Is a Standardized Measurement of the Elastic Recoil of Coronary Stents under Vascular Conditions Necessary and Meaningful?

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Summary

Laboratory testing methods for coronary stents should be as close to real conditions as possible. This study therefore investigates the possibility of a mechanical simulation of the vessel surrounding the stent. As an issue of main importance, the interest is focused upon measurements of the elastic recoil after expansion and of the radial resistance under external pressure loads for a balloon-expandable stent, using three different tubes as vascular models. Comparative measurements without a vascular simulation have also been performed. The reproducibility of results as well as systematic changes induced by the vascular simulation are discussed, as are the opportunities and limitations provided by a vessel model adapted to physiologic conditions. Currently, recoil measurements in a water bath at 37 °C without vascular simulation are regarded as the only method of gaining objective results that can be compared between individual laboratories. Although presently there is no way to model the elastic properties of actual — even diseased — blood vessels by means of single standardized test conditions that would provide the same degree of objectivity, vessel simulations may help to learn about the qualitative differences to be expected between in vivo conditions and the present stent-alone experiments.

Key Words

Stent, bench testing, mechanical properties, vascular simulation

Introduction

Testing procedures for the properties of medical products in the laboratory should always be guided by the intended clinical application. Even though national and international standards do not require a measurement under vascular conditions [1], the blood vessel is the application location for stents. This justifies attempts to include the vessel into the evaluation of stent functionality.

Physiologic models are at a disadvantage if they are too complex and, therefore, difficult to reproduce, if they do not permit a separation of the influence factors, or if the definition of the "physiologic conditions" itself is controversial. For example, charge-dependent or age-induced parameter fluctuations of the materials used that influence the elasticity or the compliance of the vascular model are problematic in this connection. As a consequence, even correctly measured physical parameters often cannot be compared with the results of measurements by other laboratories. Such measurements are not suited for standardization.

Study Purpose

This study examined the parameter values for the *elastic recoil after expansion and the radial resistance under external pressure loads*, which have to be determined for balloon-expandable stents according to EN 12006-3, under the following aspects:

- Is it possible to measure the elastic recoil in the vascular model in a reproducible way?
- Are the results comparable to those from an expansion without a vascular model?
- What is the influence of the material properties of the vascular model on the pressure-diameter curve when measuring the radial resistance by exerting external pressure?

Material and Methods

It has been shown that the parameters to be determined, recoil and radial resistance, strongly depend on the material, design, and dimensions of the stent [2-4]. To study only the influence of the vascular model, a typical slotted-tube design of a coronary stent with defined dimensions was examined in different vascular models for this study.

The coronary stent Tenax (Biotronik, Germany) with a length of 15 mm was used for the comparisons (Figure 1). It consists of 6 segments that are longitudinally connected to each other by two links each. The studied Tenax exemplars were dilated to diameter 3.0 mm (15/3.0) and 3.5 mm (15/3.5), respectively.

Three different tubes were used as vascular models:

- A silicone tube with an inner diameter of 3.0 mm and a wall thickness of 0.5 mm for the Tenax 15/3.0.
- A silicone tube with an inner diameter of 3.5 mm and a wall thickness of 0.5 mm for the Tenax 15/3.5.
- A polyurethane (PUR) tube (Pellethane) with an inner diameter of 3.5 mm and a wall thickness of 0.075 mm for the Tenax 15/3.5.

The inner diameter of the tubes that were used equals the standard diameter of the studied stent system as required by the delivery system.

According to the literature, the elastic modulus of silicone (Nusil Silicone MED 2231) is E_{200} = 1.3 MPa at room temperature [5][1] . We confirmed this value range by measurements on the tube material in a tensile test (E_{200} = 1.56 ± 0.07 MPa) [7]. A temperature rise to 37 °C will decrease the elastic modulus.

The PUR tube is made of Pellethane 2363-80A. Literature states E_{100} = 6.1 MPa at room temperature [5]. Our own measurements, which took into account the processing technology and an environmental temperature of 37 °C, resulted in the characteristic values E_{100} = 4.0 ± 0.1 MPa and E_{200} = 5.9 ± 0.2 MPa, respectively [8].

For the stent measurement with the vascular model and either with or without an additional external load, a pressure-diameter curve was recorded with a stepwise increase of the external pressure and subsequent load relief after each step; in addition, the collapse pressure was determined.

[1] E_{200} and E_{100} denote the elastic modulus determined at Dl/l = 200 % and 100 % relative elongation, respectively [6].

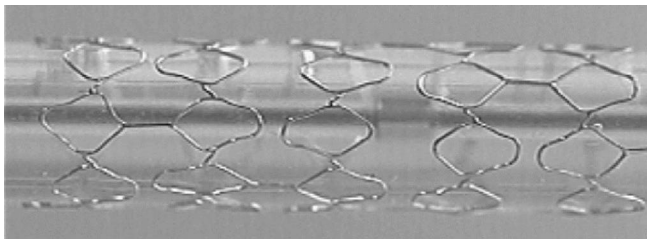

Figure 1. Tenax 15/3.5, expanded to 3.5 mm (detail).

The measurements were performed with the test set-up shown in Figure 2 [3].

The test chamber, located in the center of a biaxial laser measuring-head, is pressure sealed with a cover plate and connected to the computer-controlled pressure unit [3] through a system of tubes. Due to the measuring technique, the measured diameters are to be understood as stent diameters plus the surrounding tube material.

The stent to be tested is surrounded by a tube that simulates the vessel and separates the stent from the water-filled (37 °C) test chamber, thus allowing the application of the radial load.

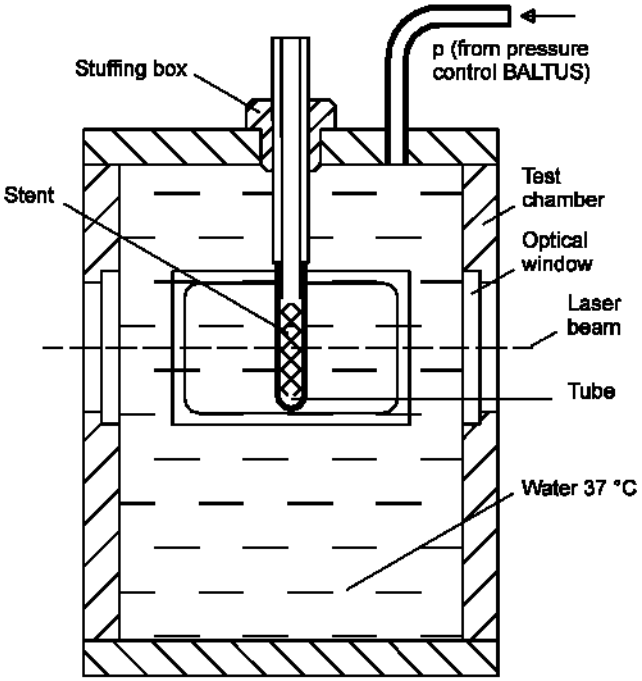

Figure 2. Test chamber for measuring the radial resistance of coronary stents under external pressure loads.

After insertion of the complete system into the vascular model, the stent was expanded by applying a pressure of 8 bar. In this state, the outer diameter d_{pmax} of the stent plus tube was measured on a segment in the middle of the stent.
Subsequently, the balloon was deflated, retracted, and the diameter d_{p0} was measured at the same position. Following the measuring regime, the pressure in the test chamber (= external pressure for stent and tube) was now raised to a peak value p_i that was increased in each step i by $\Delta p = 0.1$ bar. Next, the diameter d_{pi} was measured, and then the chamber was relieved of the pressure load again ($p = 0$). In this manner, the pressure was increased until the stent collapsed.
The diameter d_{pmax} was entered into the diagram as the first value. This value is larger than the original tube diameter (vessel overdilation) due to the inflating balloon at 8 bar. The second value in the diagram is the diameter d_{p0} after deflation and retraction of the balloon.
The elastic recoil R_0 without an additional external load (which means that only the recoil forces inherent to the stent plus tube combination are effective when using a vascular model) can be calculated as:

$$R_0 = \frac{d_{p\,max} - d_{p\,0}}{d_{p\,max}} \times 100\ \%$$

Furthermore, for the recoil R_A under an additional maximal external load, the diameter d_{p1} was used, which is defined as the diameter under maximal external pressure p_1 without stent collapse. Applicable is:

$$p_1 = p_{coll} - \Delta p$$

$$R_A = \frac{d_{p\,max} - d_{p1}}{d_{p\,max}} \times 100\ \%$$

The collapse pressure p_{coll} was determined as the external pressure value under which the support function of the stent was no longer guaranteed. This point could be easily recognized by direct observation (video). Δp was constant for all tests ($\Delta p = 0.1$ bar).
For comparative purposes, another six Tenax 15/3.5 exemplars were expanded in a controlled-temperature water bath without a surrounding vascular model and relieved subsequently, allowing a measurement of the recoil without an external load.

Results

As examples, the pressure-diameter curves for measurements in silicone tubes (Figure 3) and in the PUR tube (Figure 4) are depicted. Both figures show the measured stent diameters during expansion with the balloon pressure of 8 bar (d_{pmax}). Thus, the recoil in the simulated vessel could be calculated immediately by comparison with the diameter after pressure relief d_{p0}.
The recoil is generated from the elastic recoil of the stent after expansion and subsequent load relief, and from an additional deformation due to the restoring forces of the tube material on the stent. However, the results indicate that an additional "combined rigidity" of the tube-stent system is created from the interaction of the vascular model and the stent structure. This "combined rigidity" results from the impairment of the recoil by the friction between tube and stent filaments. Generally, a decrease of the total diameter of tube and stent is observed with increasing external pressure. The horizontal lines in the diagrams (at 4.5 mm and 3.65 mm, respectively) indicate the nominal diameters that are expected for tubes without an implanted stent. These values can be calculated from the inner diameter plus twice the wall thickness. Measurements of these reference diameters with the laser scanner were not feasible, because the method relies on that the cross section is circular, which could not be guaranteed by the tubes in this study. However, the nominal outer diameter may well serve as a guide to assess stent deformation under and after an external hydraulic pressure load.
The measuring regime was always followed until the stent collapsed, but valid determinations of the diameter were possible only as long as the circular cross section of the tube and stent was maintained. Therefore, the diagrams in Figure 3 and Figure 4 show all measured results up to the pressure.

$$p_1 = p_{coll} - \Delta p$$

An overview of the characteristic diameter values obtained is given in Table 1.
The small number of specimens tested with the surrounding vascular model does not support statistically based estimates of significance. However, the good agreement between the two measurements each that were made under the same conditions, allows at least to indicate general trends.

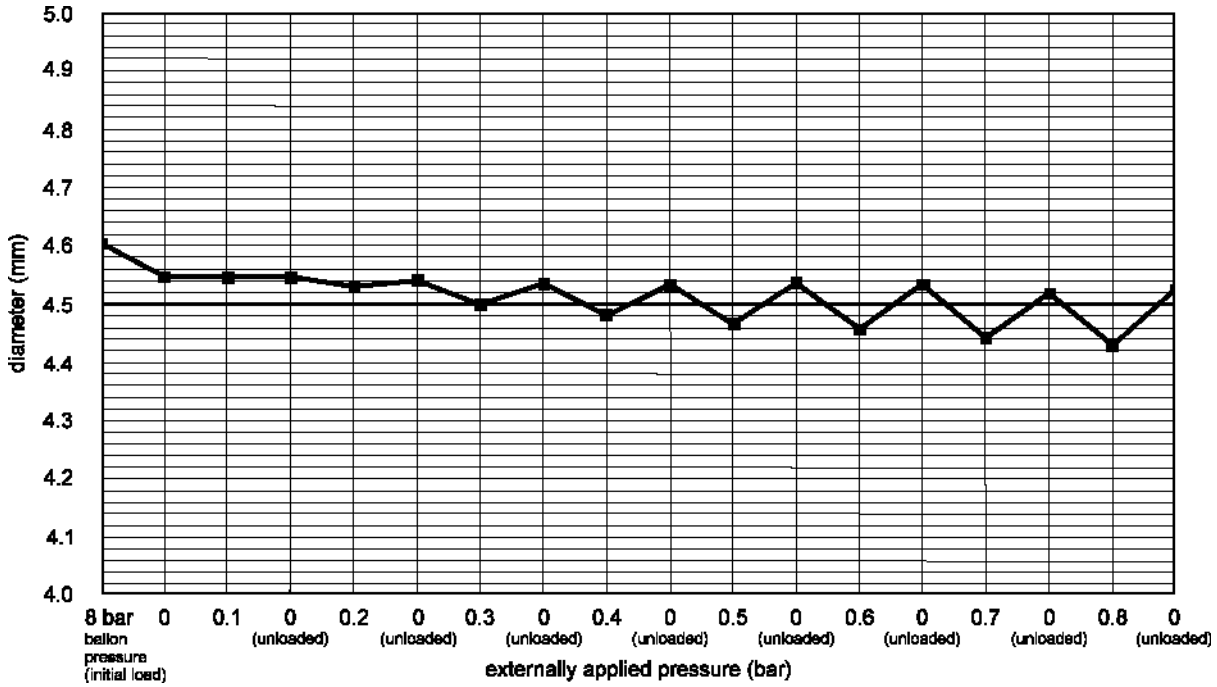

Figure 3. Pressure-diameter curve of the Tenax 15/3.5 stent during and after expansion, as well as under increasing external pressure load and subsequent pressure relief in the silicone tube (I.D. = 3.5 mm, s = 0.5 mm).

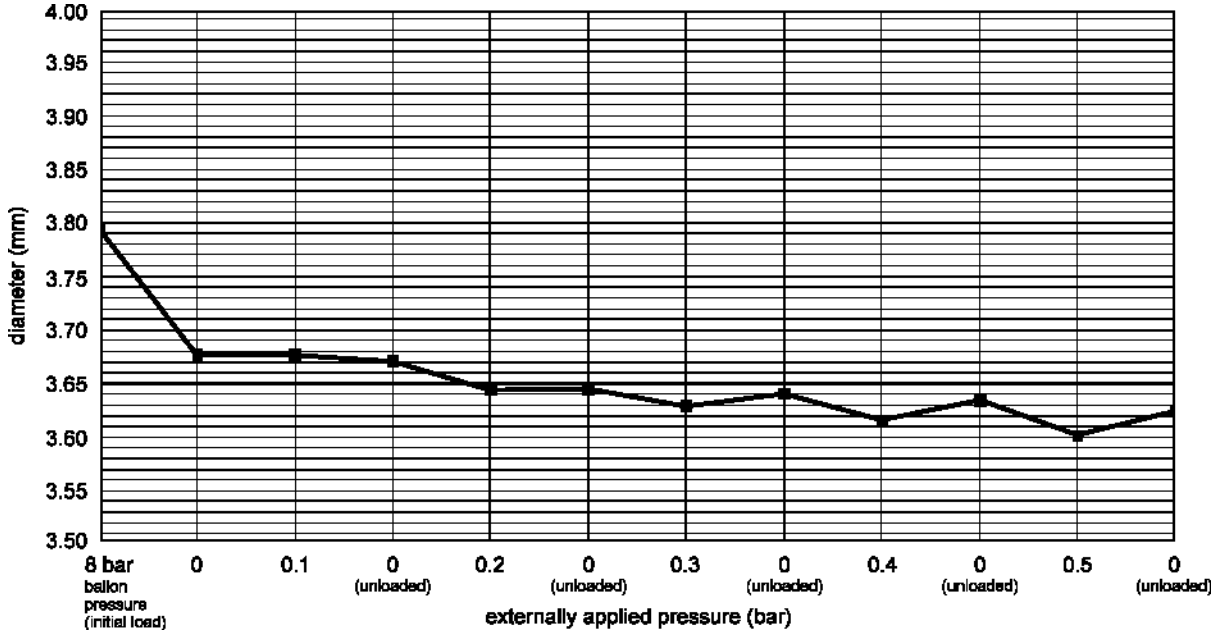

Figure 4. Pressure-diameter curve of the Tenax 15/3.5 stent during and after expansion, as well as under increasing external pressure load and subsequent relief in the PUR tube (I.D. = 3.5 mm, s = 0.075 mm).

Discussion

Recoil without Vascular Simulation

The measured recoil R_0 was determined for all six tested Tenax 15/3.5 systems without vascular simulation. The mean value of these results was R_0 = 3.78 % (Table 1).

		Diameter d_{pmax} (8bar)	Diameter d_{p0}	Recoil R_0	Diameter d_{p1}	Recoil R_A at maximal external pressure
		[mm]	[mm]	[%]	[mm]	[%]
Measurement in the silicone tube, wall thickness 0.5 mm	Tenax Complete 15/3.0	4.192	4.137	1.31	4.002	4.19
	Tenax Complete 15/3.5	4.617	4.555	1.34	4.439	3.85
Measurement in the PUR tube, wall thickness 0.075 mm	Tenax Complete 15/3.5	3.790	3.663	3.36	3.590	5.28
Free expansion (without vascular model)	Tenax Complete 15/3.5	3.577	3.442	3.78	-	-

Table 1. Diameter values as determined under the various test conditions (mean values).

This result is independent of any model assumptions and is intended to serve as a comparative value for the studied vascular models. With $n = 6$, the sample size is not large, yet it is sufficient in comparison to the other tests.

Recoil Measurement with Vascular Simulation and without External Pressure Load

Mean values of $R_0 = 1.31$ % (3.0 mm) and $R_0 = 1.33$ % (3.5 mm), respectively, were determined in the silicone tube (Table 1). These values are clearly lower than the comparative values without the vascular model. The recoil in the "biegeschlaff" PUR tube ($R_0 = 3.36$ % for 3.5 mm standard diameter) is bigger than in the silicone tube and much closer to the recoil without the surrounding tube, but still lower than without a surrounding vascular model (Table 1).

After expansion, the stent filaments are supported by the tube in a circumferential direction, but mostly in axial direction (longitudinal axis). As a consequence of the friction between the filaments and the inner tube wall, the recoil is impaired. Due to the different surface roughness and friction coefficients of the PUR and silicone materials against the metal stent (a-SiC coating layer on the 316L steel), this effect is also responsible for differences between the silicone and PUR measurements. Additionally, the thicker wall of the silicone tube (s = 0.5 mm) entails a self-supporting effect at zero external load, which will not necessarily impede a larger recoil of the stent inside the tube, but will inhibit detection of such behavior.

Recoil Measurement with Vascular Simulation and External Pressure Load

In case of the applied external pressures, the measured recoil values R_A increased only slightly. Even under the maximal external pressure p_1 that could be exerted without collapsing the stent, the decrease in diameter R_A was still less than 6 %. Larger recoils could never be observed.

Furthermore, it was observed that the stent deformation under external pressure was elastic for the measurements in silicone tubes, i.e., the initial diameter was reached again after pressure relief. A slight plastic deformation (0.035 to 0.050 mm after 0.5 bar external pressure) was measured in the PUR tube. This corresponds to 1.0 to 1.4 % of the initial diameter at 0 bar. One has to consider here that a higher radial force is required with the silicone tube in order to achieve the same deformation of the stent, because at 0.5 mm wall thickness the tube has a supporting effect by itself. This can be quantified by the fact that the silicone tube without stent collapses under external pressure only at $p_{coll} = 0.3$ bar. Thus, a plastic deformation decreasing the stent diameter due to an external pressure can no longer be precisely determined via the outer diameter of this tube.

Use of Tubes for Vascular Simulation

Can the properties of the tubes that are used be compared with those of real arterial walls? One parameter could be the elastic modulus that determines, among others, the vessel compliance under pressure loads or,

vice versa, also the radial load of the vessel on the stent through an overly dilated vessel. The literature states values for the elastic modulus of arterial vessels in an order of E = 2.0 - 8.0 x 10^6 dyn / cm² (0.2 - 0.8 MPa) [9]. Furthermore, a pressure-dilatation elastic modulus E_p is defined that describes the vessel compliance:

$$E_p = \frac{d\Delta p}{\Delta d}$$

with d .. mean (static) vascular diameter
Δp .. pressure change
Δd .. diameter change

For canine coronary vessels, a value of E_p = 3.41 x 106 dyn/cm² = 0.341 MPa is given [10]. From the material data of the vessel models, the compliance moduli E_p can be calculated via

$$E_p = E \frac{2s}{d}$$

with E .. elastic modulus
s .. thickness of the tube wall
d .. mean (static) vascular diameter .

For the d = 3.5 mm, s = 0.5 mm silicone tube one finds E_p = 0.45 MPa, while for the d = 3.5 mm, s = 0.075 mm PUR tube E_p = 0.17 - 0.25 MPa results. These values are well within physiologic boundaries, which is further supported by elastic moduli that have been determined separately for pure elastin (0.5 MPa) and active smooth muscle cells (0.8 MPa), respectively. The meshwork of collagen fibers in the adventitia would offer far more resistance to dilatation, but takes no effect here since it is gradually pulled tight not until about 30 % diameter change [11,12,13].

The compliance of a coronary vessel is therefore well simulated by both tube materials, if one accepts a single general elasticity value for a real artery. However, this is not a trivial issue, because the mechanical properties of normal vessels and vessels with stenoses are hardly comparable [14,15]. The friction behavior between vascular wall and stent filaments is also of great importance, but quantitative data are not yet available. The differences in the results between PUR and silicone tubes at least partially relate to such friction differences.

The inner surface of the lumen of normal and pathologic vessels may vary dramatically depending on the stage of atherosclerotic disease, as well as on location, concentric or eccentric cross section, and length of the stenosis. Soft plaques with fibrous inclusions or calcified rigid plaques will determine the vessel compliance, as will the friction between the stent and the vessel wall. Thus, a vascular simulation using a standardized average friction coefficient does not adequately describe the in vivo situation. Nevertheless, stents are predominantly applied to vessels with pathologic changes!

The current standard EN 12006-3 [1] does not require external loads for recoil measurements. Thus, the stent reaction to an external radial load is not captured with the recoil parameter. However, this important stent property is determined by measuring the radial resistance (pressure-diameter behavior and collapse pressure).

It is difficult to standardize the use of a vascular model because of the complicated additional effects (compliance, surface, and self-support), thus complicating the comparison of the measured values. But especially the stent diameter, a quantity that can be easily measured as a function of the balloon pressure, as well as the parameters that can be derived from it for the delivery system and the stent, should also be determined under truly standardized conditions since product characteristics can be objectively evaluated on this basis. In our opinion, a recoil measurement in a water bath at 37 °C without vascular simulation is therefore presently the only method to gain objective measurement values that can be compared quantitatively between individual laboratories.

On the other hand, the parameter range of elastical characteristics typical of physiologic or pathologic conditions is well accessible by a proper choice of tube material and geometry even with the very simple vessel simulation described in this article. The studied Tenax stent can be regarded as a typical slotted tube stent. The knowledge concerning the measuring procedure and the behavior of stents in simulated vessels that was gained with this study can also be applied to other stent types, even though the described effects, e.g., of the wall friction and the associated support function, will be of differing strength due to different structures.

Moreover, the model can obviously be extended with respect to inhomogeneously distributed elastic properties, e.g., by covering part of the inner tube surface with a rigid material to mimic calcifications. This

opens a way to learn at least qualitatively about the kind of effects to be encountered when advancing beyond the regime of current stent-alone experiments. In this way, basic mechanisms that are specific to the interaction of the stent and the surrounding vessel can be identified (and may even be quantified) under the well-defined model conditions rather than having to be extracted and isolated laboriously from observations under poorly known in vivo conditions. Vascular simulation might thus help to study further, and more complex, properties for blood vessel stents, and allow to develop strategies concerning, for example, ease of handling and product safety. However, a test standard that would allow quantification of these qualitative results does not yet exist. Such a standard can be developed only from a better knowledge of the mechanical parameters of a selection of "typical" lesion sites, which remains to be gained in future investigations.

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