◆ ENDOVASCULAR THERAPY REVIEWS ◆

Clinical Significance and Technical Assessment of Stent Cell Geometry in Carotid Artery Stenting

Gail M. Siewiorek, PhD¹; Ender A. Finol, PhD²; and Mark H. Wholey, MD³

¹Biomedical Engineering Department, Carnegie Mellon University, Pittsburgh, Pennsylvania, USA. ²Institute for Complex Engineered Systems, Biomedical Engineering Department and Department of Mechanical Engineering, Carnegie Mellon University, Pittsburgh, Pennsylvania, USA. ³Department of Radiology, University of Pittsburgh Medical Center – Shadyside, Pittsburgh, Pennsylvania, USA.

Carotid artery stenting has gained popularity due to its minimally invasive approach. However, several design concerns preclude the successful use of carotid stents. Technical issues, such as open versus closed cells, scaffolding, trackability, foreshortening, and changes in local geometry and hemodynamics, affect stent performance. Previous clinical and experimental studies have evaluated current stent models while proposing and testing novel stent designs. This review focuses on the technical aspects of carotid stent design and the clinical significance of key design parameters identified via computational and experimental modeling.

J Endovasc Ther. 2009;16:178-188

Key words: carotid artery, stent, distal protection, stenosis, thrombosis, plaque, emboli, shape memory alloy

Balloon-expandable closed-cell stents for carotid artery stenting (CAS) were first introduced in 1995. At that time, the cell structure of these stents was intended to support tissue and plaque against the vessel wall (Fig. 1) and restore the vessel to its normal dimension (Fig. 2). Unfortunately, there was a 2% incidence of carotid stent collapse: the stainless steel stent, when placed lower than the level of the mandible, could be externally compressed.¹ Consequently, the use of the balloon-expandable stent for carotid artery occlusive disease was largely abandoned. However, several valuable lessons were learned from these early studies. Not only were radial rigidity and scaffolding important features in stent design, but having a range of stent expansion rates became a necessity. The early designs lacked articulations, and

their inflexibility and high profile prevented easy trackability. To improve the flexibility and trackability of the stent, a connecting bridge between two stents was integrated into the design (Fig. 3); however, plaque prolapse was seen at this site (depending on the length of the bridge), a critical observation for future generations of stent designs.

When the non-collapsible memory alloy nickel titanium (nitinol) stent was introduced, there were several established design requirements for the delivery system: flexibility, device profile, fixation, and trackability. Moreover, stent designers took into consideration factors such as vessel conformability, scaffolding, side branch preservation, stent visibility, and recoil control to minimize migration and foreshortening (length contraction during stent expansion).² To evaluate expan-

Mark H. Wholey has disclosed that he is a consultant to Cordis, Abbott (Guidant), Medrad, Boston Scientific, Edwards LifeSciences, and Mallinckrodt. The other authors have no commercial, proprietary, or financial interest in any products or companies described in this article.

Address for correspondence and reprints: Ender A. Finol, PhD, Carnegie Mellon University, Hamburg Hall 1205, 5000 Forbes Ave., Pittsburgh, PA 15213 USA. E-mail: finole@cmu.edu

Figure 1 ♦ Cross section of a specimen demonstrating a balloon-expandable stent with excellent apposition and plaque scaffolding.

sion rates and potential plaque prolapse, protocols were designed to measure the radial forces occurring under controlled pressure conditions.

Mathematical studies have addressed several design issues involved in stent geometry, such as changes in local curvature, compliance mismatch between the vessel wall and stent material, and hemodynamics [e.g., the presence of low wall shear stress (WSS)]. Curvature changes at stent ends can induce regions of low WSS, which has been linked to neointimal hyperplasia and possible restenosis.³ Compliance mismatches in the mechanical properties of the artery and stent material can increase flow impedance, decrease distal perfusion, and disturb the flow.⁴ Blood flow

Figure 2 ♦ (A) A significant stenotic lesion with major ulceration in the right ICA (B) effectively managed with a stent precisely positioned at the ostium.

Figure 3 ♦ First-generation Palmaz-Schatz stent demonstrating the bridge that allowed increased length with flexibility.

patterns can influence delivery of blood-borne platelets, surface-adherent monocytes, and tissue-infiltrating monocytes to the artery wall, which also leads to the development of neointimal hyperplasia and restenosis.⁴ The objective of this review is to examine stent designs used in the carotid artery from clinical and computational perspectives, with special emphasis on the analysis of open-versus closed-cell stents.

CLINICALLY RELEVANT INFORMATION CONCERNING STENT DESIGN

Recently, there has been significant debate on the indications, advantages, and limitations of an open-cell stent (e.g., Precise and Acculink) in contrast with a closed-cell stent (e.g., Wallstent, Xact, and NexStent; Fig. 4). In a retrospective dual center study of 701 patients undergoing CAS, Hart et al.⁵ reported their observations within the context of a binary categorization of “open” or “closed” cell stents. The stroke and death rate in the study was 1.4%. When transient ischemic attacks (TIAs) were included, the total stroke, death, and TIA event rate was 3.7%: 4.6% in symptomatic patients and 3.0% in asymptomatic patients. When open-cell stents were

Figure 4 ♦ Closed-cell design with a detailed view of the bridge (arrows) demonstrating flexibility and conformability after expansion.

TABLE 1
Comparison of Procedural Event Rates* From the EXACT and CAPTURE 2 Trials

Event	EXACT, % (min, max 95% CI) (n=2145)	CAPTURE 2, % (min, max 95% CI) (n=4175)
Death, stroke, MI	4.1 (3.3, 5.1)	3.7 (3.1, 4.3)
Death, stroke	4.1 (3.3, 5.0)	3.4 (2.9, 4.0)
Death, major stroke	1.5 (NR)	1.4 (NR)
Death	0.9 (0.5, 1.4)	0.9 (0.6, 1.2)
Stroke	3.6 (2.8, 4.5)	2.8 (2.3, 3.3)
Major stroke	1.1 (0.7, 1.6)	0.8 (0.6, 1.2)
Minor stroke	2.5 (1.9, 3.3)	2.0 (1.6, 2.4)
MI	0.2 (0.1, 0.5)	0.4 (0.2, 0.6)

Summarized from Gray et al.⁹

EXACT: Emboshield and Xact Post Approval Carotid Stent Trial, CAPTURE: Carotid Acculink/Accunet Post Approval Trial to Uncover Unanticipated or Rare Events, CI: confidence interval, MI: myocardial infarction, NR: not reported.

used, there was an 11.1% stroke, death, and TIA event rate in symptomatic patients versus 3.0% with closed-cell stents in this subgroup. However, it should be noted that 74% of these patients received the Carotid Wallstent; hence, operator familiarity may have played a role in the outcomes.

Not surprisingly, data from other retrospective reviews provide different results. In a retrospective analysis of over 3000 patients, Bosiers et al.⁶ found statistically significant differences for total events (stroke, death, TIA) at 30 days between open- (4.2%) and closed-cell stents (2.3%, $p < 0.005$), which were particularly pronounced in symptomatic patients ($p < 0.0001$). On the other hand, Schillinger et al.⁷ failed to identify any difference between open- and closed-cell stents with respect to 30-day neurological complications, stroke, or mortality risk in a more recent multicenter review of >1600 patients. The Carotid Acculink/Accunet Post Approval Trial to Uncover Unanticipated or Rare Events (CAPTURE) surveillance study using the open-cell Acculink stent with the Accunet filter recorded a stroke rate of 4.8% in 3500 patients.⁸ Results from the Emboshield and Xact Post Approval Carotid Stent Trial (EXACT) using the closed-cell Abbott Xact stent and Emboshield filter described a stroke and death rate of 4.1% compared to CAPTURE 2, which had a stroke and death rate of 3.4%.⁹ Comparing the data from these trials (Table 1), the 95% confidence intervals for the stroke and death rate in the studies over-

lapped (EXACT: 3.3% to 5.0%; CAPTURE 2: 2.9% to 4.0%).

Wholey et al.¹⁰ provided evidence in favor of the endovascular treatment of asymptomatic patients regardless of stent cell type, citing that CAPTURE, EXACT, and the Carotid Artery Revascularization Using the Boston Scientific EPI FilterWire EX/EZ and the EndoTex NexStent (CABERNET) trials were able to meet the 3% procedural events rate guideline for asymptomatic patients published by the American Heart Association.

In another US post-market study, the CASES-PMS trial with the open-cell Precise stent, a 4.5% 30-day all stroke and death rate was reported.¹¹ Finally, in an analysis of 203 of our most recent patients at the University of Pittsburgh Medical Center who had an average 10.3-month follow-up and had undergone CAS with the Acculink stent and the Accunet filter, 1.6% developed restenosis. There was a 1.0% incidence for all strokes, 1.0% for myocardial infarction (MI), and a 7.8% all death event rate. While this is also a retrospective study based on a small sample size, it does add to the debate of which factors play a role in stent-related embolization. Table 2 summarizes the clinical outcome data.

Historically, numerous stent designs have been evaluated in clinical practice, with varying results. It is tempting to question these results on the basis of binary attributes, such as open- versus closed-cell, or one-dimensional attributes such as "wall thick-

TABLE 2

Comparison of Procedural Event Rates From Clinical Studies Evaluating Open- and Closed-Cell Stents

Event	Study	Open-Cell, %	Closed-Cell, %
Death, stroke, MI	Fairman ⁸	6.3	—
	CAPTURE 2 ⁹	3.7	—
	EXACT ⁹	—	4.1
Death, stroke, TIA	Hart ⁵	11.1	3.0
	Schillinger ⁷	4.1	6.1
Death, stroke	Fairman ⁸	5.7	—
	CAPTURE 2 ⁹	3.4	—
	Hart ⁵	1.6	0.9
	Schillinger ⁷	2.4	3.1
	EXACT ⁹	—	4.1
Death, major stroke	Fairman ⁸	2.9	—
	CAPTURE 2 ⁹	1.4	—
	EXACT ⁹	—	1.5
Death	Fairman ⁸	0.9	—
	CAPTURE 2 ⁹	0.9	—
	Hart ⁵	0.0	0.4
	EXACT ⁹	—	0.9
All stroke	Fairman ⁸	4.8	—
	CAPTURE 2 ⁹	2.8	—
	EXACT ⁹	—	3.6

EXACT: Emboshield and Xact Post Approval Carotid Stent Trial, CAPTURE: Carotid Acculink/Accunet Post Approval Trial to Uncover Unanticipated or Rare Events, MI: myocardial infarction, TIA: transient ischemic attack.

ness” or “cell size.” Such analyses may be misleading, however, because of ambiguous definitions of these characteristics or other important and often interrelated design characteristics that may not be considered. The open- and closed-cell designations serve as

good examples: it is possible to design an “open”-cell stent with a cell size smaller than that of a “closed”-cell stent, thus rendering the distinction nearly meaningless (Fig. 5). Similarly, a braided wire stent and nitinol stent might both be classified as closed cell but share no meaningful design attributes. In stent design, just as in clinical investigations, single variables often suggest some meaningful insight, but we must be mindful that outcomes are driven not by single variables but rather an interrelated system of variables.

Figure 5 ♦ (A) Closed-cell stent demonstrating the diamond configuration with radial segments and bridge connection (arrow); (B) open-cell stent with removal of a bridge connection illustrated.

TECHNICAL ASSESSMENT OF STENT DESIGN

The conflicting clinical data and consequential debates raise important questions regarding stent cell type, surface area coverage, and the identification of important factors in periprocedural embolic events. Clinical outcomes related to stent design are a function of performance attributes, such as longitudinal and circumferential stiffness and strength, scaffolding properties, conformability, and side branch preservation. Stent designers

attempt to balance these long-term performance attributes with more acute considerations related to deliverability and deployment, such as constrained profile and flexibility. All these attributes must be balanced appropriately given the anatomical challenges and the dynamic forces exerted by the vessel on the stent. Designers and engineers must select among numerous interrelated parameters (e.g., strut length and width, wire diameter and pitch, bridge configuration, material selection, and processing conditions) to achieve optimal performance.

Consider radial force as an example of the difficulty of assessing a performance attribute in stent design. The radial force required for stent apposition will be dependent on lesion characteristics and location: aorto-ostial lesions will likely require more force than internal carotid artery (ICA) stenoses. The strength of a carotid stent in the radial direction should provide adequate apposition and prevent vessel wall recoil. Excessive force in open-cell stents or rigid closed-cell stents might result in plaque disruption in a vulnerable unstable lesion due to these cell types having increased rigidity. Most nitinol stents can be adjusted for cell size and still maintain satisfactory radial force.

Given the aforementioned technical considerations, it is important to anticipate that binary classifications such as “open” and “closed” stents may be too general for meaningful retrospective clinical analysis, in particular when multiple stent designs are collectively included within these two broad classifications. Reflecting on the comparative studies available, what appears to be important are cell size and surface area coverage. Whether the stent is designed with open or closed cells may be less important than the actual cell size. For example, a closed cell with a diameter of 1000 μm is more likely to be responsible for plaque prolapse and embolization than an open cell of 500 μm .

Typically, the number and arrangement of bridge connectors differentiate open-cell from closed-cell designs (Fig. 5). If adjacent ring segments are connected at every possible junction, the design is typically classified as closed cell. In closed-cell designs, these

Figure 6 ♦ (A) Fully supported closed-cell stent, demonstrating comparable flexibility to the (B) unsupported open-cell stent.

junctions usually take the form of flexible bridge connectors, allowing some limited degree of flexion between adjacent rings (Fig. 6). If some or all of the connecting junction points are removed, the design is typically classified as open cell. Such a design with fewer connection points inherently allows more flexion and conformability. The flexion benefits of an open-cell design have a cost in scaffolding uniformity, just as the scaffolding benefits of a closed-cell design have a cost in flexion and conformability.

Vessel anatomy too plays an important role. If there are complex angulations at the carotid bifurcation where trackability becomes relevant, the flexibility of open-cell design may be required (Fig. 7A) knowing that the potential for plaque prolapse may exist. However, we can control the potential for plaque prolapse by reducing the size of the open cell. When cells open on the concave surface of an angulated carotid bifurcation, they are prone to “fish scaling” on the open surface (Fig. 7B), which can result in intimal disruption with contrast extending to the adventitia (Fig. 7C). Penetration of the struts in the open-cell configuration can extend to the adventitia, which raises the question of predisposition to restenosis or stent fracture at the deployment site (Fig. 8). However, there are open-cell stents that meet the criterion of small cell size, which affords cell conformability during flexion and provides acceptable scaffolding. Figure 9 illustrates 3 examples of stent conformability with both open- and closed-cell designs. Though Wallstent does not conform to the tortuosity of the vessel wall, its cell size is maintained,

Figure 7 ♦ (A) Larger open-cell design stents may not provide adequate scaffolding in a complex bend, but do provide conformability. (B) Open-cell design at the concave surface of the stent showing scaling. (C) Post-procedure angiogram demonstrating the open-cell struts extending beyond the intima, with focal contrast extravasation to the adventitia.

and so are the cell structures in the NexStent and Precise stents. The stability of the cell size will prevent prolapse and possibly decrease embolic events long term.

STENT MODELING AND THE IMPACT OF STENT DESIGN

The effect of stent implantation on the local geometry (i.e., curvature, compliance matching, etc.) and hemodynamics (i.e., WSS) have been modeled, as well as the effects of closed- and open-cell designs. Numerical studies are also important in the process of virtual stent design and optimization. Wu et al.¹² investigated the delivery and expansion of various stent designs in both straight and curved vessels using finite element modeling (FEM) of 2 segmented-design nitinol stents in a stenosed, tortuous carotid bifurcation. Each

Figure 8 ♦ Strut fracture in a nitinol carotid stent as seen on a 3-dimensional reconstructed angiogram.

simulated stent had the same dimensions (diameter 8 mm, length 22 mm, thickness 0.23 mm) and consisted of 12 stent unit rings in a cylindrical pattern connected longitudinally every other ring; however, they differed in the number and length of strut units. S_{ori} has 6 strut units 3.15 mm long and S_{mod} has 9 strut units 2.10 mm long (Fig. 10). S_{ori} had more regions of malapposition than S_{mod} and had a greater percentage of residual stenosis (37% versus 46%); however, S_{mod} caused more ICA tortuosity changes and higher stress concentrations at the lesion. The authors noted that the final stenosis rate for both designs was unacceptable. Despite the drawbacks, Wu and colleagues concluded that stents with shorter struts (e.g., S_{mod})

Figure 9 ♦ Stent configurations in a carotid glass model with variations in conformability: (from left to right) NexStent, Wallstent, and Precise stents.

Figure 10 ♦ (A) Cross section of the carotid artery. CCA: common carotid artery, ICA: internal carotid artery, ECA: external carotid artery. (B) The CCA–ICA angle of the vessel, which reflects ICA tortuosity. (C) The segmented nitinol stent models S_{ori} and S_{mod} . Reproduced with permission from *Journal of Biomechanics*.¹² Copyright 2007 Elsevier.

may lead to more favorable clinical outcomes due to increased radial expansion force and increased conformability to tortuous vessels, which may contribute to less irregular blood flow patterns. However, one must note the tradeoffs between the 2 designs.

Alderson and Zamir¹³ investigated mathematically the effects of stent stiffness and stent length on wave reflections. Rigid stents placed in elastic blood vessels caused an impedance mismatch that created sites of wave reflections at the stent ends. A stent produces a higher stress at its entrance than at the exit, which aids flow through the stent. Short stents located at the entrance of a diseased vessel affect hemodynamics the least while long stents placed near the end of a diseased vessel aid flow the most.

Compliance is a measure of the elasticity of the material or the ratio of the rate of volume change to the rate of pressure change. Moore et al.¹⁴ conducted computational studies of compliance mismatch between a stent and an artery, which can affect the stresses in the arterial wall and can contribute to restenosis. The authors found that compliance-matching stents, which provide a smooth transition between the compliance of the stent and of the artery, reduce this stress concentration on the wall.

Prabhu¹⁵ used the FEM approach to model the loading conditions that a self-expanding

nitinol stent experiences during the manufacturing process (stent expansion to maximum diameter, stent annealing, and stent crimping into delivery system) and stent deployment in an artery. The stress due to localized contact between the stent and artery wall could be as much as an order of magnitude higher than the circumferential stress due to stent expansion. The author recommended a stent that more uniformly contacts the artery wall to more evenly distribute contact stresses and eliminate localized stresses.

Kolachalama et al.¹⁶ utilized fluid-structure interaction simulations and a pulsatile one-dimensional Navier-Stokes solver to investigate the effect of arterial and stent geometries on pressure changes within the vessel. A metric pressure variation factor (PVF), which indicates the extent of deviation of pressure from the ideal pressure in a cross section of a vessel, was evaluated to capture the gross effect of pressure changes. A nondimensionalized maximum pressure p_m^* was computed to capture local effects of pressure change. It was found that p_m^* is less sensitive to changes in stent length and Young's modulus of the stented portion of the artery than PVF, suggesting that localized pressure peaks can be generated by a short length stent that has low compliance relative to the artery.

Several investigators have explicitly modeled open- and closed-cells stents. Although

the studies target balloon-expandable stents, which are no longer used during CAS, important information can be derived from these studies. Xia et al.¹⁷ investigated closed-cell balloon-expandable stents using FEM and the repeated unit cell (RUC) approach to model the Palmaz-Schatz stent and a NIR-type stent with V- and S-shaped links. The longitudinal stress was the most dominant component due to rotational deformation of stent struts around the bridge following expansion. The authors found that stents with V- or S-shaped links were easier to inflate than the Palmaz-Schatz stent, but tended to foreshorten more. S-shaped links are more flexible than V-shaped links. The expanded diameters of the stent only, balloon only, and stent and balloon were mapped as a function of internal pressure. The stent size increased nonlinearly from the original 3-mm diameter to ~5.5 mm as the internal pressure increased from 0 to ~0.4 MPa. This study demonstrated that modeling a quarter of the stent can reduce computational time while keeping the same measurements of displacement and stress (measurement differences of <1.0%). Ju and others¹⁸ expanded upon Xia's work by modeling computationally the closed-cell stents with RUC and with a free end (RUC⁺). The study compared 3 models (Panel, RUC, RUC⁺) applied to 2 stents (Palmaz-Schatz and a sinusoidal stent); they found good agreement with previous studies without losing computational accuracy by way of RUC⁺, yielding the most accurate results for the inner surface and ends of the stent.

Computational models of stents in the coronary artery were created by Lim et al.¹⁹ with open or closed cells linked by bend-shaped structures. A realistic, transient, non-uniform balloon expansion process was analyzed using FEM. Seven commercially available stents (Palmaz-Schatz PS153, TenaxTM, MAC Standard, MAC Q23, MAC Plus, Coroflex, and RX Ultra Multi-link) were compared to recommend design parameters that could reduce the risk of restenosis due to foreshortening or dog-boning (flaring of stent ends due to excessive balloon expansion). Although dog-boning is an issue related to balloon-expandable stents, Lim's modeling can be applied to self-expanding stents to

Figure 11 ♦ Design parameters for a generic stent showing the 3 parameters identified by the optimization algorithm: strut spacing (h), radius of curvature (ρ), and axial amplitude (f). Reproduced with permission from *Medical and Biological Engineering and Computing*.²¹ Copyright 2007 Springer.

suggest design parameters. Foreshortening ranged from 0.0% to 12.9% and dog-boning values were all positive, indicating overexpansion of the distal portion of the stent. Hence stents with closed cells tend to have greater foreshortening and dog-boning.

Bedoya et al.²⁰ investigated stent design parameters based on vessel stress distribution and radial displacement using a generic stent model with varying connector bar lengths (strut spacing), radius of curvature at the crown junctions, and axial amplitude. Strut spacing changed the results most drastically in that small strut spacing (1.2 mm versus 2.4 mm) could induce higher stresses over a larger region. The authors recommended that future stents be designed with large strut spacing, blunted corners at bends, and high axial amplitude. Timmins et al.²¹ expanded upon Bedoya's work by using FEM to numerically evaluate arterial wall stress, lumen gain, and cyclic deflection (change in radial position from diastolic to systolic pressure) of a stent. An algorithm was developed to optimize the stent design using single values representing these 3 variables in the computational model. The goal of the algorithm was to maximize lumen gain and cyclic deflection while minimizing wall stress, using different weighting factors to emphasize the importance of various parameters. When circumferential stress reduction or a cyclic deflection increase was emphasized, strut spacing and axial amplitude increased while radius of curvature

Figure 12 ♦ (A) Virtual histology images at 3 sites in a carotid artery demonstrating lesion characteristics that are best suited for endarterectomy rather than stenting. Note the absence of any fibrous cap and the exposure to the necrotic core at the intimal surface. (B) Pre-occlusive ICA lesion with complex ulceration and angulation. (C) Virtual histology of the lesion demonstrating significant necrotic core and dystrophic calcification at the intimal surface. These lesion characteristics are best suited for endarterectomy.

decreased. Increasing the emphasis on lumen gain resulted in a decrease in strut spacing and axial amplitude and an increase in radius of curvature (Fig. 11).

FUTURE CONSIDERATIONS

With the recent introduction of intravascular ultrasound (IVUS) with virtual histology (VH, Fig. 12A), it may be possible to better understand the influence of lesion characteristics (e.g., plaque thrombogenicity) on the significance of cell size, stent flexibility and conformability, and plaque prolapse potential, all of which are predictors of stroke. The angiographic anatomy may provide a better understanding of lesions more demanding of flexibility or scaffolding, while IVUS-VH might suggest lesions that are thrombotically active and not suited for stenting regardless of cell design (Fig. 12B,C). It is reasonable that the cell structure itself can be designed to act as its own embolic protection filter, i.e., an increase in surface area coverage may prevent plaque prolapse and em-

Figure 13 ♦ (A) A high-grade stenotic lesion with ulcerative changes best suited for a closed-cell stent; (B) post-procedure angiogram showing flow pathway restored.

bolization, effectively trapping plaque fragments between the stent and the vessel wall. With VH and an improved understanding of the lesion characteristics, we are better equipped today to choose the most appropriate stent.

Early clinical experience with helically designed stents has demonstrated excellent flexibility and kink resistance. The helical design allows uniform cell size at flexion points without scaling. In terms of clinical applications, the calcified and tortuous lesion may be best suited to the more conformable open or helically designed stents. Greater scaffolding may be more appropriate for vulnerable plaque or ulcerative lesions with type C characteristics (i.e., with high anatomical risk, Fig. 13).

It is evident that there remain unresolved clinical issues and the need for improvement in stent designs. While we look forward to robust datasets and device evolution, it is important that the interventional community remains mindful of the lessons of recent clinical trials in both the US and Europe. An interesting observation in the CAPTURE study was the existence of a difference (although not statistically different) in outcomes among inexperienced versus experienced operators.²² Major stroke and death occurred at a 3.4% event rate in the lesser experienced operators compared to 1.1% in the operators with significant experience. Furthermore, an

all stroke and death rate comparison of inexperienced and experienced operators also showed a difference of 6.9% versus 4.6%, respectively. The lesson to be learned is that interventionists can and do achieve excellent stenting results when they are well trained in the use of the devices, the stenting protocol, and appropriate patient selection.

Current stent geometries can be evaluated using computational modeling tools, which also aid in the design and optimization of new stents that improve upon the shortcomings of the existing devices in the market. Computational studies can parametrically vary stent geometries, which give insight into which configurations have the most favorable effects on the vessel wall and local hemodynamics. Virtual design of medical devices using FEM reduces the prototype building and evaluation periods.

CONCLUSION

The desired features of a carotid artery stent include (1) scaffolding adequate to control plaque prolapse but with acceptable flexibility, (2) conformability, and (3) radial strength to track the lesion, appose the vessel wall, and control recoil. Evidently, no matter how tight the scaffolding, there is still no stent available on the market to control emboli $\leq 100 \mu\text{m}$ in diameter, which comprise most of the emboli produced during CAS. When used with an efficient distal protection filter and combined with a flow reversal system, we may be able to reduce most of the embolic events occurring in the periprocedural period. Nonetheless, predicting and controlling delayed post-procedure events remains a challenge.

By means of FEM and design optimization algorithms, new stent configurations can be evaluated in a simulated environment prior to prototype building. The modeling issues relevant to stent design include (1) curvature changes, (2) compliance mismatches, and (3) low wall shear stress. Previous studies have found that stents with shorter and fewer struts, as well as reduced thickness, yield better results. Compliance-matching stents aid blood flow and reduce stress concentrations. Careful consideration must be taken in the design of the stent edges, as the ends

have a tendency to induce elevated mechanical stresses, cause large curvature changes that lead to low wall shear stress, and damage the arterial wall after expansion.

Acknowledgment: The authors thank Roseanne R. Wholey for her assistance with the preparation of this article.

REFERENCES

1. Wholey MH, Al-Mubarak N, Wholey MH. Updated review of the global carotid artery stent registry. *Catheter Cardiovasc Interv.* 2003;60:259–266.
2. Palmaz JC. Intravascular stents in the last and the next 10 years. *J Endovasc Ther.* 2004; 11(Suppl II):II-200–II-206.
3. Wentzel JJ, Whelan DM, van der Giessen WJ, et al. Coronary stent implantation changes 3-D vessel geometry and 3-D shear stress distribution. *J Biomech.* 2000;33:1287–1295.
4. Duraiswamy N, Schoepfoerster RT, Moreno MR, et al. Stented artery flow patterns and their effects on the artery wall. *Annu Rev Fluid Mech.* 2007;39:357–382.
5. Hart JP, Peeters P, Verbist J, et al. Do device characteristics impact outcome in carotid artery stenting? *J Vasc Surg.* 2006;44:725–731.
6. Bosiers M, de Donato G, Deloose K, et al. Does free cell area influence the outcome in carotid artery stenting? *Eur J Vasc Endovasc Surg.* 2007;33:135–141.
7. Schillinger M, Gschwendtner M, Reimers B, et al. Does carotid stent cell design matter? *Stroke.* 2008;39:905–909.
8. Fairman R, Gray WA, Scicli AP, et al. The CAPTURE registry: analysis of strokes resulting from carotid artery stenting in the post approval setting: timing, location, severity, and type. *Ann Surg.* 2007;246:551–558.
9. Gray WA, Chaturvedi S, Verta P. 30-day outcomes for carotid artery stenting in 6320 patients from two prospective, multicenter, high surgical risk registries. *Circ Cardiovasc Interv.* 2009; in press.
10. Wholey MH, Barbato JE, Al-Khoury GE. Treatment of asymptomatic carotid disease with stenting: pro. *Semin Vasc Surg.* 2008;21:95–99.
11. Katzen BT, Criado FJ, Ramee SR, et al. Carotid artery stenting with emboli protection surveillance study: thirty-day results of the CASES-PMS study. *Catheter Cardiovasc Interv.* 2007;70:316–323.
12. Wu W, Qi M, Liu XP, et al. Delivery and release of nitinol stent in carotid artery and their interactions: a finite element analysis. *J Biomech.* 2007;40:3034–3040.

13. Alderson H, Zamir M. Effects of stent stiffness on local haemodynamics with particular reference to wave reflections. *J Biomech.* 2004;37:339–348.
14. Moore J, Berry J, Rolland P, et al. Finite element modeling and in vivo evaluation of a compliance matching stent. *Ann Biomed Eng.* 2000;28(Suppl 1):61.
15. Prabhu S. Finite element modeling of the interaction of a self-expanding stent with an artery. Proceedings of the Materials and Processes for Medical Devices Conference; November 14–16, 2006; Boston, MA, USA.
16. Kolachalama VB, Bressloff NW, Nair PB, et al. Predictive haemodynamics in a one-dimensional human carotid artery bifurcation. Part I: Application to stent design. *IEEE Trans Biomed Eng.* 2007;54:802–812.
17. Xia Z, Ju F, Sasaki K. A general finite element analysis method for balloon expandable stents based on repeated unit cell (RUC) model. *Finite Elem Anal Des.* 2007;43:649–658.
18. Ju F, Xia Z, Sasaki K. On the finite element modelling of balloon-expandable stents. *J Mech Behav Biomed Mater.* 2008;1:86–95.
19. Lim D, Cho SK, Park WP, et al. Suggestion of potential stent design parameters to reduce restenosis risk driven by foreshortening or dogboning due to non-uniform balloon-stent expansion. *Ann Biomed Eng.* 2008;36:1118–1129.
20. Bedoya J, Meyer CA, Timmins LH, et al. Effects of stent design parameters on normal artery wall mechanics. *J Biomech Eng.* 2006;128:757–765.
21. Timmins LH, Moreno MR, Meyer CA, et al. Stented artery biomechanics and device design optimization. *Med Biol Eng Comput.* 2007;45:505–513.
22. Gray WA, Yadav JS, Verta P, et al. The CAPTURE registry: results of carotid stenting with embolic protection in the post approval setting. *Catheter Cardiovasc Interv.* 2007;69:341–348.