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Research article

A comparison of endothelial cell growth on commercial coronary stents with and without laser surface texturing

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ABSTRACT

Complete endothelialisation of coronary stents is an important determinant of future thrombotic complications following coronary stenting. Stent surface texture is an important factor that influences endothelial cell growth. With the emergence of second and third generation coronary stents, is limited comparative data describing endothelial cell growth in contemporary stent platforms, and limited data available on approaches used to rapidly modify the surfaces of commercial coronary stents to improve endothelialisation. In this study we have determined the *in vitro* proliferation of the primary human coronary artery endothelial cells on the commonly used 4 types of commercial coronary stents and found that the inner surface of BioMatrix drug-eluting stents (DES), after eliminating of the polymer and drug coating, had significantly higher endothelial cell proliferation compared to that of other bare metal stents (BMS): Multi-Link8, Integrity and Omega. The surfaces of the 3 types of BMS which are smooth, displayed similar endothelial cell proliferation, suggesting the importance of surface features in manipulating endothelial cell growth. Laser surface texturing was used to create micro/nano patterns on the stents. The laser treatment has significantly increased endothelial proliferation on the inner surfaces of all 4 types of stents, and Multi-Link8 stents displayed the highest (>100%) improvement. The laser textured BioMatrix stents had the highest absolute number of endothelial cells growth. Our results provided useful information in the endothelialisation potential for the commonly used commercial coronary stents and suggested a potential future application of laser surface bioengineering to coronary stents for better biocompatibility of the device.

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1. Introduction

Coronary heart disease (CHD) is the major course of death worldwide. In combination with balloon angioplasty, the implantation of coronary stents has been a highly effective treatment for CHD, which helps opening up the narrowed coronary arteries and re-establish blood flow to supply the heart. However, despite efforts, complications, e.g., restenosis and thrombosis, are still happening to a considerable population of patients, some of which are life threatening [1,2]. Therefore, there is an increasing demand for improving the biocompatibility of coronary stents.

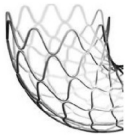
Coronary stents are made of a range of materials, mainly metal alloys such as stainless steel and more recently, titanium, cobalt-chromium and platinum as well as biodegradable polymers [3]. Drug-eluting stents (DES) have significantly reduced in-stent restenosis, by inhibiting abnormal vascular smooth muscle cell (VSMC) proliferation though polymer-aided coating of anti-proliferative or immuno-suppressive drugs on the surface of stents [4–6]. More recently, biodegradable polymer coatings were employed in order to eliminate the inflammatory reactions to the polymers after drug release and potentially reduce the risk of stent thrombosis [7]. These approaches have significantly improved the biocompatibility of coronary stents. However, in addition to inhibiting VSMC growth, the local release of the potent anti-proliferative drug also significantly inhibits endothelial cell (EC) growth, which increases the risk of stent thrombosis [2,8–12].

ECs are the major cell type that is responsible for the maintenance of the homeostasis of vasculature. Through generating nitric oxide, ECs inhibit platelet activation, prevent thrombus formation and keep the underlying VSMC quiescent from proliferation [13–16]. Poor endothelialisation of coronary stents is associated with stent thrombosis and restenosis [11,17,18], and importantly, the endothelial coverage of stent is largely dictated by surface properties of the device [19–25]. This applies to both bare-metal stents (BMS) and DES since the DES would essentially become a BMS after the acute phase drug release, particularly for the biodegradable polymer-coated DES. Therefore, engineering coronary stents with suitable surface properties has been a major focus of innovation.

We previously reported using laser surface engineering approach to change surface texture of the first-generation stent material, stainless steel 316L, as well as a first generation of BMS made up of 361L and found multi-fold increases in the adhesion and proliferation of human coronary artery ECs (hCAECs) on the laser-treated surfaces [26]. For the new generation stent that are made of cobalt chromium and platinum, whilst they have significant advantages including better radial strength and flexibility, complications of in-stent restenosis and incomplete endothelialisation still remain to be significant problems regardless the type of material used for the stent struts [27–29]. Different type of stents has different surface properties which could differentially manipulate EC growth behaviours; however, there has not been previous data comparing their ability in accommodating EC growth across different types of commercial stents. In addition, it is not known if laser surface engineering could also improve endothelialisation of coronary stents that are made up of the new generation materials.

Laser surface engineering of stents is still a novel field. The latest laser technology advances which can provide a larger range for texture design (from nano to micron scale) combined with surface chemistry alteration, demonstrate some distinctive advantages for stent surface modifications [30]. Surface topography affects many surface properties including wettability, protein adsorption, cell orientation, attachment, and proliferation. Surface chemistry is also important in surface engineering of stents as different cells have preferential surface chemistries for attachments. The complex and combined effects of these properties significantly affect the biological response to textured surfaces.

Table 1
Coronary stents used in the study.

Stents	Company	Image *(the design)*	Stent type	Strut material	Strut thickness *(Measured on SEM images)*
Omega	Boston Scientific		BMS	Platinum chromium	77 μm
Integrity	Medtronic		BMS	Cobalt chromium	97 μm
Multi-Link8	Abbott Vascular		BMS	Cobalt chromium	73 μm
BioMatrix	Biosensors		DES (PLA biodegradable, Biolimus A9)	316L Stainless steel	114 μm

One of the future directions for laser surface engineering to improve the surface biocompatibility is changing the polarisation direction to create complex LIPSS (Laser-induced periodic surface structures) textures. Furthermore, using different rastering strategies at various pulse durations and performing laser texturing in different gases could also be employed to generate novel biomimetic textures [30].

In the current study, we have conducted an investigation into comparing the most commonly used commercial coronary stents including Multi-Link8, Integrity, Omega and BioMatrix for their abilities in manipulating hCAEC proliferation and determined the effect of laser surface engineering of stent on the growth of endothelial cells.

2. Materials and methods

2.1. Coronary stents and stent materials

Coronary stents used in the study were Multi-Link8, Integrity, Omega and BioMatrix which were purchased from Abbott Vascular, Medtronic Ltd, Boston Scientific and Biosensors International Group Ltd, respectively (Table 1). Stainless steel 316L was used as the stent material for initial optimising conditions of laser surface engineering.

2.2. Cell culture and cell proliferation

Human primary coronary artery endothelial cells (hCAECs, PromoCell, Germany) were cultured in Endothelial cell growth medium MV (PromoCell, Germany) at 37 °C in a 5% CO2 incubator. 0.5-1 $\times$ 10^5 hCAECs cells per well were seeded in 24-well plates (Corning, Fisher Scientific) containing cut-opened and flattened coronary stents or coupons of stent materials and cultured for 3 days. The stents or coupon materials were then transferred to a clean well. The viable cells growing on the surface of stents or materials were quantified either by DAPI staining followed by cell counting using the ImageJ software on images taken from the fluorescent microscope, or by CellTiter-Glo Luminescent Assay Kit (Promega) according to manufacturer's instructions and read with an illuminometer (Promega).

To investigate the cell behaviour under flow condition, viable cells growing on stent surfaces were determined by using an ibidi flow system (ibidi GmbH, Germany). This system generates shear stress by producing defined flow over the cells, which mimics the condition of blood flow of arteries in vivo. The system includes two pumps that work simultaneously to allow comparing of cell growth on a blank and a textured surface. The pumps are controlled by software that regulates the outgoing pressure applied to the fluid and therefore determines the shear value or flow rate. 1 $\times$ 10^5 hCAECs per well were seeded in 24-well plates containing cut-opened and flattened coronary stents and grew for 24 h. Stents were then transferred into ibidi flow chambers (μ-slide I Luer, ibidi) that were connected to ibidi pumps for supplying Endothelial cell growth medium MV. The system was running for 2 days under a flow rate that produced 15 dyn/cm^2 shear stress. The viable cells growing on the surface of stents were quantified either by counting cell numbers after DAPI staining or by a luminescent assay using CellTiter-Glo Luminescent Assay Kit (Promega).

2.3. Laser surface processing

The 4 types of commercial coronary stents mentioned above were cut open and flattened, and the stent material stainless steel 316L were cut into 1 $\times$ 1 cm ready for laser surface-texturing. A diode pumped Nd:YVO_4 laser with a 532 nm wavelength, pulse duration of 8 ns was used in the study. The laser processing conditions are: 30 kHz frequency, scanning speed 100–300 mm/s and hatch distance 50 μm in nitrogen gas. The laser beam with a near Gaussian intensity distribution (beam quality factor M2 $\sim$ 1.5) was focused onto the surface of the samples with a spot diameter of 55 μm. The laser fluence (energy density) was varied between 1.5 J/cm^2 and 4 J/cm^2. The laser-textured samples were sterilized in 100% ethanol and then washed with Phosphate buffered saline (PBS) solution prior to being placed in a 24-well plate with luminal surface facing up for seeding hCAECs for proliferation experiments as mentioned above.

2.4. Characterization of surface topography and roughness

Surface topography of stent material and stent was analysed using scanning electron microscopy (SEM). Surface roughness measurements were obtained by using MeX software. The software uses SEM images and automatically retrieves 3D information which can be used to obtain metrology measurements.

2.5. Statistics

At least 3 repeated tests were carried out for each experimental condition. Data were normalised to the stent surface areas for cell growth and expressed as mean $\pm$ SE. Differences between groups were analysed using a Student's *t*-test. $P < 0.05$ was considered statistically significant.

3. Results

3.1. Comparison of EC proliferation between different types of coronary stents

Details of the coronary stents used in the study are listed in Table 1. The Multi-Link8, Integrity and Omega stents are 3 types of

commonly used BMS; and BioMatrix is a DES.

The Biolimus A9 coating on BioMatrix DES was first stripped by ethanol treatment in order to eliminate the cytotoxicity effect of Biolimus A9 to EC (Fig. 1A and B(a,b)). This represents the late stage of post stent implantation when the stent strut is exposed after drug release. This is a key stage for the development of late-stent thrombosis. The drug-eluted BioMatrix stents, named BioMatrix stent strut, were used for the comparison of EC growth between DES and BMS for all the following experiments.

The proliferation of hCAECs on the luminal surfaces of the 4 types of coronary stents were then determined. Fig. 1C summarised data from 4 independent experiments which was presented as the percentage of proliferation to that of the BioMatrix. Results showed that there were no significant differences of cell proliferation observed among the 3 types of BMS (Omega, Integrity and Muli-Link8), however, the proliferation of hCAECs on Biomatrix stent struts was significantly higher than that of the 3 other stent types (Fig. 1C).

3.2. *Effect of laser texturing of coronary stents on endothelial cell proliferation*

To create an optimal surface texture of coronary stents for better endothelial cell growth with minimum laser processing time, we tested a range of laser scanning speeds (100–300 mm/s) on the surface of stent material, stainless steel 316L. The laser beam was raster scanned over the surface of the targets using a computer controlled galvo-scanning system equipped with a flat field lens. The full experimental set-up can be found in our previous publication [26]. The speed was increased until a similar texture to our previous study was observed. Results showed that the laser treatment generated micro ripples with nano features (Fig. 2A, B, 2E). A trend of

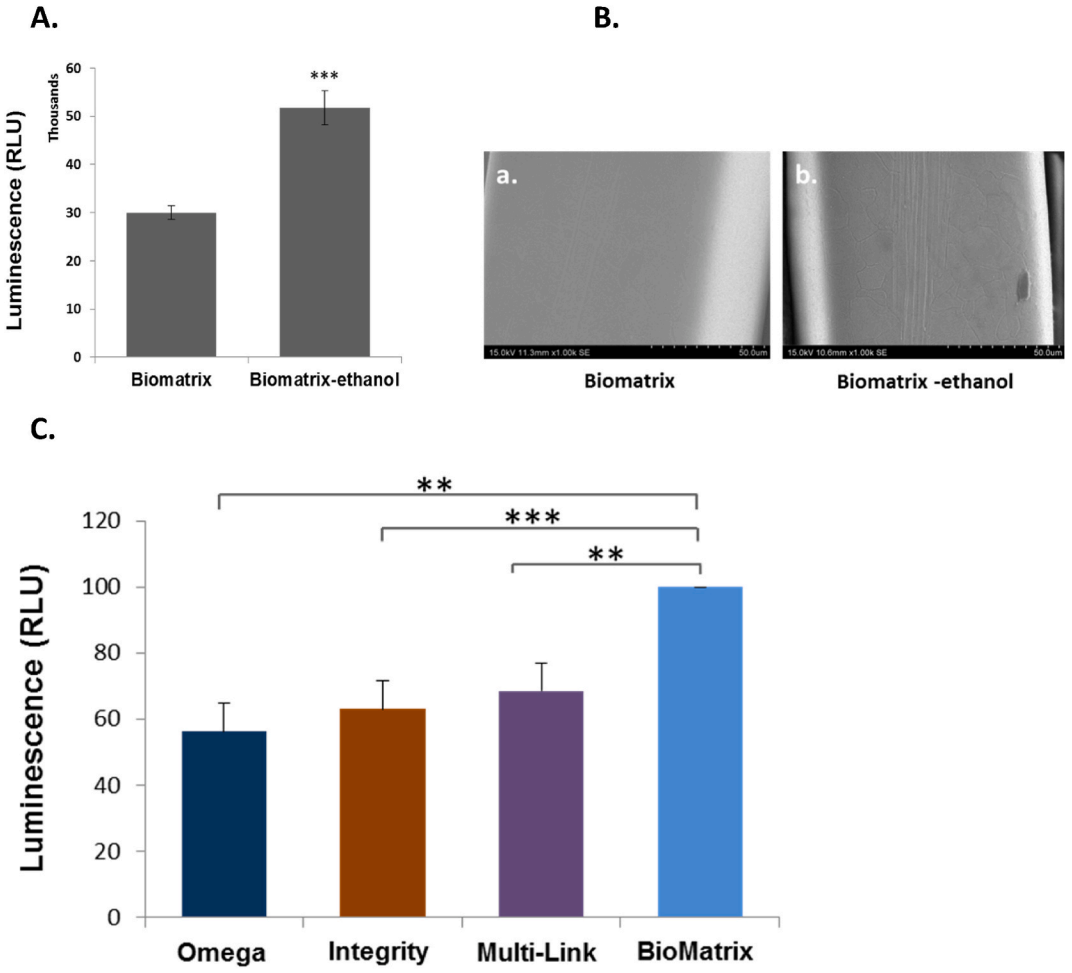


Fig. 1. The proliferation of hCAECs on different commercial coronary stents. A. The drug coating on the BioMatrix coronary stents were first eliminated by ethanol treatment, which enhanced the proliferation of hCAECs on the ethanol-treated BioMatrix stents. n = 3. **B.** SEM images of BioMatrix stents before and after ethanol treatment. **C.** Comparison of the proliferation of hCAECs on the surface of the 4 types of stents. The 4 types of stents were cut open and mounted in a 24-well plate with inner surface facing up. 1×10^5 hCAECs were seeded on to the stent and incubated in a CO_2 incubator for 3 days. The viable cells growing on the stents were measured by CellTiter-Glo Luminescent assay. Results are the summary of 4 independent experiments with 4 repeats each. Data are normalised as the percentage to BioMatrix and presented as mean ± SE. **p < 0.01, ***p < 0.001.

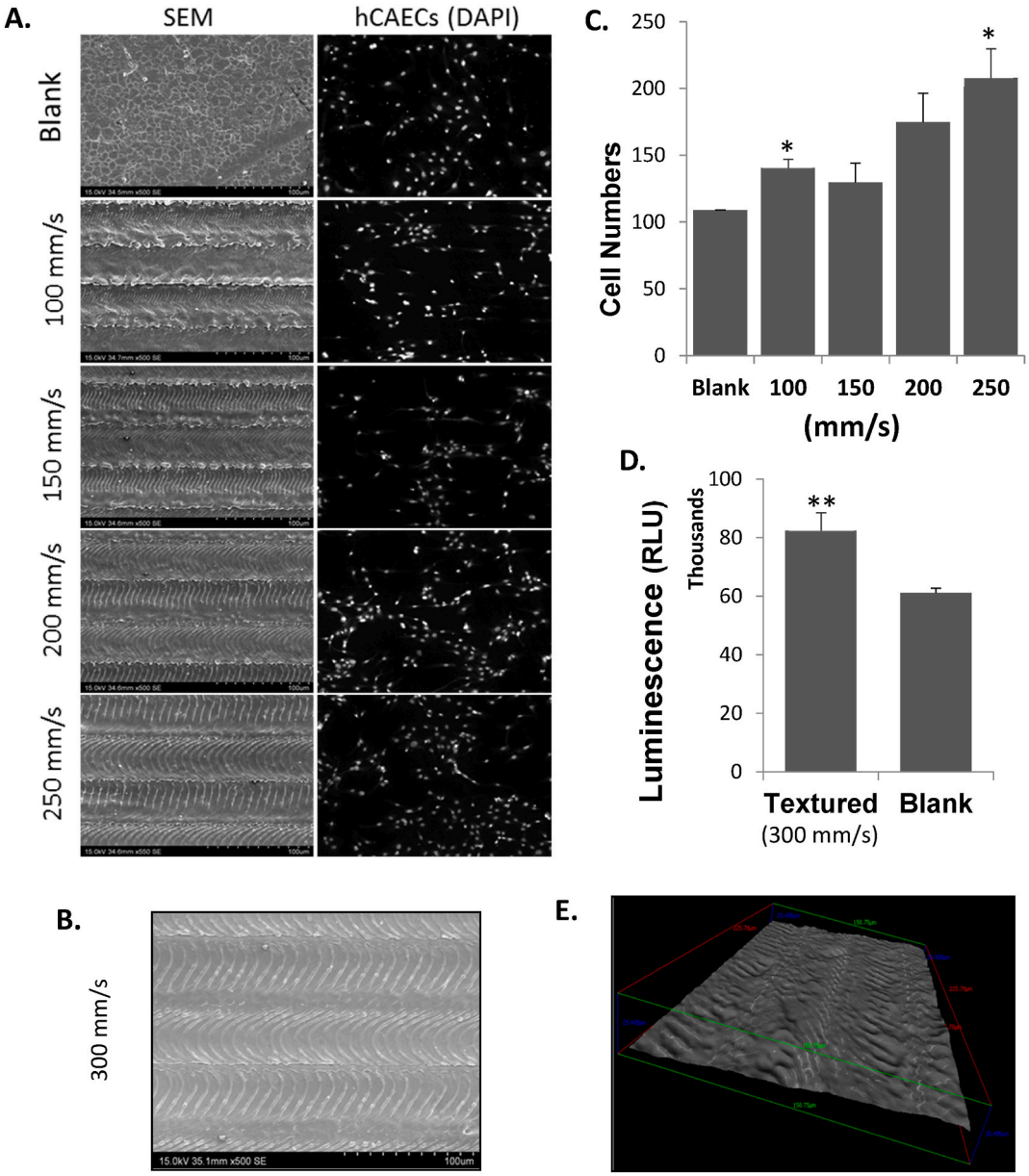


Fig. 2. Proliferation hCAECs on laser-textured stainless steel 316L surfaces. Coupons of stainless steel 316L were textured using a range of different scanning speed at a constant power of 100%, frequency 30 kHz and hatch distance 50 μm in nitrogen gas and imaged by SEM (**A & B**). hCAECs were cultured on the laser-textured surfaces for 72 h and visualised by DAPI staining of the nuclei (**A**) or analysed for proliferation by either cell counting (**C**) or chemiluminescence assay using CellTiter-Glo kit (Promega) (**D**). Mex 3D image of laser-textured stainless steel 316L surface at a scanning speed 300 mm/s (**E**). Data are presented as mean ± SE. Compared to the Blank, *$p < 0.05$, **$p < 0.01$. $n = 3$.

Table 2
Surface characters of laser-textured stainless steel surface using laser processing conditions of 100% power, 30 kHz frequency and scanning speed 300 mm/s. Data are presented as mean ± SE, n = 4.

Stents	Ra (nm)	Rz (μm)	Rsk	Rku
Horizontal (against ripples)	403 ± 88	2.08 ± 0.43	−0.18 ± 0.34	4.73 ± 1.02
Vertical (along ripples)	314 ± 61	1.48 ± 0.25	0.20 ± 0.08	3.14 ± 0.03

increased cell proliferation with increased laser scanning speed was observed (Fig. 2C). Compared to the non-textured surface, the micro- and nano-combined structure generated at a laser scanning speed of 300 mm/s had significantly less surface roughness (Rz 1–2 μm, Fig. 2A and B, Table 2) while still be able to promote EC proliferation (Fig. 2D). At 300 mm/s textures were highly repeatable, clean and precise with a lower roughness compared to our previous study which creates less damage to stent material properties and therefore this speed was chosen in this study.

Based on above finding, the inner surfaces of the 4 types of coronary stents were subject to laser surface-texture at a laser scanning speed 300 mm/s. The micro/nano mixed structure was successfully recreated on the inner surfaces of coronary stents as shown by the SEM images in Fig. 3A. The proliferation of hCAECs on the laser-textured stent surfaces was then visualised under fluorescent microscope (Fig. 3A) and quantified by luminescent assays (Fig. 3B). Results showed that when comparisons were made across different

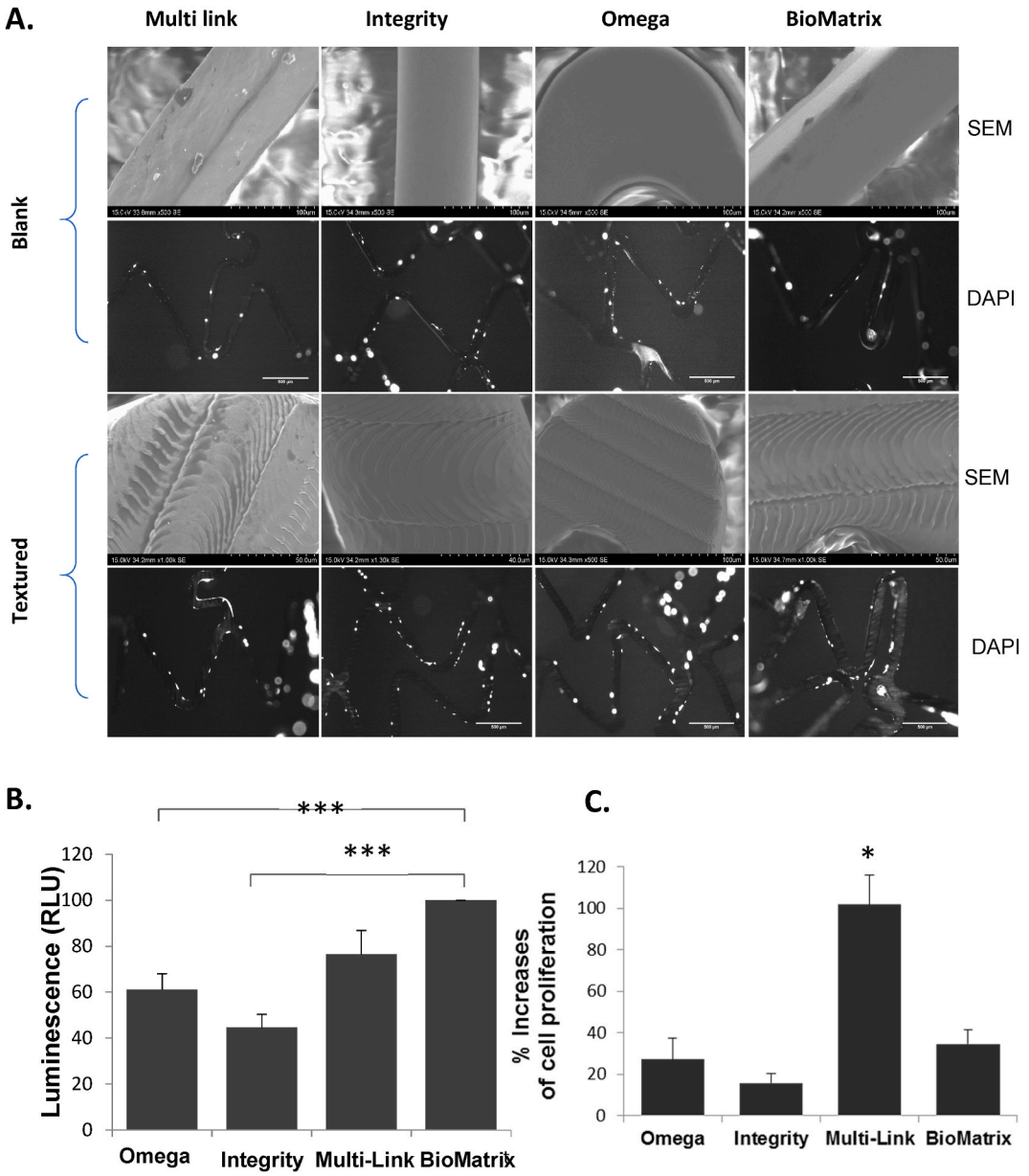


Fig. 3. Proliferation of hCAECs on laser-textured coronary stents. Stents were cut open and the inner surfaces were subject to laser-texturing under a processing condition of 100% power, 30 kHz frequency, scanning speed 300 mm/s and hatch distance 50 μm in nitrogen gas. SEM images are shown in **A.** 1×10^5 hCAECs were grown on the stent surfaces in a 24-well plate for 3 days. The cells were then stained by DAPI and visualised under fluorescent microscope (**A**). Viable cells growing on the stents were measured by CellTiter-Glo Luminescent assay (**B**). The increments of hCAEC proliferation after laser surface-texturing of each coronary stent type compared to its blank are shown in **C**. Results are the summary of 4 independent experiments with 4 repeats each. Data are normalised as the percentage to BioMatrix and presented as mean ± SE. ***$p < 0.01$.

types of stents by measuring viable cells 72 h after growing on the laser-textured stent surfaces using normalised data from 4 independent experiments, the highest hCAEC proliferation was again observed on the laser-textured BioMatrix stent strut (Fig. 3B), suggesting a relatively superior ability of the BioMatrix strut surface in supporting EC growth.

The impact of laser treatment on the EC proliferation for each individual type of stents was then analysed. Data were summarised from 4 independent experiments and presented as percentage changes of cell proliferation to their non-textured counterparts. Results showed that laser surface-texturing has significantly increased the proliferation of hCAECs on the surfaces of all 4 types of coronary stents. The Multi-Link stents showed the highest increment (102%) as compared to its non-laser-treated control (Fig. 3C).

3.3. Impact of flow on endothelial cell proliferation on laser texturing of coronary stents

The flow test results showed that when comparing blank and textured stents under flow by measuring viable EC cells after 48 h, the highest hCAEC proliferation was again observed on the BioMatrix stent (Fig. 4) with a considerable difference to other stent types, suggesting that BioMatrix performed even better in supporting EC growth when under flow. The results were obtained from the normalised data from 3 different experiments.

4. Discussion

Early endothelial coverage is important to reduce the risk of both early- and late-stent thrombosis and allows shorter dual antiplatelet regimes particularly in those patients at risk of bleeding complications [14,15]. Multiple factors including drug coating, surface texture, stent material, surface chemistry and even stent design could potentially influence EC growth behaviour on the stent surface [19,31]. In the current study we have made direct comparison of the *in vitro* proliferation of hCAECs between the surfaces of 4 types of commonly used coronary stents and applied laser surface engineering technology to improve endothelial growth. To our knowledge this is the first comparative study using a range of actual commercial coronary stents.

Cell response to the surface is dominated by surface topography and chemistry. In this study, a surface with both suitable texturing and favourable chemistry to EC cells is produced by using laser surface machining. The novelty of the method used, and the resultant surface has been discussed elsewhere [26].

We previously reported using biomimicry to modify the surface of stainless steel 316L stent material which resulted in a threefold increase in the adhesion and eightfold increase in the proliferation of endothelial cells [26]. This surface resembled the texture of inner rat aorta (observed under environmental SEM) and was produced in nitrogen atmosphere. The study compared same morphology created in argon, nitrogen and air atmospheres to investigate the effect of topography and chemistry on endothelial cells and suggested the importance of the surface chemistry, including the dramatic increase of chromium nitride, for the interaction of endothelial cells with the material surface. Therefore, in the present study, laser machining is performed in nitrogen.

The improved performance of this unique surface texture and chemistry in terms of coronary stents were discussed in our previous paper [26]. The novel super-hydrophilic surface that was created represented a specific combination of micro/nano structures and a particular surface chemistry, which was ideal for increased EC adhesion and proliferation. Surface hydrophilicity increased protein absorption, making it highly suitable for cell attachment. Furthermore, the increased surface area, enabled accommodating more cells in a given-sized surface.

In the present study, we have used our previous method with an improved processing speed, to compare 4 types of commonly used commercial coronary stents for their ability in accommodating hCAECs proliferation *in vitro*. Briefly, we have used a diode pumped Nd: YVO_4 nanosecond laser with a 532 nm wavelength, pulse duration of 8 ns, 30 kHz frequency, scanning speed of 300 mm/s and hatch

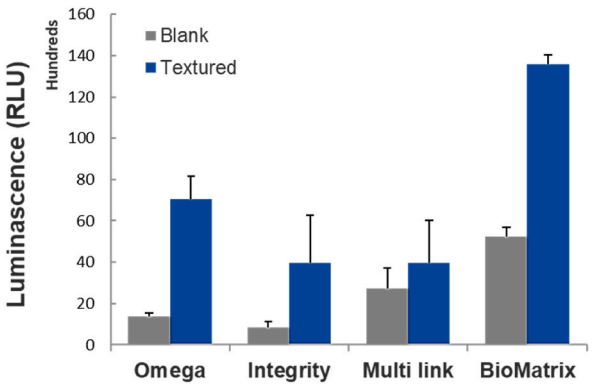


Fig. 4. Proliferation of hCAECs on laser-textured coronary stents under flow. Stents were cut open and the inner surfaces were subject to laser-texturing under a processing condition of 100% power, 30 kHz frequency, scanning speed 300 mm/s and hatch distance 50 μm in nitrogen gas. 1×10^5 hCAECs were grown on the stent surfaces in a 24-well plate for 24 h following by growing in ibidi chambers under flow for 48 h. Pump shear stress was 15 dyn/cm^2. Viable cells growing on the stents were measured by CellTiter-Glo Luminescent assay. Results are the summary of 3 independent experiments with 3 repeats each. Data are presented as mean ± SE. *$p < 0.01$. A. B.

distance of 50 μm, in nitrogen gas.

It is also important to note that although femtosecond and picosecond lasers are advantageous in terms of higher accuracy and shorter processing time, nanosecond lasers continue to dominate the significant laser ablation and processing business due to the massive cost difference between the systems. This indicates another advantage of the method used in the present study.

Flat sheets of stainless steel 316L stent material was used for initial investigations and to observe the effect of changing laser parameters on surface structures (Table 2). Due to the value of stents used in this study, the number of stents provided was limited. Considerable number of additional stents are needed to perform surface characterisation (topography and chemistry) with statistical analysis, before and after laser machining which could be a beneficial addition to continue this study.

Surface topography affects many surface properties including the wettability, which could significantly influence the biological response to the surface. Higher wettability provides a larger attachment area between liquid environment and the surface, resulting in increased interaction between biomedical instruments and the biological environment. Additionally, different surface chemistries can alter wettability. Contact angle is an indication of the surface wettability. Contact angle measurement method and data were provided in our previous study [26]. All surfaces created in the present study were superhydrophilic with a contact angle near zero indicating a highly wettable surface.

We have observed differences in endothelialisation between contemporary stent platforms, and found that BioMatrix stent struts, after stripping-off the polymer and drug coating, displayed a highest hCAEC proliferation compared to that of the other 3 stents. We show that despite the differences in materials that are used in the manufacture of coronary stents, i.e., cobalt chromium for Multi-Link8 and Integrity stents and platinum chromium for Omega stents, the surfaces of the 3 types of BMS had similar rate of hCAEC proliferation, suggesting that stent materials of BMS are unlikely to play an important role in manipulating EC growth and the smooth surface of BMS may have a universal effect on EC growth behaviour regardless the types of metal alloys used for making the stents.

BioMatrix stent, after stripping off the polymer, showed the highest hCAEC proliferation before and after laser texturing in static (Figs. 1C and 3B respectively) and under flow (Fig. 4). Under SEM, the stripped off surface was observed rougher than other stents (Fig. 5A–D), indicating the important role of surface features in cell proliferation compared to the stent material. This could be the reason for high proliferation before laser machining. In addition, the laser beam may have interacted differently with this surface due to the difference in roughness, which could explain the high proliferation of laser textured BioMatrix. This notion supports our previous study and other publications that surface texture is important for ECs growth on the stent surfaces [19–24]. Detailed investigations of topographies and chemistries created by laser machining on 4 stent types could expand this further in future.

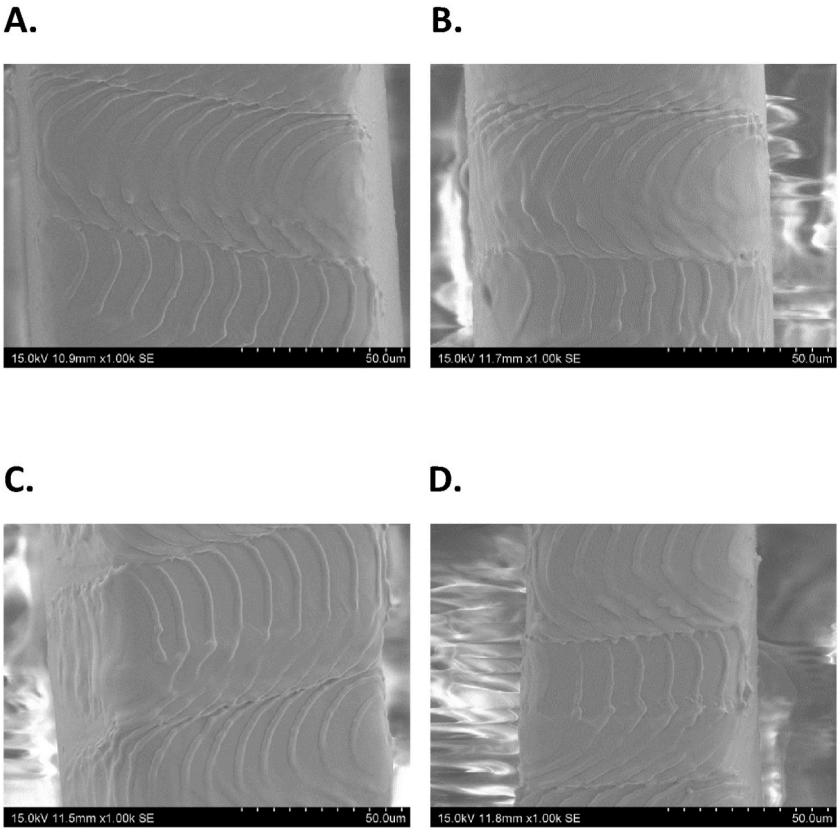


Fig. 5. SEM image after laser texturing. (**A**) BioMatrix after 5 min soaking in ethanol, (**B**) Integrity, (**C**) Multi link-8, (**D**) Omega. Size of scale bar is 50 μm on each image.

An important issue for DES is the likelihood of developing late-stent thrombosis due to incomplete EC coverage caused by the non-selective cytotoxic drug coating [11]. Drug therapy post PCI is a delicate balance between preventing such stent thrombosis events, particularly in the phase when the stent struts have not become adequately endothelialised, and minimising the duration of dual anti-platelet therapy, particularly in those patients at risk of bleeding complications. The polymer- and drug-eliminated BioMatrix stent surface used in our EC proliferation experiment represents such a stage, and the results provide an indication of promising endothelialisation of BioMatrix stents compared to the BMS tested. One of the reasons behind could be that the surface of the BioMatrix stent strut was made relatively rougher than the BMS for effective polymer and drug coating, although this prediction is to be approved by additional surface characterisation.

To produce an optimal surface texture for better EC growth, advanced laser surface-texturing was conducted on the 4 types of coronary stents. Results showed that comparing before and after laser treatment, EC proliferation had increased for all the stents tested with the increment of Multi-Link8 stents to be the highest (102%) and the Integrity the least (Fig. 2). Whilst both Integrity and Multi-Link8 stents are made of cobalt chromium, there were significant differences in the endothelialisation of these 2 stents suggesting that additional factors to stent material play a role in endothelialiaisation. It is notable that the Integrity BMS is made of a single, continuous strand of wire formed into a sinusoid (Table 1), a considerably different design from that of the Multi-Link8 stents. The round wire strut of Integrity stents has a relatively smaller flat inner surface for laser-texturing (Fig. 3A). It is likely that the laser beam was not in-focus on the acentric area of the Integrity stent surface which had produced slightly different texture, not ideal for EC proliferation (Fig. 5).

If comparing Multi-Link and Omega stents, no significant difference in cell proliferation was observed when tested before and after laser machining (Figs. 3B and 4 respectively), however, the order was reversed. It is important to note that there could have already been intrinsic differences for cells growing on different stents after laser texturing, therefore they may have different responses to shear stress. Nonetheless, the exact molecular mechanisms need to be further investigated in future. We believe the results from the flow experiment is closer to in vivo situation.

The results support a role of laser surface engineering in improving the endothelialisation of coronary stents and have extended our finding to the commonly used coronary stents that made of new generation of stent materials.

Compared to our previous study [26], we have significantly increased the laser scanning speed in the current study. One of the reasons behind was to minimise the entire processing time for translating our research to future cost-effective manufacturing. It is estimated that using the 300 mm/s scanning speed to texture a stent of 18 mm length and 3.5 mm diameter, 21.6 s would be sufficient for each stent. Additionally, higher laser scanning speed would produce less rough surface as evidenced by the fact that Rz value of the textured surfaces has been brought down from the previous 1–20 μm to about 1–2 μm when the scanning speed increased from 60 mm/s to 300 mm/s. The smaller surface texture would give less damage to the material properties of the stents, which helps retains their radial strength.

When using the BioMatrix DES for comparison of the EC growth with other BMS, the PLA supported-Biolimus A9 coating was first stripped by ethanol treatment. The reason for this was that following implantation of DES, a time point when the acute phase of drug release and polymer degradation have completed represents a key stage for the development of late-stent thrombosis. Such DES surfaces at this stage should be used for comparison of EC growth between DES and BMS. Result showed that after 5 min incubation in ethanol, the proliferation of hCAECs on the inner surface of the BioMatrix stents was significantly restored (Fig. 1A), suggesting the elimination of the anti-proliferative drug. SEM imaging showed appearance of a rougher stent surface after ethanol treatment which indirectly reflected the elimination of the polymer coatings (Fig. 1B).

5. Conclusion

In summary, we have compared 4 types of commonly used commercial coronary stents for their ability in accommodating hCAECs proliferation *in vitro*, and found that BioMatrix stent strut, after stripping-off the surface coating, displayed significant higher endothelial growth compared to that of the 3 types of BMS (Multi-Link8, Integrity and Omega). The lack of differences for hCAEC proliferation between the 3 types of BMS suggests an important role of surface texture, instead of material, in manipulating EC growth. Laser surface engineering could rapidly produce texture on the surface of coronary stents which has significantly increased hCAEC proliferation to different extend. The over 100% increment of EC proliferation on laser-textured Multi-Link8 stent that is made of cobalt chromium alloy has extended our previous finding to the new generation device, further demonstrating the power of laser technology to be used in future bioengineering of biocompatible coronary stents including the newer generations of polymer stents.

Additional information

No additional information is available for this paper.

Declaration of generative AI and AI assisted technologies in the writing process

During the preparation of this work the author(s) did not use any generative AI or AI assisted technologies.

CRediT authorship contribution statement

Nazanin Mirhosseini: Writing – review & editing, Writing – original draft, Validation, Project administration, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Lin Li:** Writing – review & editing, Validation, Supervision,

Methodology. **Zhu Liu:** Writing – review & editing, Validation, Supervision, Methodology. **Mamas Mamas:** Writing – review & editing, Validation, Methodology. **Douglas Fraser:** Writing – review & editing, Validation, Methodology. **Tao Wang:** Writing – review & editing, Validation, Supervision, Methodology.

Declaration of competing interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests: Nazanin Mirhosseini has patent Methods of manufacturing superhydrophilic implants licensed to The University of Manchester.

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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