

Development Options for Coronary Stent Coatings

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Abstract

In our research we investigated the effect of different surface qualities on the adhesion of polylactic acid coating on 316L steel. During the study, samples were electropolished and surface treated with aqua regia and Vilella reagent, and the coating was applied by dipping technology using a Yaskawa robot arm. Tribology tests were carried out to determine the adhesion, in addition to the force and friction conditions, the coating damage resulting from the measurements was also recorded. In addition to the coating adhesion assay, contact angle measurements were also performed. Based on our measurement results, it was found that compared to the three different surface treatment methods, the surface of the samples treated with Vilella was more adherent to the polymer than to the conventional electropolished surface. In this case, the frictional force was also much lower than that of the polished specimens, therefore this surface treatment method may be suitable for improving the adhesion of the coating.

Keywords: coronary stent, surface treatment, coating adhesion, PLA.

1. Introduction

Coatings were initially used to enhance the biocompatibility of stents, but nowadays their primary goal is the controlled delivery of drugs used to inhibit neointimal hyperplasia. Regarding the optimal stent coating, the literature highlights three main aspects. The first and one of the most important factors is that the coating must be done on a dirt-free metal surface. Second, the surface roughness should be minimal relative to the thickness of the coating to ensure good uniformity and adhesion of the film layer. Furthermore, finally, the coating must be stable both when applied to the stent and after implantation. [1, 2]

The three drugs studied for the treatment of restenosis are heparin, sirolimus, and paclitaxel. Heparin was effective in reducing both thrombosis and neointimal proliferation, while sirolimus and paclitaxel were mainly used in their antiproliferative effects to block neointimal hyperplasia. [3, 4]

Proper surface treatment plays an important role in the design and manufacture of a stent.

Roughening of the surface may increase the adhesion of the coating, but a surface that is too rough tends to thrombogenicity. A typical surface treatment method for stent types used in clinical practice is electropolishing. [5, 6]

In our previous research, we observed that the coating may detach from the surface of the stent already during implantation, due to the fact that the plastic coating carrying the drug does not adhere properly to the surface of the metal frame of the support. [7] The aim of the present research is to develop a surface treatment method to improve the adhesion of the polylactic acid coating on 316L type austenitic stainless steel, taking into account previous experience [8].

2. Methods

For the research, 30×30×0.3 mm, 316L type, austenitic stainless steel inserts were used. 5 specimens were electropolished, 5 specimens in aqua regia (20 mL methanol, 15 mL 65 % HNO₃, 45 mL 37 % HCl, up to 60 s, without power source), and 5 in Vilella reagent (200 mL ethanol, 2 g picric

acid, 10 mL of 37 % HCl, up to 60s, at 5V). Surface treatments were performed at 30 ± 1 °C. Furthermore, the samples were cleaned with acetone and ethanol according to the pre-programmed values in the robot before and after the surface modification, and then allowed to dry (3× acetone and 3× ethanol purification, respectively).

2.1. Electropolishing

Of the various surface treatment methods, electropolishing is one of the most popular surface modification methods due to its many advantages. One of these is that it can be performed on complex shapes, such as coronary stents of different shapes and geometries. It can be used to create a well-passivated, smooth, even and clean surface. During the polishing process, the metal is oxidized on the surface of the anode, causing the ions to dissolve into the electrolyte solution and then migrate to the cathode where they are reduced to form hydrogen. The electropolishing process is influenced by a number of parameters. Such parameters include anodic current density, applied potential, electrolyte temperature, polishing time, electrolyte composition and concentration, mixing method, and anode-cathode surface ratio. [9-12] Based on this research and our experience, electropolishing was performed at 5 V and 0.01 A/mm² in room temperature solution

(25 ± 1 °C) with ultrasonic vibration, phosphoric acid (H₃PO₄), sulfuric acid (H₂SO₄) and distilled water. It was performed in a 6:3:1 mixture to which 20 mL of glycerol was mixed. The polishing time was 3 minutes. (Figure 1).

2.2. Chemical etching

Electropolishing can create a very smooth surface, but the adhesion of the coating can be aided by roughening the surface by chemical etching. Similar experiments have already been carried out in the development of polyurethane coatings, in which it has already been demonstrated that the smaller the roughness of the stent surface, the easier it is for the coating to peel off. [13] However, polyurethane is no longer a typical drug carrier in medical technology, so taking into account the research and development trends, the present research examines the adhesion of polylactic acid.

Roughening of the surface was solved by chemical etching. In order to make the process repeatable, the process was automated with a robot arm (Yaskawa You teach me). The milling agent was stirred continuously with a magnetic stirrer (IKA RCT basic) at 30 ± 1 °C. The duration of the surface treatment was 60 seconds, during which the robot arm moved the specimen in a circular manner in the solution (Figure 2).



Figure 1. Electropolishing process.



Figure 2. Surface treatment with a Yaskawa „You teach me” robotic arm.

2.3. Coating process

The coating was also applied to the surface-treated tiles automated with the robotic arm. The polylactic acid was dissolved in chloroform. The dipping time was 10 seconds, and after extraction, the coating formed on the tile was dried by rotating it along its axis.

2.4. Qualification of the coatings

The weight of the tiles was measured before, after and after the surface treatments. The effectiveness of the surface treatment was examined by electron microscopy (Zeiss EVO MA 10) and confocal microscopy (Sensofar, neox PLu). Surface roughness was also measured on reference, surface-treated and coated samples. The adhesion of the coating was evaluated by a tribological test (High Temperature Tribometer - CSM Switzerland). As the moisture-binding effect of a given plastic is important in medical technology, we also performed edge angle measurements as part of the research.

3. Results

3.1. Microstructure examination

The surface treatment methods used resulted in different surface morphologies, the scanning electron micrograph (SEM) of which is shown in [Figures 3–5](#). Electropolishing, as expected, gave a smooth surface, while for samples milled with Vilella and aqua regia, the surface roughening was achieved. Each of the surfaces obtained after the treatments may be suitable for applying the polymer coating of our choice.

3.2. Weight measurement

The largest weight loss was caused by the electropolishing process (6–8 %), followed by the weight loss caused by aqua regia (0.5–1 %) and Vilella etchant (0.2–0.3 %).

After coating preparation, the weight of the inserts increased uniformly by 6 to 7 mg, which supports the success of the automation of the coating.

3.3. Surface roughness

Surface roughness was determined by confocal microscopy (Neox Flu, Sensofar). The surface roughness of the samples etched in the royal water was the highest, of the treated samples only the roughness less than the reference value was measured on the electropolished specimen ([Table 1](#)).

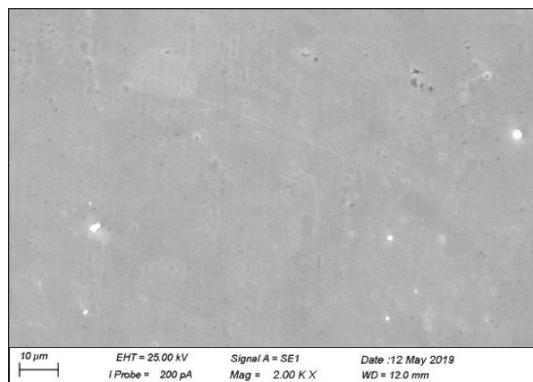


Figure 3. Electron microscopic images of the surface of the electropolished 316L sheet.

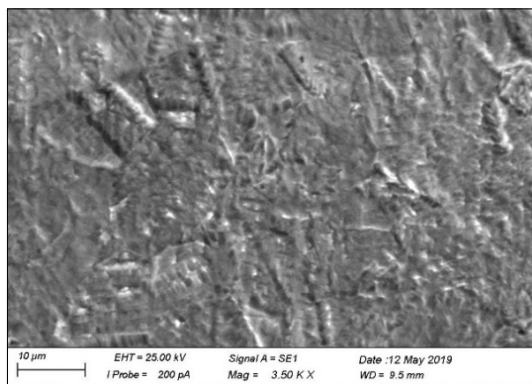


Figure 4. Electron microscopic images of the surface of the aqua regia etched 316L sheet.

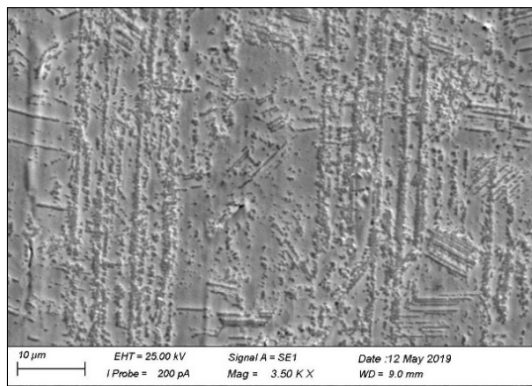


Figure 5. Electron microscopic images of the surface of the Vilella etched 316L sheet.

The values after coating preparation are summarized in [Table 2](#). After coating, the average surface roughness of each sample decreased, which is advantageous because the low surface roughness during implantation reduces the force required to push the implant intravascularly.

Table 1. Surface roughness measured on the samples (*Sa* = average of the deviations). The shown values are the average values of the 5 samples.

Sample	Sa (μm)
Aqua regia etched	5.12
Vilella etched	4.53
Electropolished	2.66
Reference	2.73

Table 2. Surface roughness measured on the samples after the coating process (*Sa* = average of the deviations). The shown values are the average values of the 4 samples.

Sample	Sa (μm)
Aqua regia etched	1.33
Vilella etched	1.37
Electropolished	1.53
Reference	1.30

3.4. Coating thickness

The thickness of the coating was also determined by confocal microscopy. To this end, samples were selected from all types in which a part of the coating was mechanically removed from half of the coated section so that neither the raw material nor the remaining coating section was damaged; partial layer thickness could not be measured properly at the original boundary, as it would show a gradual thinning. By examining the total length of the boundary section thus formed (marked in yellow in **Figure 6**), the coating thickness is obtained by calculating the difference between the average level of the sample surface (green area in **Figure 6**) and the average coating thickness level (orange area in **Figure 6**). Unfortunately, the robotic arm method we currently use did not allow the samples to be dipped in their full length into the polymer, as the chuck would have been immersed together with the sample in the plastic solution. The polylactic acid coating would solidify on the chuck and insert together, and even the coating on the insert would be damaged as the chuck was removed. In the future, our goal is to further develop the current method in such a way that we are able to immerse the samples in polylactic acid along their entire length so that the layer thickness is as uniform as possible over the entire length of the sample.

The software of the applied confocal microscope made it possible to determine the exact height

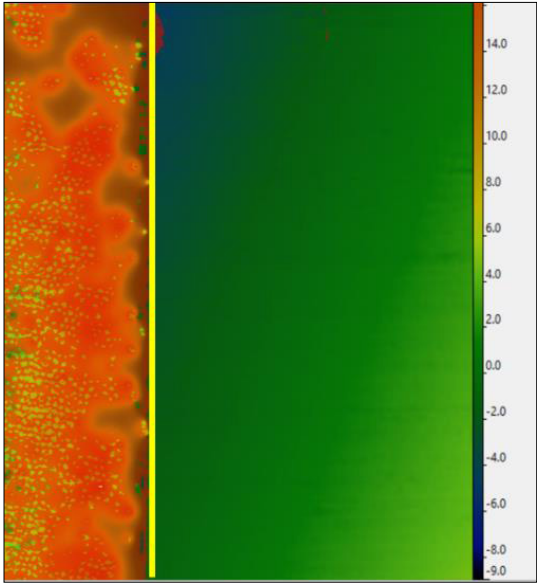


Figure 6. Colour scale for determining layer thickness. The border section created during the incision is marked in yellow. The scale on the right shows the depth and height of the levels for each colour in μm.

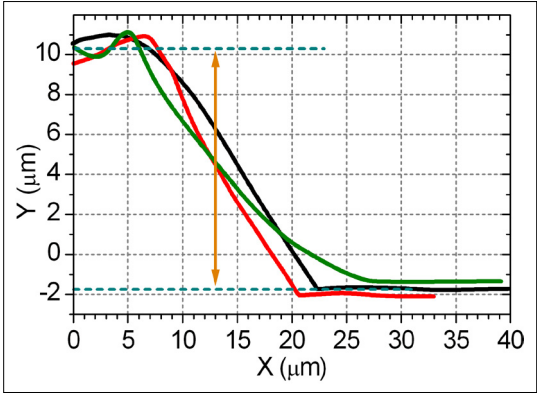


Figure 7. Results of measurements with confocal microscopy; the yellow arrow-line shows the layer thickness.

and depth values at the selected points as well as along the lines, in addition to the colour map representation. **Figure 7** shows the values of 3 line analysis. The measurement lines were recorded in a direction perpendicular to the boundary line. The average height of the coating layer as well as the average height of the plane of the plate were determined, and the thickness of the coating was obtained from the difference between the two. Based on the results, the average thickness of the coating prepared by us is 10.45 μm, which meets

the values of the literature and is almost identical to the typical coating thickness of the stents currently on the market.

3.5. Qualification of coating adhesion

The adhesion quality of the coating was determined by tribological testing, for which we used the High Temperature Tribometer CSM THT instrument in the laboratories of PROMATECH in Kosice. During the test, using the same load force, in our case 50 N, a needle on a disc was moved circularly on the surface of the insert. A detail of the “ditch” formed by the needle is shown in **Figure 8**.

Based on the experiment, we found that under the effect of the same loading force, the trench with the smallest depth (8.76 μm) was formed in the case of the samples treated with Vilella. Based on this, it can be said that the applied polymer coating was here the most resistant to the same load. In this case, the needle did not even reach the boundary of the metallic material of the sample, based on an average coating thickness of 10.45 μm . In contrast, there was a deeper difference beyond the coating thickness for the aqua regia-treated (17.5 μm) and electro polished (16.9 μm) samples. The deepest scratch depth (35.2 μm) was obtained for untreated but polymer-coated samples, i.e. it was the least “resistant” to the same load.

3.6. Edge angle

When measuring the edge angle, the wettability of the inserts was examined. From the data, we can deduce whether the blood wets the surface or not. This is important because the hydrophobic or hydrophilic nature of the surface can affect the effect of the coating on restenosis. Smaller flange angles means better wetting (minimum 0°), while larger flange angles (maximum 180°) can mean worse wetting. At a flange angle of less than 90°, it is said that the liquid wets the solid surface of the substrate, and at flange angles greater than 90°, we speak of non-wetting. Our measurement results are summarized in **Figure 9**.

From **Figure 9**, it can be seen that the polymer applied to the surface of the Vilella-treated samples has a wettability above 90°, i.e. the coating is more hydrophobic, even in the case of electropolished samples only. From this we can conclude that the blood wets the surface of our polymer less. This is good because it will cause the platelets, to stick less to the wall of the ‘foreign’ substance and thus be less likely to cause thrombosis.

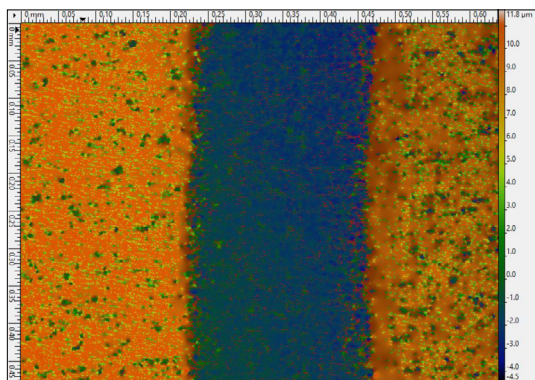


Figure 8. A detail of the “ditch” formed by the needle during tribological examination.

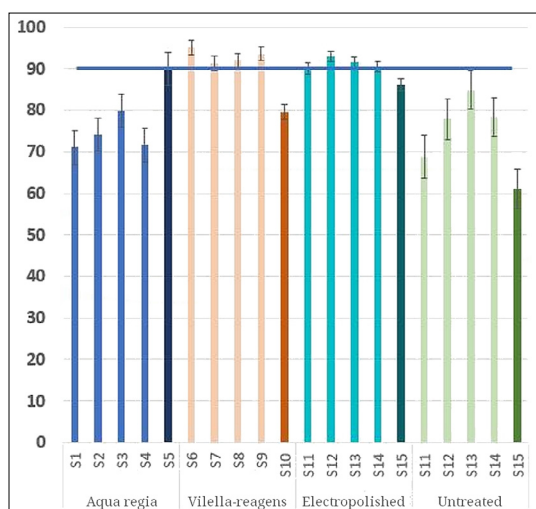


Figure 9. Results of edge angle measurement. Samples S5, S10, S15 and S20 are uncoated reference pieces.

4. Conclusions

Summarizing the above results, we found that etching solutions used in the literature can be used well for material type 316L and create a nice uneven surface, which affects the properties of the coating adhesion. Furthermore, we found that comparing the three different surface treatment methods to the surface of the Vilella-treated inserts, the polymer adhered better than to the electropolished surface formed by the currently most widely used method. In this case, the coefficients of friction were also much lower than in the case of polished samples, so this surface treatment method may be suitable for strengthening the adhesion of the coating. Furthermore, using this surface treatment method based on the edge

angle values, i.e. the wettability conditions, we can say that we have obtained a better result than in the conventional electropolishing method.

Nevertheless, the surface treatment and coating processes have been automated and a suitable surface treatment method has been found for better coating adhesion on 316L inserts, however, further research and studies on stents are also needed.

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