon angioplasty in the treatment of coronary artery disease. Stent Restenosis Study Investigators. **New Engl J Med** 1994; 331:496-501
2. Serruys PW, de Jaegere P, Kiemeneij F, et al. A comparison of balloon-expandable-stent implantation with balloon angioplasty in patients with coronary artery disease. **N Engl J Med** 1994; 331:489-495
3. Rau T, Schofer J, Schluter M, Seidensticker A, Berger J, Mathey DG. Stenting of non acute total coronary occlusions: predictors of late angiographic outcome. **J Am Coll Cardiol** 1998; 31:275-280
4. Park SJ, Shim WH, Ho DS, Raizner AE, et al. A paclitaxel-eluting stent for the prevention of coronary restenosis. **N Engl J Med** 2003; 16:1537-1545
5. Morice MC, Serruys PW, Sousa JE, et al. A randomized comparison of a sirolimus-eluting stent with a standard stent for coronary revascularization. **N Engl J Med** 2002; 346:1773-1780
6. Lemos PA, Saia F, Ligthart J, et al. Coronary restenosis after sirolimus-eluting stent implantation. Morphological description and mechanistic analysis from a consecutive series of cases. **Circulation** 2003; 108 :257-260
7. Abbott WM, Megerman J, Hasson JE, L'Italien G, Warnock DF. Effect of compliance mismatch on vascular graft patency. **J Vasc Surg** 1987; 5:376-382

8. Stewart SFC, Lyman DJ. Effects of a vascular graft/natural artery compliance mismatch on pulsatile flow. **J Biomechanics** 1992; 25:297-310
9. LoGerfo FW. Biomechanics of prosthetic arterial graft failure. **J Vasc Surg** 1989; 5:581-583
10. Kinley CE, Marble AE. Compliance: a continuing problem with vascular grafts. **J Cardiovasc Surg** 1980; 21:163-170
11. Baird RN, Abbott WM. Pulsatile blood flow in arterial grafts. **Lancet** 1976; 2:948-949
12. Berry JL, Manoach E, Mekkaoui C, et al. Hemodynamics and wall mechanics of a compliance matching stent: in vitro and in vivo analysis. **J Vasc Interv Radiol.** 2002;13:97-105
13. Berry JL, Santamarina A, Moore JE Jr, et al. Experimental and computational flow evaluation of coronary stents. **Ann Biomed Eng.** 2000; 28:386-398
14. Sutton CS, Tominaga R, Harasaki H, et al. Vascular stenting in normal and atherosclerotic rabbits: studies of the intravascular endoprosthesis of titanium-nickel-alloy. **Circulation** 1990; 81:667-683
15. Barth KH, Virmani R, Strecker EP, et al. Flexible tantalum stents implanted in aortas and iliac arteries: effects in normal canines. **Radiology** 1990; 175:91-96
16. Vernhet H, Juan JM, Demaria R, Lauraire MC, Senac JP, Dautzat M. Acute changes in aortic wall mechanical properties after stent placement in rabbits. **J Vasc Interv Radiol** 2000; 11:634-638
17. Rolland PH, Charifi AB, Verrier C, et al.. Hemodynamics and wall mechanics after stent placement in swine iliac arteries: comparative results from six stent designs. **Radiology** 1999; 213:229-246
18. Vernhet H, Demaria R, Juan JM, Oliva-Lauraire MC, Senac JP, Dautzat M. Changes in wall mechanics after endovascular stenting in the rabbit aorta: comparison of three stent designs. **AJR** 2001;176(3):803-7
19. Hoeks APG, Ruissen CJ, Hick P, et al. Transcutaneous detection of relative changes in artery diameter. **Ultrasound Med Biol** 1985; 11, 51-59
20. Megerman J, Hasson JE, Warnock DF, L'Italien GJ, Abbott W.M. Noninvasive measurements of nonlinear arterial elasticity. **Am J Physiol**, 1986; 250, 181-188
21. Vernhet H, Demaria R, Juan JM, Oliva-Lauraire MC, Quéré I, Gariépy J, Sénac JP, Dautzat M. Validation of a newly developed B-mode image processing technique

- versus wall tracking ultrasound for the study of wall mechanics in small caliber arteries. *Clin Physiol Funct Imaging* 2002; 22:180-6
22. Piamsomboon C, Roubin GS, Liu MW, et al. Relationship between oversizing of self-expanding stents and late loss index in carotid stenting. **Cath and Cardiovasc Diag** 1998; 45:139-143
 23. Schwartz RS, Huber KC, Murphy JG, et al. Restenosis and the proportional neointimal response to coronary artery injury: result in a porcine model. **J Am Coll Cardiol** 1992; 19:267-274
 24. Friedman MH, Hutchins GM, Barger CB, Deters OJ, Mark FF. Correlation between intimal thickness and fluid shear in human arteries. **Atherosclerosis** 1981; 39:425-436
 25. Vernhet H, Demaria R, Juan JM, Oliva-Lauraire MC, Senac JP, Dauzat M. Arterial stenting and overdilation: does it change wall mechanics in small calibre arteries? *J Endovasc Ther* 2002; 9:855-862
 26. Vernhet H, Demaria R, Juan JM, Oliva-Lauraire MC, Marty-Double C, Sénac JP, Dauzat M. Wall mechanics of the stented rabbit aorta: Long-term study and correlation with histologic findings. **Journal of Endovascular Therapeutics** 2003 ; 42 :389-394
 27. Rogers C, Tseng DY, Squire JC, et al. Balloon-artery interactions during stent placement: a finite element analysis approach to pressure, compliance, and stent design as contributors to vascular injury. **Circ Res.** 1999;84:378-383
 28. Berry JL, Manoach E, Mekkaoui C, et al. Hemodynamics and wall mechanics of a compliance matching stent: in vitro and in vivo analysis. **J Vasc Interv Radiol.** 2002;13:97-105
 29. Wentzel JJ, Krams R, Schuurbiens JC, et al. Relationship between neointimal thickness and shear stress after Wallstent implantation in human coronary arteries. **Circulation** 2001;103:1740-1705
 30. Carlier SG, van damme LCA, Blommerde CP, et al. Augmentation of wall shear stress inhibits neointimal hyperplasia after stent implantation. Inhibition through reduction of inflammation? **Circulation** 2003; 107:2741-2746
 31. Schwarzacher SP, Tsao PS, Ward M, Hayase M, Nienbauer J, Cooke JP, Yeung AC. Effects of stenting on adjacent vascular distensibility and neointima formation: role of nitric oxide. **Vascular Medicine** 2001;6:139-144
 32. Vernhet H, Jean B, Lust S, Laroche JP, Bonafe A, Quere I, Senac JP, Dauzat M. Wall mechanics of the stented extra-cranial artery. **Stroke** 2003: 34:222-224

33. Giannattasio C, Failla M, Emanuelli G, Grappiolo A, Boffi L, Corsi D, Mancina G. Local effects of atherosclerotic plaque on arterial distensibility. **Hypertension** 2001;38:1177-1180
34. Hughes, T. and Lubliner, J., On the 1D theory of blood flow in the larger vessels, **Math. Biosciences**, 1973, 18: 161-170
35. Reuderink, PJ., Hoogstraten, HW., Sipkema, P., Hillen, B. and Westerhof, N. , Linear and Nonlinear one-dimensional models of pulse wave transmission at high Womersley numbers, **J. Biomechanics.**, 1989, 22: 819-827
36. Nicoud, F., Vernhet, H. and Dautzat, M., Changes in fluid mechanics after endovascular stenting: the pressure waves point of view, **ESAIM Proceedings**, 2002, 12:134-139
37. Chorin, J.A., A numerical method for solving incompressible viscous flow problems, **J. Comp. Physics**, 1967,2: 12-26
38. Berry, JL., Santamarina, A., Moore, JE., Roychowdhury, S. and Routh, WD., Experimental and computational flow evaluation of coronary stents, **Annals of Biomedical Engineering**, 2000: 28: 386-398
39. Bénard, N., Coisne, D. and Perrault, R., Simulation of blood flow in stented artery: Blood rheological properties effects, **XXVIII Congrès de la Société de Biomécanique**, 2003
40. Robinson, KA., Roubin, GS., Siegel,RJ., Black,AJ., Apkarian,RP. and King, SB., Intra-arterial stenting in the atherosclerotic rabbit, **Circulation**, 1988, 78: 646-653
41. Tortoriello, A. and Pedrizzetti, G., Flow-tissue interaction with compliance mismatch in a model stented artery, **J. Biomechanics**, 2004, 37:1-11
42. Nicoud, F., Vernhet, H. and Dautzat, M., A Numerical Assessment of Wall Shear Stress Changes after Endovascular Stenting, **J. Biomechanics**, 2005, in press
43. Nicoud, F., Hemodynamic changes induced by stenting in elastic arteries, **Annual Research Briefs**, Center for Turbulence Research, NASA Ames/Stanford Univ, 2002, 335-348