Impact of Calcium Quantifications on Stent Expansions

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Severely calcified plaque is of great concern when planning and implementing a stenting intervention. In this work, computational models were developed to investigate the influence of calcium characteristics on stenting outcomes. The commonly used clinical measurements of calcium (i.e., the arc angle, maximum thickness, length, and volume) were varied to estimate stenting outcomes in terms of lumen gain, stent underexpansion, strut malapposition, and stress or strain distributions of the stenotic lesion. Results have shown that stenting outcomes were most sensitive to the arc angle of the calcium. A thick calcium with a large arc angle resulted in poor stenting outcomes, such as severe stent underexpansion, D-shaped lumen, increased strut malapposition, and large stresses or strains in the plaque. This was attributed to the circumferential stretch of the tissue. Specifically, the noncalcium component was stretched significantly more than the calcium. The circumferential stretch ratios of calcium and noncalcium component were approximately 1.44 and 2.35, respectively, regardless of calcium characteristics. In addition, the peak stress or strain within the artery and noncalcium component of the plaque occurred at the area adjacent to calcium edges (i.e., the interface between the calcium and the noncalcium component) coincident with the location of peak malapposition. It is worth noting that calcium played a protective role for the artery underneath, which was at the expense of the overstretch and stress concentrations in the other portion of the artery. These detailed mechanistic quantifications could be used to provide a fundamental understanding of the impact of calcium quantifications on stent expansions, as well as to exploit their potential for a better preclinical strategy. [DOI: 10.1115/1.4042013]

Keywords: calcium, stent deployment, percutaneous coronary intervention, arc angle, stent underexpansion, malapposition, D-shaped lumen, arterial stress, finite element method

Introduction

Severely calcified plaque is well acknowledged as a strong factor affecting stenting outcomes [1,2]. It is evident from high resolution, intravascular optical coherence tomography (OCT) that stent implantations in severely calcified lesions are often imperfect. They often result in noncircular deployment, stent underexpansion, stent malapposition, and calcium fracture, which increase the risk of complications, including vessel microdissection, thrombosis, and in-stent restenosis [3–5]. The calcium score (i.e., the calcium area multiplied by a weight factor depending on the peak radio-density) is commonly used to predict the risk of coronary heart disease [6,7]. In addition, calcium geometric characteristics are associated with optimal stent expansion [1,8]. Specifically, the statistical analysis of 1155 native vessels by intravascular ultrasound and coronary angiography classified the calcium into the arc angle, maximum thickness, length, and depth [9]. In an intravascular optical coherence tomography study of 51 coronary artery lesions, it was found that the calcium angle (i.e., the arc angle of the calcium), instead of thickness and length, correlated with lumen gain after stenting [8]. Lindsay et al. showed that the arc angle of the superficial calcification correlated well with the percentage of malapposed struts [10]. To the contrary, another intravascular ultrasound study of 60 native vessels showed that acute stent expansion had no correlation with calcium characteristics in terms of the calcium score, length, maximum thickness, and area [11]. This might be associated with multiple postdilations adopted for lesions with a large calcium score. These postdilations could induce the calcium fracture and reduce the

plaque burden on stents. The analysis of 37 calcified lesions showed that calcium with an arc angle larger than 227 deg and a thickness less than 0.67 mm was prone to crack formation [12]. The quantitative characterizations of stent expansion will help to unveil calcium characteristics that impact stent expansion the most, and thus help provide a better plan for prestent and poststent strategies.

Finite element method has demonstrated its efficiency in providing mechanistic details of stenting beyond the in vitro and in vivo experimental data [13–16]. For calcified plaque, the major focus of computational studies has been on plaque vulnerability, especially the thin-cap fibroatheroma with a large lipid core [17]. Wong et al. showed that a thinner fibrous cap resulted in much higher stress concentrations, and thus an increased risk of fracture [18]. Generally, the thin-cap fibroatheroma with a lipid core was considered to be softer than the artery wall. However, a severely calcified plaque would be much stiffer than the artery wall, which poses challenges on stent expansion [19]. Pericevic et al. showed that the stiffness of a concentric plaque affected stent-induced arterial stresses dramatically [20]. Specially, a stiffer concentric plaque mitigated the stress level on the artery wall. Zhao et al. further illustrated that the presence of the calcified plaque reduced tissue prolapse [21]. The presence of severely calcified plaque is of great concern when planning and implementing a stenting intervention. However, there are scarce computational studies on the role of calcium characteristics in stent expansion.

The aim of this work is to quantitatively inspect the influence of calcium characteristics on stenting outcomes in term of lumen gain, stent underexpansion, strut malapposition, and lesion stress or strain distributions using finite element models. Four calcium parameters commonly used in clinical studies (i.e., the arc angle, maximum thickness, length, and the volume) were varied to quantitatively reveal the underlying mechanism of poor stenting

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Fig. 1 Representative model of the stent-plaque-artery interaction (top), where the calcium has the length (L) of 5 mm, the arc angle (θ) of 270 deg, and maximum thickness (δ) of 0.06 mm (bottom)

Table 1 Characteristics of the calcium

Models		Ca60-0.6	Ca180-0.6	Ca270-0.6	Ca270-0.4	Ca270-0.2
Arc angle (deg)		60	180	270	270	270
Maximal thickness (mm)		0.06	0.06	0.06	0.04	0.02
Calcium volume (mm ³)	Length 13 mm	6.54	17.7	22.8	15.9	8.31
	Length 5 mm	3.45	9.28	12.00	8.48	4.49

outcomes associated with a severely calcified plaque. The major determinant of stent underexpansion and malapposition was identified. The stress and strain distributions of both the plaque and artery were associated with stent performance as well as with potential tissue adaptation.

Materials and Methods

The commercial Express stent (Boston Scientific, Natick, MA) was implanted into a stylized artery with lumen diameter of 3 mm, wall thickness of 0.75 mm, and length of 26 mm (Fig. 1). The stent had a nominal diameter of 3 mm, thickness of 0.13 mm, and length of 16 mm. The asymmetric plaque had a parabolic longitudinal profile with the length of 13 mm. An edge ratio of 2:1 at the narrowest occlusion led to a lumen diameter of 1.5 mm, which was half of the reference lumen diameter, referred to as a 50% diameter stenosis. Calcium and the noncalcium components were considered for the plaque. The major characterizations of the calcium arc were quantified using its arc angle and maximum thickness in the transvers plane (Fig. 1 bottom), as well as the total length [9]. The variations in calcium geometrical parameters resulted in ten cases as listed in Table 1. Specifically, the arc angles of the calcium, also referred to as calcium angles, were 60 deg, 180 deg, and 270 deg, which represented clinical classifications in quadrants. For the case of the arc angle of 270 deg, the maximum thickness of the calcium was reduced from 0.6 mm to 0.4 mm and 0.2 mm by changing the superficial layer of the calcium into a noncalcium component. This could represent the plaque modification process, which is usually adopted prior to stenting in lesions with a large calcium angle [22]. The aforementioned five cases with the calcium length of 13 mm were considered as baseline models. The other five cases were obtained by reducing the calcium length to 5 mm, which is the plaque length most frequently observed in patients with calcified plaque [9]. In total, ten models were labeled, based on their calcium characteristics, as Ca θ - δ_N - L_N , in which θ represents the degree of the calcium angle,

δ_N represents the maximum thickness in mm, and L_N represents the longitudinal length in mm (i.e., either 5 mm or 13 mm). The corresponding calcium volume was calculated to estimate the potential role of calcium volume on the acute outcomes of stenting. Considering the model symmetry about the xz -plane, one half of the model was constructed to reduce the computational cost (Fig. 1).

The stent was made of 316L stainless steel, whose mechanical behavior is described as a perfect linear elastic-plastic material with a Young's modulus of 190 GPa, Poisson's ratio of 0.3, and yield stress of 207 MPa [23]. All tissues, including the artery and two plaque components, were modeled as an isotropic hyperelastic material using the following constitutive equation:

$$U = \sum_{i,j=1}^3 C_{ij}(I_1 - 3)^i(I_2 - 3)^j \quad (1)$$

where U is the reduced third-order polynomial strain energy density function and I_1 and I_2 are the first and the second invariants of the left Cauchy-Green tensor, defined as

$$I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \quad (2)$$

$$I_2 = \lambda_1^{-2} + \lambda_2^{-2} + \lambda_3^{-2} \quad (3)$$

The material coefficients C_{ij} were adopted from our previous work [21] as shown in Table 2. The perfect plastic model was used to describe the permanent deformation of the noncalcium component of the plaque, which is necessary for modeling the stent recoil process [24]. The plasticity was assumed to initiate at 34% strain when the stress reached the yield stress of 0.07 MPa [25].

Table 2 Material coefficients

	C ₁₀ (MPa)	C ₀₁ (MPa)	C ₁₁ (MPa)	C ₂₀ (MPa)	C ₀₂ (MPa)	C ₃₀ (MPa)	C ₀₃ (MPa)
Artery	0.10881	−0.101	−0.1790674	0.0885618	0.062686		
Plaque	0.04				0.003		0.02976
Calcium	−0.49596	0.50661	1.19353	3.6378		4.73725	

Table 3 Minimal lumen area (mm²) after stenting

Models	Ca60-0.6	Ca180-0.6	Ca270-0.6	Ca270-0.4	Ca270-0.2
Length 13 mm	7.91	5.78	4.24	4.48	4.91
Length 5 mm	7.93	5.74	4.10	4.46	4.86

Fig. 2 Minimal lumen area in relation to the combination of arc angle and volume of the calcium. The MLA had no correlation with the volume, but it was reduced with a large calcium angle.

The model was meshed into 39770 eight-node reduced integration elements (C3D8R element type). The element sizes of 0.02 mm, 0.12 mm, and 0.14 mm for stent, plaque, and artery, respectively, were adopted based on the mesh convergence study (i.e., difference in maximum von Mises stress smaller than 1.5%). The two ends of the artery were constrained as symmetrical planes. A general contact algorithm was used for all the contact surfaces to resist contact penetrations. A friction coefficient of 0.05 was adopted at the tissue–stent interface based on the measurements reported by Dunn et al. [26]. The uniform displacement was used to mimic the stent crimping, expansion, and recoil processes. The stent was first crimped from its nominal diameter of 3.0 mm to 1.0 mm, and then was expanded to a diameter of 4 mm. When the radial displacement was removed, the stent recoiled back. The model was solved using ABAQUS (Providence, RI).

Results

The Calcium Angle as the Major Determinant of Stent Underexpansion and Lumen Shapes. Following stenting, the minimal lumen area (MLA) was measured to evaluate the lumen gain (Table 3). The MLA following the stenting of less than 85% of the reference lumen area was referred to as stent underexpansion. It is clear that all the MLAs following stenting were less than 6.01 mm² (85% of reference lumen area) except for cases with a calcium angle of 60 deg. Specifically, as the arc angle increased from 60 deg to 270 deg, the MLA decreased by 46.4% (i.e., from 7.91 mm² to 4.24 mm²). As the calcium length increased 2.6 times, the MLA alternation was less than 3.30%. For the arc angle of 270 deg, as the calcium thickness increased three times, the MLA reduced by 15.6% (i.e., from 4.86 mm² to 4.10 mm²). The lumen gain following stenting was most strongly dependent on the calcium angle, and next the thickness. The calcium length had minimal influence on the lumen gain and the stent underexpansion.

Fig. 3 Stent induced stretch ratio (a) minimal lumen area of Ca 270-0.6-13 before (left) and after (right) stenting. (b) Five models with the same calcium length of 13 mm. Minimal alternations in the stretch ratio of each plaque components were observed regardless of the calcium angle and thickness (Ca_θ–δ_N–L_N: θ represented the degree of calcium angle, δ_N represented the maximum thickness in mm, and L_N represented the longitudinal length in mm).

The combined effects of arc angle and volume of the calcium on MLA are depicted in Fig. 2. Results were divided into three groups by the arc angle and five groups by the volume. It was obvious that the arc angle, not the volume, distinguished the levels of the MLA. The MLA ranged from 4.103 mm² to 4.908 mm² for cases with a calcium angle of 270 deg, from 5.5 mm² to 6 mm² for a calcium angle of 180 deg, and from 7.5 mm² to 8 mm² for a calcium angle of 60 deg. No correlation was found between the calcium volume and the MLA.

The cross section profiles of the MLA before and after stenting were demonstrated using the case of Ca270-0.6-13 (Fig. 3(a)). Points A and B represent the radial projection of calcium ends onto the inner wall before stenting. Points C and D represent the locations on the inner wall with maximum and minimum plaque thicknesses, respectively. Also, point C was radially aligned with the maximum calcium thickness. After stenting, the aforementioned positions were denoted with a prime ('). The stent-induced stretch was expressed as the length ratios of arc A'C'B' to ACB, as well as arc A'D'B' to ADB, also referred to as the stretch ratios of the calcium zone and noncalcium zone, respectively. Figure 3(b) depicts the stretch ratios for the five baseline cases. It is interesting to observe that the stretch ratios for all these cases are 1.44±0.094 for the calcium zone, and 2.35±0.13 for the noncalcium zone. As the calcium became thinner and deeper for the cases with a calcium arc of 270 deg, the inner surface of the calcium zone was slightly stretched more and the noncalcium zone less. Specifically, the stretch ratio of the calcium zone was increased by 21.21% due to the thinner calcium.

Fig. 4 Transvers plane cut of the stented artery for five models with a calcium length of 13 mm. A thicker calcium with a large angle resulted in a D-shaped lumen and more malapposition areas ($Ca_{\theta-\delta_N-L_N}$: θ represented the degree of calcium angle, δ_N represented the maximum thickness in mm, and L_N represented the longitudinal length in mm).

Table 4 Calcium characterizations before and after stenting

		Ca60-0.6-13	Ca180-0.6-13	Ca270-0.6-13	Ca270-0.4-13	Ca270-0.2-13
Open angle (deg)	Initial value	149	90.5		54.4	
	After recoil	144	101.5	76.7	75.7	72.1
Chord (mm)	Initial value	1.37	2.70		2.18	
	After recoil	1.42	2.99	3.11	3.19	3.16
Calcium height h_c (mm)	Initial value	NA	0.54		1.31	
	After recoil	NA	0.47	1.27	1.36	1.47
Noncalcium height h_n (mm)	Initial value	NA	0.96		0.19	
	After recoil	NA	2.43	0.83	0.90	0.97

It is worth noting that the final lumen shape was also sensitive to the calcium angle and the thickness (Fig. 4). Specifically, the lumen was a desirable nearly circular shape for the case of Ca60-0.6-13 and became an approximate D-shape for the case of Ca270-0.6-13. The D-shape could revert toward the nearly circular shape by reducing the thickness of the calcium, as might be done in an atherectomy. This could be explained by the deformations of the rigid calcium arc, which were quantified using the following four parameters as illustrated in Fig. 3. The open angle of the calcium (i.e., $\angle EFG$) is formed by three positions E, F, and G, which represent the two ends and middle point at the outer surface of the calcium. The line EG is the chord length of the calcium arc. The distances from points C and D to the chord are referred to as the calcium height (h_c) and noncalcium height (h_n), respectively. The calcium characterizations before and after stenting were quantified for the baseline cases as listed in Table 4. It was clear that the stent deployment altered the calcium shape. Before stenting, the three cases of the calcium angle of 270 deg exhibited the same open angle, chord length, calcium height, and noncalcium height, regardless of the calcium thickness. The open angle increased 32–41%, with the lower value corresponding to the thinner calcium. The calcium height was reduced 3.1% (from

1.31 mm to 1.27 mm) for the 0.6 mm calcium arc. However, it was increased for the other two thinner ones. The chord elongation ranged from 42% to 46%. It is interesting to see that the 0.2 mm-thick calcium arc had a smaller chord than the 0.4 mm one. This indicated that there were contour changes of the compliant calcium arc, which were reflected in the increased ratio between the calcium height and the chord. The h_c -to-chord ratio was calculated as 0.408, 0.426, and 0.465 for the three 270 deg cases with calcium thicknesses of 0.6 mm, 0.4 mm, and 0.2 mm, respectively. The noncalcium height significantly increased 3.4–4.1 times. The constrictions from both calcium and the noncalcium components resulted in a lumen size smaller in the x direction than in the y direction, especially for a thicker calcium arc. The rigidity and contour of the calcium arc could drive the stent recoil with more confinement in the vertical direction (y) than in the horizontal one (x). Together, heavily calcified plaques with a large arc angle caused the commonly observed D-shaped lumen to form. A reduced calcium thickness or arc angle would alter the under-expanded D-shape to a nearly circular shape. For a smaller calcium angle, there were fewer alternations in the shape of the calcium arc. Much larger increases were observed in the noncalcium component height, as expected. The stent expansion

Fig. 5 Malapposition of the stent struts (longitudinal plane cut) for models with different length Ca 270-0.6-13 (top) and Ca 270-0.6-5 (bottom) ($\text{Ca}_{\theta}-\delta_N-L_N$: θ represented the degree of calcium angle, δ_N represented the maximum thickness in mm, and L_N represented the longitudinal length in mm)

Table 5 The maximal gap between the malapposed stent strut and lumen (mm)

Models	Ca60-0.6	Ca180-0.6	Ca270-0.6	Ca270-0.4	Ca270-0.2
Length 13 mm	0	0.122	0.638	0.427	0.144
Length 5 mm	0	0.133	0.592	0.382	0.117

mainly relied on the stretch of the noncalcium component. The noncalcium component near the edge of the calcium, along the circumferential direction, was concave downward, which is associated with plastic deformation and malapposition.

The Role of Calcium Geometry on Stent Malapposition. Stent malapposition, also referred to as incomplete contact of the stent to the vessel wall, is defined as the separation of the stent strut from the lumen surface. Stent struts in relation to the vessel wall at the location of the MLA are shown in Fig. 4 for all the baseline cases. It is clear that the stent was well apposed to the vessel wall for a small calcium angle of 60 deg. Stent malapposition occurred around the calcium edge for the calcium angle of 180 deg and worsened for the increased calcium angle of 270 deg. The malapposition of the stent struts was mitigated as the calcium thickness reduced from 0.6 mm (Ca270-0.6-13) to 0.2 mm (Ca270-0.2-13). In addition, the longer calcium caused a larger area of stent malapposition (Fig. 5), even though it had minimal impact on the peak malapposition of the stent struts at the transverse plane of the MLA.

The maximal gap between the malapposed stent strut and the lumen surface was quantified, as listed in Table 5, to indicate the severity of the malapposition. It is clear that there were no malapposed struts for cases with a calcium arc of 60 deg regardless of the calcium length. The maximum gap of the malapposed strut increased from 0.124 mm to 0.638 mm as the arc angle of the calcium increased from 180 deg to 270 deg (Ca180-0.6-13 versus Ca270-0.6-13). As the maximum thickness of the calcium decreased from 0.6 mm to 0.2 mm (Ca270-0.6-13 versus Ca270-0.2-13), the maximal malapposition decreased from 0.638 mm to 0.144 mm; a thinner and deeper calcium improved the stent apposition.

The Role of Calcium on Tissue Mechanics. The arterial stress distributions at the full expansion and recoil states for all five baseline cases are shown in Fig. 6. When the stent was fully expanded, the volume percentage of high stress increased with the larger calcium angle and thickness. This was due to the overstretch of the artery wall. Stress or strain concentrations were always at the interface between the calcium and the noncalcium components. The peak Von Mises stress increased from 0.23 MPa to 0.70 MPa as the calcium angle increased from 60 deg to 270 deg. It was then reduced to 0.49 MPa as the calcium thickness changed from 0.6 mm to 0.2 mm. To the contrary, the opposite trend was observed after stent recoil. Specifically, the volume

percentage of high stress on the artery wall decreased with a larger calcium angle and thickness. In addition, the higher stresses were shifted to the ends of the calcium arc. This implied that the calcium arc with a larger angle and/or thickness correlated with a larger recoil, which in turn was associated with more severe stent underexpansion and malapposition.

The histogram analysis of the maximum principal strain (MPS) in the calcium component of the plaque is depicted in Fig. 7. Regardless of the expansion state, the peak MPS, as well as the volume percentage of higher strain in the calcium, increased with a larger calcium angle and thinner calcium. This implied a higher risk of calcium fracture for a larger calcium angle and thinner calcium. In addition, minimal shape alternations were observed in the case of Ca60-0.6-13 (Table 4). The calcium arc with a larger arc angle was heavily bent during the stent expansion.

The maximum principal strain in the noncalcium component of the plaque after stenting is shown in Fig. 8. The peak MPS occurred at the area adjacent to the calcium edges. They were 1.14, 1.40, and 2.29 for the cases with calcium angles of 60 deg, 180 deg, and 270 deg, respectively. It was reduced to 1.73 and 1.26 as the calcium thickness reduced to 0.4 mm and 0.2 mm, respectively. The abnormally high value of the MPS might be associated with the dissection in the noncalcium component. As the nonrecoverable plastic strain accumulated during expansion, the concave surface formed near calcium ends, which was consistent with the location of strut malapposition.

Discussion and Conclusion

The goal of this work was to characterize and clarify the mechanism of stenting in a calcified lesion. The influence of the calcium quantifications on acute stenting outcomes, in terms of the lumen gain, stent underexpansion, malapposition, and tissue mechanics, was delineated. The calcium was categorized by four commonly used clinical parameters (i.e., the arc angle, maximum thickness, length, and volume) [8,9]. Three independent calcium parameters, the arc angle, maximum thickness, and length, were varied to investigate their roles on stenting outcomes.

It was found that the lumen gain and its associated stent underexpansion were most sensitive to the calcium angle and next to the calcium thickness. A thicker calcium with a larger arc angle led to stent underexpansion as well as the clinically observed D-shaped lumen [27]. A small calcium angle, of 60 deg, had minimal influence on stent expansion. The desired lumen gain could be achieved with a nearly circular shape, which conformed to expanded stents. A reduced thickness in the calcium enhanced the lumen gain as well as propelled the lumen shape toward a nearly circular one (Fig. 4). This could be explained by the deformation of the rigid calcium as characterized in Table 4. For stenting a lesion with a thicker calcium, the calcium height was slightly reduced and the chord elongated by half due to its large structural rigidity. This was compensated by increasing the noncalcium height 3–4 times to accommodate stent expansion. It was credited to the better compliance and stretchability of the noncalcium component in comparison to the calcium. It was noted that the circumferential stretch ratios of the calcium and noncalcium components remained the same regardless of the calcium quantifications (Fig. 3(b)). The stretch ratio magnitude might be related to the balloon diameter at the full expansion state. Together, the commonly observed D-shaped lumen was formed for the heavily calcified plaque with a large arc angle. An increased compliance of the calcium arc, such as a small arc angle and thinner calcium, could revert the D-shaped lumen toward a nearly circular one. This implied the importance of pre-stent management including plaque modifications [28].

The noncalcium component of the plaque near the edges of the calcium arc was most stretched and exhibited a circumferential concave surface especially for a large calcium angle, which resulted in strut malapposition. Specifically, the peak strain region in the noncalcium component was consistent with the locations of

Fig. 6 Stress distribution in the artery at fully expansion of the stent (left) and after recoil (right). During stent expansion, the percentage of high stresses increased with a large calcium angle or a large thickness, while it exhibited an opposite trend after the recoil ($Ca_{\theta}-\delta_N-L_N$; θ represented the degree of calcium angle, δ_N represented the maximum thickness in mm, and L_N represented the longitudinal length in mm).

the maximum malapposition. Large circumferential stretch of tissues warranted well-apposed struts. This was evidenced in the case with the calcium angle of 60 deg regardless of its longitudinal length. The maximum gap of the malapposed strut always occurred at the MLA for the calcium with an arc angle larger than 180 deg. It was sensitive to both the calcium angle and thickness. The calcium length had minimal impact on the maximum gap of the malapposed strut, but it positively correlated with the area of the malapposed struts. This was consistent with the observations by Alegría-Barrero et al. [10].

The thicker calcium, especially with a large calcium angle, indeed played a protective role for the artery underneath, which agrees with previous observations [20,21]. However, this

protection effect caused by the calcium was at the expense of stress concentrations occurring around the edges of the calcium arc (Fig. 6). The peak arterial stress was located at the MLA plane at the fully expanded state, but it was shifted to the longitudinal ends of the calcium after recoil. In addition, a thicker calcium with a larger calcium angle exhibited abnormally high arterial stresses at the fully expanded state. After recoil, a smaller percentage area of higher stresses was observed. This was due to incomplete stent expansion. A large strain in the noncalcium component was associated with tissue damage, as observed by Fitzgerald et al. [29]. They demonstrated that tissue dissection occurred adjacent to the calcium edge, and the existence of the calcium would aggregate the dissection. It was also associated with cellular

Fig. 7 Strain histogram in the calcium at fully expansion (top) and after recoil (bottom). The peak MPS as well as the volume of higher strain in the calcium increased with a larger calcium angle and a thinner calcium ($\text{Ca}_\theta\text{-}\delta_N\text{-}L_N$: θ represented the degree of calcium angle, δ_N represented the maximum thickness in mm, and L_N represented the longitudinal length in mm).

mechanotransduction and tissue remodeling, including in-stent restenosis [15,30].

The calcium arc suffered larger strains in the cases with a large arc angle and/or a small thickness. For a large calcium angle, the peak tensile strain occurred at the inner surface of the calcium. It was attributed to the bending of the calcium arc, which was due to both the edge tension by the noncalcium component and the compression by the stent. The arc bending vanished for a small calcium angle of 60 deg, due to the minimal shape changes in the calcium during the stent expansion. The abnormally large strain in the calcium could induce a calcium fracture. Our results indicated that thinner calcium was prone to fracture. This was consistent with the suggestions for plaque modification in cases of severely calcified lesions [12].

The geometry of the artery and plaque components was idealized for this comparative study considering its advantage for inspecting the underlying mechanism of stent–artery interaction and for isolating the contribution of each parameter on stent expansion. On the other hand, our simulations do not represent all clinically relevant cases, and a patient-specific model would be useful for preclinical solutions. An OCT reconstructed plaque and an magnetic resonance imaging reconstructed artery might change the level of underexpansion and malapposition, as well as the stress and strain distributions of tissues. The material properties of artery and plaque components could be improved to be nonhomogeneous, anisotropic, and time dependent [31]. Nonhomogeneous calcium was associated with discontinuities in mechanical properties, which could lead to higher in situ stresses. A larger yield

Fig. 8 Strain distribution in the noncalcium component of the plaque, with the peak value at the interface between calcium and the noncalcium components ($\text{Ca}_\theta\text{-}\delta_N\text{-}L_N$: θ represented the degree of calcium angle, δ_N represented the maximum thickness in mm, and L_N represented the longitudinal length in mm)

stress of the noncalcium component was expected to induce less plastic deformation within the lesion, and thus less lumen gain and severe malapposition following the stenting. The aforementioned variations in material properties of noncalcium component could alter the magnitude of stent malapposition and stresses in the tissue, but not the location of stress concentration and malapposition induced by the material mismatch. For the stenting treatment of heavily calcified stenosis, a balloon predilation and/or a rotational atherectomy was generally needed for removing the lesion and/or inducing the calcium fracture [12]. Both predilation and calcium fracture were not considered in this work. The strain or stress distributions could change after considering the calcium fracture, but the major determinants of the stent expansion were expected to be the same (i.e., the calcium angle and thickness). In spite of these simplifications, this study demonstrated the basic mechanism of stenting in a heavily calcified lesion in terms of lumen formation, underexpansion, malapposition, and stress or strain distributions of tissue.

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