Available online at www.sciencedirect.com
ScienceDirect
Comput. Methods Appl. Mech. Engrg. 291 (2015) 42–63
Computer methods in applied mechanics and engineering
www.elsevier.com/locate/cma

Solid-beam finite element analysis of Nitinol stents

J. Frischkorn*,[1], S. Reese[1]

RWTH Aachen University, Institute of Applied Mechanics, Department of Civil Engineering, Mies-van-der-Rohe-Strasse 1, 52074 Aachen, Germany

Received 13 August 2014; received in revised form 16 March 2015; accepted 18 March 2015
Available online 1 April 2015

Abstract

This paper discusses the finite element (FE) modelling of Nitinol stent structures by means of a recently introduced solid-beam finite element technology. FE stenting simulations based on standard 3D solid elements are computationally very expensive. An adequate FE discretization of the stent structure requires a large number of elements for two reasons: Several elements are needed in the thickness directions of the stent struts in order to resolve localized martensite transformations. Additionally, due to the slenderness of the struts, a dense discretization in longitudinal direction is required to retain reasonable element aspect ratios. In combination with non-linearities emerging from the non-linear material behaviour, large deformations, and contact, stenting simulations become very challenging and time consuming. The solid-beam finite element technology is considered to reduce computational costs by working with only one element in thickness direction and allowing larger aspect ratios while the results are still acceptable to be used in the iterative design process. The here suggested formulation based on an eight-node brick element geometry is suitable to efficiently model beam-like structures with prismatic cross-sections without the necessity of abstracting the beam axes from the three-dimensional (3D) geometry. All relevant locking phenomena are alleviated by a tailored combination of different techniques. The combination of the new finite element technology with a sophisticated material model that represents the pseudoelastic behaviour of Nitinol is an additional aspect to be investigated in this regard. Validations at the example of an isolated strut show that very accurate results are obtained for thick and thin strut geometries. Moreover, we successfully simulate the crimping–expanding process of an intracranial stent showing a very complex geometry. The results match those based on high density solid meshes very well. The computational benefit becomes clearly evident.

Keywords: Finite element technology; Beam elements; Stent implantation; Pseudoelasticity

1. Introduction

Stents are tubular medical devices which are implanted into the human body to treat a variety of vessel diseases like arteriosclerosis or aneurysms. They are applied in various regions of the body: e.g. coronary, carotid, iliac, femoral or even intracranial arteries. Main objectives are the recovery of blood flow in stenotic arteries or the stabilization of aneurysms. Moreover stent-like structures serve as scaffolds in artificial heart valves.

* Corresponding author.
E-mail addresses: jan.frischkorn@rwth-aachen.de (J. Frischkorn), stefanie.reese@rwth-aachen.de (S. Reese).
[1] Tel.: +49 0 241 80 25001; fax: +49 0 241 80 22001.

http://dx.doi.org/10.1016/j.cma.2015.03.011
0045-7825/

Using minimally invasive surgery, the stent is applied in its crimped state by a catheter (e.g. percutaneously through the femoral artery). After the required position is reached the stent is released from the catheter. Subsequently, the stent's mode of operation is usually based on a strong diameter expansion such that the blood vessel is held open. Depending on the expanding mechanism, balloon-expandable (BE) and self-expandable (SE) stents are distinguished. The vast majority of the BE stents is made of stainless steel 316L while nickel–titanium alloys (Nitinol), also known as shape-memory alloys (SMAs), are in general the materials of choice for SE stents. In this paper we focus on the latter. Nowadays, most stents are manufactured by laser cutting from tubes. The resulting topology is characterized by beam-like structures (often with rectangular cross-sections) which are connected by knots. A survey of the variety of stent designs is provided in [1].

Although there are some stent applications that utilize the temperature driven shape memory effect for the expanding or recoiling process, e.g. the vena cava filter of Simon et al. [2], the majority of SE stents exploits the pseudoelastic or superelastic behaviour of Nitinol. It is based on a stress induced phase transformation from an austenitic (A) to a martensitic (M) crystallographic structure and a back transformation during unloading. Different transformation stresses for the $A \rightarrow M$ and the $M \rightarrow A$ transformation lead to a stress–strain hysteresis. Thereby, strain magnitudes of up to about ten per cent can be recovered during unloading. Compared with BE stents several advantages of Nitinol stents can be named (cf. [3,4] for details). In contrast to the BE stents which are manufactured in their crimped configuration (smallest diameter), the diameter of an undeformed SE stent is approximately ten per cent larger than the vessel diameter. As a consequence no over-expansion is needed to account for spring back which in turn reduces the damage to the vessel. Besides good biocompatibility (passive Ti-oxide layer) and high kink resistance, Nitinol stents provide a very good biomechanical compatibility: The hysteretic stress–strain behaviour, also characteristic for many biological tissues, provides low opening forces (chronic outward force) while a high resistance (radial resistive force) against diameter decrease is obtained (see e.g. [5,6]).

In several publications the finite element (FE) method is applied to investigate the stenting procedure on a computational basis. Most of the simulations are performed by means of the commercial software packages ANSYS or ABAQUS using the pseudoelastic material formulation of Auricchio and Taylor [7]. For realistic simulations several challenges need to be accomplished: non-linearities emerging from large deformations, pseudoelasticity and massive contact interaction between stent and artery. The complex interaction between the stent and the soft tissue of the artery can currently be treated in the best way by explicit FE methods. In simplified simulations, where the artery is for instance represented by a rigid surface or shell elements, implicit methods are well applicable, too. Explicit stenting simulations based on patient specific data of a carotid artery can be found in [8,9]. Wu et al. [10] utilize the implicit FE method for similar simulations. Implicit FE simulations are also applied by Thériault et al. [11] to analyse a commercial Nitinol stent which is equipped by surrounding polyethylene rings. They show that the relaxation behaviour of the polyethylene rings can be used to steer the speed of the stent expansion with the aim to reduce tissue trauma. Azaouzi et al. [12] simulate the deployment and the pulsatile loading of a Nitinol stent inside an artery. In Azaouzi et al. [13] an optimization procedure is used to reduce the mean stress and the stress amplitude to which the stent is exposed during long-term cyclic pulsation due to heart beating. Kleinstreuer et al. [14] analyse stent grafts under cyclic loading for abdominal aortic aneurysms. Combined experimental and FE-based evaluations of pulsatile fatigue of diamond-shaped Nitinol specimens are performed by Pelton et al. [15].

In addition to the non-linearities, FE simulations of stent structures usually require high-density solid meshes to obtain converged results that account for localized martensite transformations. Fine meshes are also needed for adequate resolution of the artery in order to evaluate the impact of contact stresses accurately. Thus, the computational effort of realistic stenting simulations can become very high. Although most regions of the considered stents are characterized by beam-like parts the application of conventional beam elements is difficult (except for stents made of braided wires see e.g. [16]). In this work we investigate the applicability of a recently introduced solid-beam finite element formulation (Q1STb) of Frischkorn and Reese [17] towards the modelling of stent structures with the aim to reduce the computational effort. For this purpose the material model of Reese and Christ [18] is considered to represent the pseudoelastic behaviour. The solid-beam element is built upon an eight-node hexahedral element. Special technologies are incorporated to avoid various locking effects that are inherently included in the linear kinematics. Inspired by the solid-shell formulation of Schwarze and Reese [19] all relevant locking effects are removed by a tailored combination of the enhanced assumed strain method (EAS) and the assumed natural strain method (ANS) which are embedded in a reduced integration concept. The material non-linearities that evolve in the thickness directions are taken into account by using several quadrature points in thickness integration.