Nanoroughness, Surface Chemistry and Drug Delivery Control by Atmospheric Plasma Jet on Implantable Devices

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Alessandro Patelli,[*,†] Federico Mussano,[‡] Paola Brun,[¶] Tullio Genova,[‡,§] Emmanuele Ambrosi,[‖] Niccoló Michieli,[†] Giovanni Mattei,[†] Paolo Scopece,[⊥] and Lorenzo Moroni[#]

†*Dept. Physics and Astronomy, Padova University, via Marzolo 8, 35131 Padova, Italy*

‡*CIR Dental School, Dept. Surgical Sciences, Torino University, 10126 Torino, Italy*

¶*Dept. Molecular Medicine, Unit of Microbiology, Padova University, 35121 Padova, Italy*

§*Dept. Life Sciences and Systems Biology, Torino University, 10124 Torino, Italy*

‖*Dept. Molecular Sciences and Nanosystems, Venezia University, 30172 Venezia, Italy*

⊥*Nadir srl, 30172 Venezia, Italy*

#*MERLN - Institute for Technology-Inspired Regenerative Medicine, Maastricht University, 6229 ER Maastricht, The Netherlands*

E-mail: alessandro.patelli@unipd.it

Phone: +39 049 827 7039

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Abstract

Implantable devices need specific tailored surface morphologies and chemistries to interact with the living systems or to actively induce a biological response also by the release of drugs or proteins. These customised requirements foster technologies that can be implemented in additive manufacturing systems. Here we present a novel approach based on spraying processes that allows to control separately topographic features in the submicron range ($\sim$ 60 nm - 2 μm), ammine or carboxylic chemistry and fluorophore release even on temperature sensitive biodegradable polymers such as polycaprolactone (PCL). We developed a two-steps process with a first deposition of 220 nm silica and poly(lactic-co-glycolide) (PLGA) fluorescent nanoparticles by aerosol followed by the deposition of a fixing layer by atmospheric pressure plasma jet (APPJ). The nanoparticles can be used to create the nano-roughness and to include active molecule release, while the capping layer ensures stability and the chemical functionalities. The process is enabled by a novel APPJ which allows deposition rates of 10 - 20 nm$\cdot$s^{-1} at temperatures lower than 50 °C using argon as process gas. This approach was assessed on titanium alloys for dental implants and on PCL films. The surfaces were characterized by FT-IR, AFM and SEM. Titanium alloys were tested with pre-osteoblasts murine cells line, while PCL film with fibroblasts. Cell behaviour was evaluated by viability and adhesion assays, protein adsorption, cell proliferation, focal adhesion formation and SEM. The release of a fluorophore molecule was assessed in the cell growing media, simulating a drug release. Osteoblast adhesion on the plasma treated materials increased by 20% with respect to commercial titanium alloys implants. Fibroblast adhesion increased by a 100% compared to smooth PCL substrate. The release of the fluorophore by the dissolution of the PLGA nanoparticles was verified and the integrity of the encapsulated drug model confirmed.

1 Introduction

Implants technology, stimulated by tissue engineering advances, has faced in the last years an always increasing complexity. The driving force has been the chase of mimicking bio-surfaces in order to improve scaffolds or prostheses integration in tissues[1]. However, the accurate replication of the biological surfaces is a challenging task, since materials and structures are difficult to emulate and their mechanisms are still not fully understood[2]. The pivotal features of an implant, which have been found allowing to influence cells growth and differentiation on the implants surfaces, are mainly three: the geometrical and mechanical properties, the surface chemistry and the surface topography.

The mechanical properties, mainly the scaffolds stiffness, are bulk characteristics and are known to guide cells differentiation and growth through mechanotransduction[2]. The geometrical shape, as for example pores interconnection, influences also vascularisation and cells motility driving their growth on preferential directions or improving differentiation[3].

For the surface chemistry aspect, usually the surfaces have to maintain a moderate hydrophilicity (contact angle 40-60°)[4]; negative charge on surfaces improves the proliferation while positive charge increases spreading and differentiation[5,6]. The desired surface chemistry is often obtained by wet chemistry processes or by the use of silane agents. However, in the last years the atmospheric plasma technology is no more confined to the mere activation of the surface[7,8], but it is consolidating for the deposition of stable functional coatings[9–11] even on 3D structures[12]. The surface chemical groups influence mainly the protein adsorption[13] by their polarity and by covalent bonding, inducing a selection also on their folding and orientation[14].

The topography effects can be divided into two main ranges: micrometre and nanometre scales. The roughness of the same order of magnitude of cells size influences mainly their adhesion and the presence of a preferential orientation can also drive their motility and growth in the desired directions[15]. Micro-scale roughness is also known to favour osteoblasts growth due to the enhanced precipitation of minerals in the micro-pores[1]. Due to

these evidences, for example, a large number of commercial dental implants presents sand blasted and acid etched (SBAE) surfaces, showing micro-scale porosity. The nano-roughness superposes a further control since cells feel the nano scale features owing to their filopodia[2,16]. Preferentially disordered structures with a critical distance of the features of 60-70 nm allow the binding of integrins to the surfaces and their maturation to focal adhesion point with their clustering in groups at least of 7-8 proteins[17–19]. The roughness in the range of few tens up to few hundreds of nanometres is a key factor in cells interaction with the implants[20,21]. Rounded shapes in this range allow also to increase tension stress in focal adhesion points influencing the mechanotransduction signal chain and are demonstrated to influence cells differentiation and growth[19]. However, it is always difficult to decouple the effect of the surface nano-topography and chemistry since the production processes affect both concurrently[22]; moreover the suitable technologies depends on scaffolds materials and can encounter limitations on centimetre scale and on 3D shapes[23–25].

Furthermore, the next generation of implants requires smarter features such as the possibility to create topographic or chemical gradients to guide cell growth in the scaffold[2,26] or to release ions or active molecules[6]. For example, in bone implants the presence of antibacterial coatings can prevent infections and therefore prosthesis loosening[27], the release of ions like Ca and Sr improves osteoblast growth[28] or the delivery of proteins, such as growth factors, induces cell differentiation[29].

Scaffold production technologies are still under continuous development to satisfy the complexity of desired features for more functional tissue regeneration[30–32]. Despite the different fabrication methods and materials available a localised control over the topographic and chemical functionaliastion aspects is an open challenge. As an example, titanium based prostheses are characterised by acid etching and anodisation processes respectively for micro- and nano- roughness production, but it is still quite difficult to impart a nanostructures gradient with these methods[1,33]. In a similar way, on polymers the nano- and micro- structures induced by solvent and particulate leaching[1,34] or surface solvent etching[25] affect the whole

scaffold. On the other side, the lithographic technologies find some constraints in centimetre scale 3D scaffolds[35]. However, some first attempts are presented in literature for surface chemistry gradients control on polymers using graphed hydrophilic groups[26] or inducing hydrophilic gradient form the surface to the bulk thanks to plasma processes in vacuum[12,36] or in atmosphere[37].

Here, we present a novel approach to tackle surface chemistry and topography based on a two-step spraying technique: at first nano-particles are deposited by aerosol of a dispersion and secondly a fixing layer is deposited by atmospheric pressure plasma jet (APPJ). In order to highlight the potential of this technology, we present its effects on cell growth tested on two different substrates: grade 1 titanium for dental implants and a thermal sensitive polycaprolactone (PCL) 20 μm thick film. For the two substrates we used different cells lines, respectively pre-osteoblasts and fibroblasts, the first for an application oriented testing (bone tissue engineering), the latter for their lower sensitivity to nano-topography[38]. Drugs or proteins release can be achieved by the use of bio-resorbable nano-particles embedding the active molecules. In this way, the smart additive is protected by plasma reactive ions and radicals and it can be incorporated in the coating without being altered[39]. Moreover, the introduced plasma process allows to select protein, drug or mixture, and which size of particles to use independently. This drug release functionality is investigated using fluorescent nano-particles of poly(lactic-co-glycolide) (PLGA). The presented technology offers an easy way to tailor the scaffolds surfaces, allowing to precisely control key properties such as surface chemistry, roughness and drug release, addressing them separately (fig. 1). Due to substrates temperatures lower than 50 °C during the plasma process, the surface morphology of thermal sensitive material is preserved even at the nanoscale. The fixing layer, which is few hundred nanometres thick, appears like a permeable well-adherent snowfall on the nanostructures. As the whole process is based on spraying techniques, it can be implemented on 3D printers allowing the design of gradients.

2 Experimental section

2.1 Sample preparation

Atmospheric pressure plasma jet

The low temperature of plasma polymer layer deposition was achievable thanks to an innovative design of the jet device (Plasma Stylus Noble, Nadir srl) that couples two generators frequencies: 17 kHz and 27 MHz[40]. The jet is based on a double dielectric barrier discharge (DBD) scheme with a double couple of ring electrodes positioned externally to an alumina tube and powered respectively the upstream couple with an high voltage at kHz and the downstream at MHz (fig. 1). The jet uses argon as process gas. A schematic representation of the atmospheric pressure plasma complete deposition system is available in the Supporting Information (fig. S2). This configuration assured deposition rates of 10 - 20 $nm \cdot s^{-1}$ at temperatures lower than 50 °C avoiding the use of helium. For the experiment Ar 5.0 purity was used as process gas (Sapio Ar 5.0 purity, 4.5 slm), nitrogen (Sapio N_2 5.0 purity) was flushed thanks to an outer coaxial duct at 10 slm to confine the process atmosphere and avoid oxygen content in the plasma area. On the other side, the chemical precursors were introduced by an inner alumina capillary just before the RF electrodes (fig. 1).

Substrates preparation

Grade 2 commercially pure titanium samples of cylindrical shape (⌀ 7 mm, height 3 mm) were supplied by Titanmed s.r.l. with different surface finishing: T1 as worked by lathe, i.e. machined, and T2 sand blasted (large grit) – acid etched (SBAE). The samples were cleaned with acetone and rinsed with isopropanol/H_2O solution 70%, then were sterilised in ultrasound bath for 5 min in isopropanol and rinsed in Milli-Q water. The root mean square roughness (R_{rms}) for T1 samples was about 110 nm in a scanning area of 40×40 μm (Supporting Information fig. S1), while T2 samples roughness is in the micrometer range.

PCL films (440744 Aldrich, average M_n 80.000) were produced by hot moulding. The

mirror polished surface of the Si (100) wafers was functionalised by spinning a solution of perfluorodecyltriethoxysilane (FAS-17, Aldrich 658758, purity 97%) 1% in ethanol followed by drying in oven in air at 250 °C. Two silicon wafers was then used as moulds introducing between them 300 mg of pellets at 90°C and applying a pressure of about 70 kPa. The films obtained were about 20 μm thick characterised by a spherulite surface structure[41] with a RMS roughness of about 50 nm in a scanning area of 40×40 μm (Supporting Information fig. S1). The PCL films were then cut in discs (⌀ 1 cm).

Aerosol nanoparticle deposition

Monodispered silica nano-particles were synthesized using Stöber process[42] starting from tetraethyl orthosilicate (131903 Aldrich, purity 98%), ammonium hydroxide solution (338818 Aldrich, 28% NH_3 in H_2O) and ethanol (EtOH, 1.00983 EMD MILLIPORE) in proportion 1:3:42. The solution was left covered under stirring overnight at 30 °C. The resulting particle average size was about 220 nm with a FWHM of 10 nm. SEM images and size distribution are presented in the Supporting Information (fig. S4). The particles were rinsed in deionised water in ultrasound bath 3 times and then re-dispersed at 0.5 w% in EtOH.

In order to test the possibility to release active molecules, red fluorescent nano-particles of poly(lactic-co-glycolic acid) (PLGA) were supplied by Phosphorex (DEGRADEX® LGFR200, excitation 652 nm / emission 668 nm). The particles, of 200 nm of nominal diameter, were dispersed by ultrasonic bath at 0.5 w% in EtOH just prior the aerosol deposition. SEM imaging was used to verify the dispersion particle with the deposition of a solution drop on silicon substrate, the size distribution of the particle appeared broad with a PDI of about 0.4 (Supporting Information fig. S13).

The deposition via aerosol was performed using a plotter and air pressure nebuliser[43]. The plotter was moved at 50 mm·s^{-1} in parallel scanning lines with step of 1 cm. The nozzle distance from the samples was set to 8 cm, the suspensions was dosed by a syringe pump at 2 ml·min^{-1} and sprayed by a 3 bar air flux. The substrates were kept at 40 °C on a hot

plate to increase the EtOH evaporation rate without affecting the substrates.

Plasma polymer deposition

For the fixing layer deposition, the APPJ was mounted on the same plotter of the aerosol nozzle at a distance of 2 mm from the substrates. As in Mussano et al.[10], two different coatings were deposited and characterised by opposite polarities: positive polarity with amine groups using 3-aminopropyltriethoxysilane (3-APTES, Sigma Aldrich, $\geq$ 98%) as precursor and negative polarity with carboxylic/carboxylate groups starting from methylmethacrylate (MMA, Sigma Aldrich, $\geq$ 99%). The precursors was introduced in the plasma, using the inner capillary just prior the RF electrodes, fluxing Ar in a bubbler containing the liquid monomer at room temperature. For all the processes the power supply at about 17 kHz is set to an output of 8-9 kV peak-to-peak at about 8 W, while the RF power is adapted depending on the substrate using also pulsing mode. The process parameters are presented in the Supporting Information (tab. S1). For FT-IR analyses in transmission mode, the coatings were deposited on also silicon (100) substrates. The deposition rates for all the coatings were in the 10 - 20 nm$\cdot$s^{-1} range.

2.2 Surface characterisation

Scanning Electron Microscopy

The scanning electron microscopy (SEM) was used to investigate the morphology of the deposited nanostructures and how they relate to cell growth (FE-SEM ZEISS Sigma VP and HD). The samples were observed at low accelerating voltages ($\leq$ 5 kV) in order to avoid metallization. Samples cross sections on silicon wafers were obtained by crack propagation on the substrate at room temperature. The osteoblasts on titanium substrates were fixed with 1% glutaraldehyde (Sigma G7776) buffered in 0.1 M sodium cacodylate (Agar, UK) (4 °C, 1 h) after 24 hours incubation period to achieve a suitable cell density for SEM observation of individual cell morphology. Then, cells were dehydrated through a series of ethanol/water

solution from 20 to 100% ended with hexamethyldisilazane (Sigma) soaking and air-drying. The fibroblasts cells on PCL films were fixed with 1% glutaraldehyde (Sigma) buffered in 0.1 M sodium cacodylate (Agar, UK) (4 °C, 1 h) after a 4-day of incubation period. The samples, after washing, were soaked in a solution of 1-butyl-3-methylimidazolium tetrafluoroborate 2.0% in milliQ water (Sigma) for 60 s at room temperature; the excess of solution was removed by a pipette followed by drying in nitrogen flux.

Atomic Force Microscopy

Surface topography was characterised by atomic force microscopy (AFM, model NT-MDT Nova Solver-PRO) in tapping mode. The data was analysed by Gwyddion software levelling the data only by flat plan subtraction and a following rows alignment using the median differences tool. The 2D radial power spectral density function ($W_r(k)$) has been used to describe the roughness as a function of spatial frequencies.

FTIR Spectrophotometry

Silicon wafer coated samples were used for transmission FT-IR spectrophotometry characterisation. Measurements were performed by using a Perkin Elmer Spectrum One spectrophotometer by performing 32 scans for each sample.

2.3 Biological characterisation

Protein adsorption

To quantify the amount of protein adsorbed, the titanium disks were incubated in presence of Fetal Bovine Serum (FBS) in Phosphate Buffered Saline (PBS) at 2% concentration, at 37 °C for 30 minutes. Subsequently, the samples were washed twice with PBS and the adsorbed protein was eluted from the disks using Tris Triton buffer (10mM Tris (pH 7.4), 100 mM NaCl, 1mM EDTA, 1mM EGTA, 1% Triton X-100, 10% Glycerol and 0.1% SDS) for 10 minutes. Total protein amount was quantified using SERVA BCA Protein Assay Micro Kit

(SERVA Electrophoresis GmbH, Heidelberg, Germany) Pierce™BCA Protein Assay Kit (Life Technologies, Carlsbad, California, USA) following the manufacturer's instructions.

Cell assays

To study the biological response *in vitro*, pre-osteoblastic murine cells (MC3T3-E1, ECACC, Salisbury, UK) and primary skin fibroblasts were used. MC3T3-E1 cells and fibroblasts were respectively maintained in Alpha MEM and DMEM supplemented with 10% fetal bovine serum (Life Technologies, Milan, Italy), 100 U·ml^{-1} penicillin, 100 μg·ml^{-1} streptomycin. Both cell types were passaged at subconfluency to prevent contact inhibition and were kept under standard culture conditions (humidified atmosphere of 5% CO_2 in air, at 37 °C). Culture and experimental conditions were previously established in our laboratories and adapted to areas of different tested surfaces.

Cell adhesion

Osteoblasts adhesion was evaluated on titanium samples using a 24-well plate (BD, Milan Italy) as support. Cells were detached using trypsin for 3 minutes, carefully counted and seeded at 3×10^3 cells/disk in 100 μl of growth medium on the disks with different roughness. The 24-well plates were kept at 37 °C, 0.5% CO_2 for 10 min. Before and after fixation in 4% Paraformaldehyde in PBS for 15 min at room temperature, cells were washed two times with PBS and then stained with 1μM DAPI (Molecular Probes, Eugene, California, USA) for 15 min at 37 °C to visualize cell nuclei. Images were acquired using a Nikon Eclipse T-E microscope with a 40× objective. The cell nuclei were counted using the "Analyze particles" tool of NIH ImageJ software (ImageJ, U. S. National Institutes of Health, Bethesda, Maryland, USA, http://imagej.nih.gov/ij/).

Fibroblast adhesion was assessed using 3-(4,5-dimethylthiazolyl)2,5-di-phenyltetrazolium bromide (MTT) test. PCL films with or without coating or nanoparticles were incubated with 100.000 fibroblasts/samples in 250 μL culture medium. Samples were incubated at 37

°C for 15 min, rinsed in PBS to remove non-adherent cells and added with MTT for 4 hrs. Reaction was stopped by acidification and the optical density (O.D.) was spectrophotometrically recorded. A standard curve prepared by seeding decimal dilution of cells was used to calculate cell number. Data are reported as ratio over not functionalised substrates.

Cell viability

To assess the viability, MC3T3-E1 cells were plated at a density of 2500 cells/sample in 24-well culture dishes, at 1, 3 and 7 days, using CellTiter-Glo® (Promega, Milan, Italy). The CellTiter-Glo® Luminescent Cell Viability Assay quantifies viable cells owing to the ATP content. The amount of ATP is indeed directly proportional to cell number.

Fibroblasts (100.000/samples) were cultured for 24 hrs at 37 °C, washed and then incubated for 4 hrs with MTT to assess metabolic activity in viable cells[44]. Reaction was stopped and recorded as described above. A standard curve prepared by seeding decimal dilution of cells was used to calculate the number of viable cells.

Cell proliferation

Fibroblasts were labelled with CFSE cell tracker (Carboxyfluorescein succinimidyl ester; ThermoFisher), a green fluorescent dye retained within cells for long periods, and cultured on PCL films with or without coating or nanoparticles at 37 °C for 7 days. Once incorporated within cells the dye is not transferred to adjacent cells, but it halves within daughter cells thus following each cell division up to 7-8 divisions. Percentage of positive cells were quantified by FACS analysis.

In a parallel set of experiments, CFSE-related fluorescence was visualized by confocal microscopy. Briefly, after 7 days in colture cells were fixed in 4% paraformaldehyde for 15 min at room temperature, washed twice in PBS and mounted in a drop of mounting media (Sigma). Samples were analyzed and photographed using a Leica TCS-NT/SP2 confocal microscope

Focal adhesion

Fibroblasts were cultured on PCL films with or without coating or nanoparticles for 24 hrs at 37 °C. For Western blot analysis total proteins were extracted in RIPA buffer (150 mM NaCl, 50 mM Tris-HCl, 0.25% wt/vol sodium deoxycholate, 0.1% Nonidet P-40, 100 µM NaVO4, 1 mM NaF, 1mM phenylmethylsulfonyl fluoride, 10 μg·ml^{-1} aprotinin, 10 μg·ml^{-1} leupeptin). Particulate material was removed by centrifugation and proteins (25 μg/line) were separated with 10% SDS PAGE and then transferred to nitrocellulose membranes. Membranes were probed with anti-FAK phospho Y397 antibody (clone: EP2160Y; Abcam) and then incubated with the proper HRP-conjugated secondary antibodies. Control loading were performed using anti-β-actin antibody (Sigma). Bands were visualized using enhanced chemiluminescence (Millipore). Images were captured using Hyper Film MP (GE Healthcare) and densitometry analysis was performed using NIH ImageJ software.

Statistical analysis

Results are reported as mean±SEM of two independent experiments, each performed in triplicate. Statistical analysis was performed using GraphPad Prism 3.03 (San Diego, California, USA). Statistical significance was calculated using one-way ANOVA test followed by the Newman-Keuls post-hoc test; $p<0.05$ was considered statistically significant.

3 Results and discussion

3.1 Nanostructured coating design

The here proposed technology aims to decouple the roughness and surface chemical functionalisation on any substrate at low temperature. The process involves two steps: particles spraying followed by a APPJ layer deposition to fasten the nanoparticles to the surface (fig. 1). As first step, silica monodispersed particles of 220 nm in diameter were selected as building block to create controlled roughness. The choice of such size is due to the expected size of

the focal adhesion (FA) points of the cells on the surfaces that is of the order of few hundreds of nanometres. For the maturation of a stable FA point, indeed, it is needed at least the clustering of 6-7 integrins, with proteins spacing of some tens of nanometres[17]. Moreover the rounded surface would introduce further tension stress to the connections, facilitating the FA widening[19]. The particles were dispersed in ethanol at 0.5 w% and then sprayed as an aerosol on the surfaces. The use of the alcohol was due to its higher vapour pressure in order to avoid the "coffee rings" effect[45] of the drops evaporating on the surface.

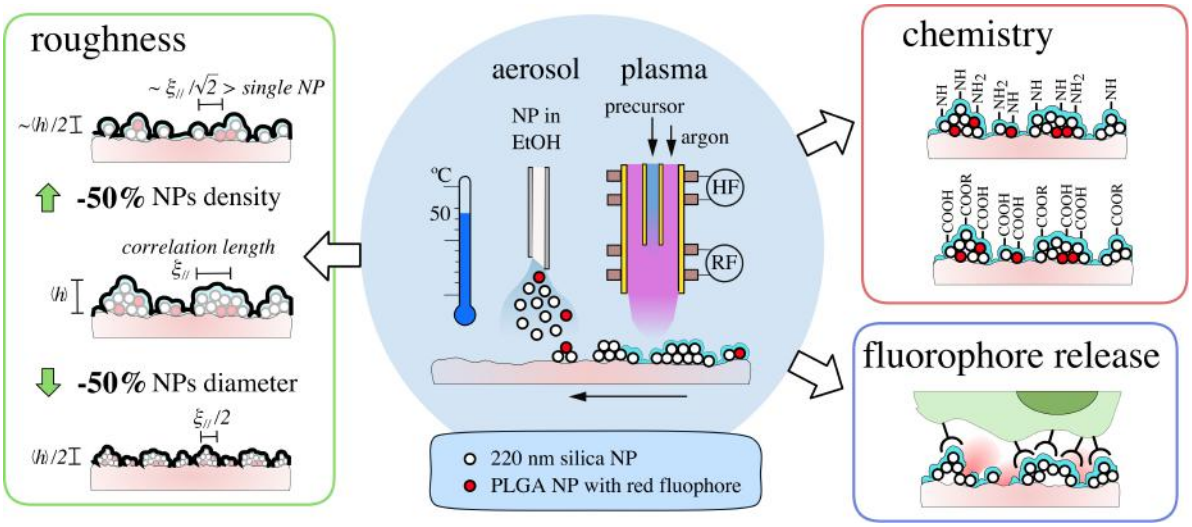

Figure 1: Scheme representing the proposed technology and its control of roughness, chemistry and molecules release. In the centre the deposition process is summarised with the aerosol deposition of the nanoparticles (NPs) and the following deposition of a fastening layer by APPJ, scanning the samples. The film properties are controlled independently: the roughness is controlled laterally and vertically by NPs size and density in the submicron range; the surface chemistry of the deposited APPJ plasma polymer can offer amine or carboxylic functionalities by precursor choice; during cells growth when PLGA NPs with a red fluorophore embedded are used, the fluorophore is released in the culture media.

As a second step a plasma polymer layer was deposited by atmospheric plasma jet on the particles. The role of the coating is two-fold: to fix the particles on the surface, assuring particles adhesion and therefore roughness stability, and to offer the desired chemical functionalisation. In this work, we used similar coatings of the ones presented in Mussano et al.[10] that showed chemical and mechanical stability after ageing for 48 h in water solution and after UV sterilisation. A 30% reduction of the RF average power was carried out to adapt

the process to the substrates (Supporting Information tab. S1). The plasma was driven in pulsed mode at a frequency of 500 Hz and duty cycle of 50% for the titanium substrate to further reduce potential substrate heating due to cylinders corners. However, the average power was kept equal to the one used in continuous mode on PCL and the FT-IR spectra can be nearly overlapped. The thermal load of the plasma process was evaluated by the heating of a copper mass in place of the samples in static conditions[46] and the thermal power resulted of about 185 ± 30 mW (Supporting Information fig. S7), which is in the range of jets used in plasma medicine applications[47]. The deposition rates were about 20 nm·s^{-1} for the MMA and 10 nm·s^{-1} for APTES precursors. The IR spectra in fig. 2a show for the MMA and APTES the presence of the functional carboxyls/esters and amino groups respectively.

The results of this two-steps deposition process can be observed in cross section in fig. 2b. The particles appeared covered by a plasma polymer about 100 nm thick: a thicker layer was observed on the top of the particles while it decreased when getting closer to the substrate, due to a shadowing effect of the particles, themselves. Therefore, the coating thickness was less uniform than in vacuum processes[42]. Nevertheless, the coating was present even at the base of the particles, gluing them to the substrate. Indeed, the atmospheric pressure nature of the plasma deposition assures an isotropic flow of the radicals. Therefore, the thickness is determined mainly by the local acceptance angle: the top of the particle is exposed to more than a semi-plane of incoming radicals while the substrate below the particle sees its semi-plane partially occupied by the particle itself. Moreover, as the coating grows, the acceptance angle for shadowed areas further decreases, since the coating grown on the upper part and the one grown on the substrate get in contact. As can be observed by the SEM cross section (fig. 2b bottom) no voids could be detected at this later growth stage between the coating on the particle and on the substrate. However, a geometrical junction is always visible and suggests anyhow the presence of discontinuities in the coating. Therefore, in order to ensure a good adhesion of the particles on the surfaces, the fixing layer must have the same thickness of the particle radius. At the same time, the presence of these defects around

the particles guarantees coating permeability, which can be tailored just by controlling its thickness.

The particles were randomly positioned leaving part of the substrate area free or arranging up to in stack of no more than 3 levels. The presence of substrate areas free of particles is needed to allow the particles fastening by the coating. However, the fixing layer successfully bonded also the particles to each other, when at least part of each was in view of the plasma (Supporting Information fig. S6). Particles in the lower levels, if fully shadowed by the others, could not be coated resulting in a loss of mechanical stability. Due to that effect if all the surface would be covered by 3 layers no anchoring to the substrate can be achieved. Therefore, the choice of particles size is relevant in order to reach the desired roughness range with a limited number of particles deposited.

On the other side, the layer formation is obtained by a plasma enhanced chemical vapour deposition (PECVD) at atmospheric pressure, therefore the control of the surface functionalities is obtained by precursors and process parameters selection. The condensation of the coating on the surface is obtained via the creation of precursor's fragments along the interaction with plasma reactive species and by their recombination in the gas phase and on the surface[48]. As a matter of fact the resulting coating is characterised by a branched network of heterogeneous molecules. However, chemical radical or ionic chain growth are still present allowing to preserve some functional properties of the conventional polymer. In fig. 2a the FT-IR spectra of the fixing coatings are presented, in order to verify if precursor functional groups were preserved during the deposition. For the coatings obtained with the APTES precursor, nitrogen containing groups are visible as primary amines at 1558 cm^{-1}, at 1660 cm^{-1} as stretching vibration of the oximes and at 3200–3400 cm^{-1} as amides. The presence of the primary amines confirms the retention of some aminopropyl chains of the precursor[10]. The presence of oximes and amides is probably due to precursor oxidation during the deposition process, as supported also by the presence of the hydroxyls groups at 3000–3500 cm^{-1}. The oxygen was dragged in the plasma region by the jet movement on the plotter

even if nitrogen was fluxed in the outer duct of the torch to confine the process atmosphere. Nevertheless the precursor oxidation was partially prevented as shown also by the methyl groups visible at 1465–1375 and 2980–2880 cm^{-1}. Moreover, the silica coating backbone is recognizable in the FT-IR spectra at 1200 and 1000 cm^{-1} that prevents its dissolution in aqueous media[49,50].

The MMA precursor was used to obtain carboxylic acid surface chemical functionalities. In this case the process is adjusted in order to produce carboxyl groups thanks to chemical reactions. During the deposition process the precursor is fragmented leading to the formation of aldehydes and ketones[51]. The oxidation of the aldehydes during the process can lead to the formation of the carboxyl groups[51]. This reaction process is supported by the FT-IR spectra obtained, where a broad absorbance peak centred at 1720 cm^{-1} can be preferentially attributed to carboxyl groups (absorbing in the range 1700—1725 cm^{-1}) than to esters (1730—1735 cm^{-1}) and to the presence of ketones (1705—1720 cm^{-1}) and aldehydes (1730—1740 cm^{-1})[52]. The presence of the carboxyl groups is suggested also by the intense OH vibration band between 3200–3500 cm^{-1}. Part of the methyl groups coming from the precursor was maintained saturated in the coating as can be deduced by the presence of the adsorption signal at 2850–3000 cm^{-1} and its absence in the 800–1000 cm^{-1} range[51]. The C=C signal at 1635 cm^{-1} is frequently observed in MMA plasma deposited coatings and is related to unsaturated structures that may form during the plasma process[53]. However, in order to verify the electrophilic character of the coating the toluidine blue O (TBO) test was performed[54], obtaining a density of negatively charged groups of about $4.8 \pm 0.6 \cdot 10^{14}$ cm^{-2} .

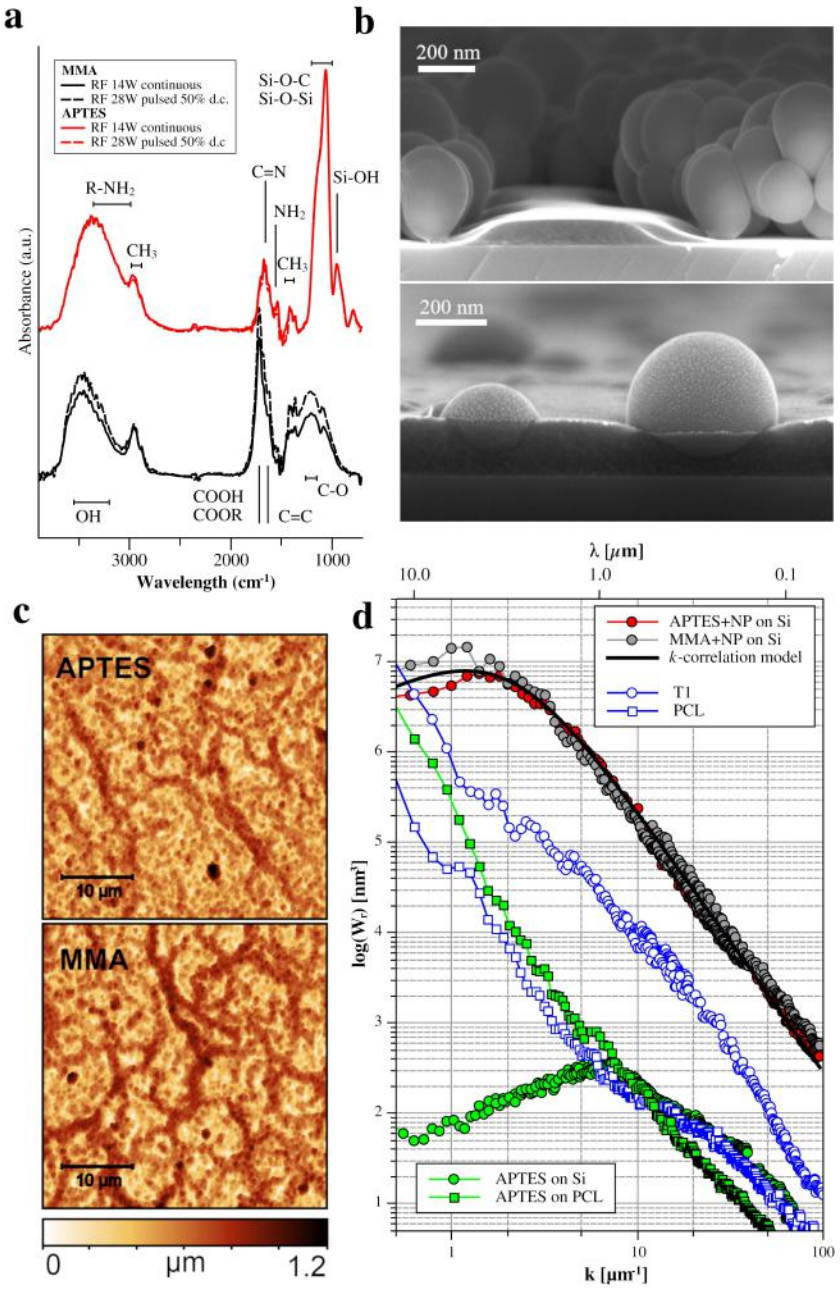

Figure 2: (a) FT-IR spectra of the plasma polymers deposited on silicon starting from MMA and APTES precursors using the parameters for titanium and PCL substrates. (b) SEM cross-section images of the SiO_2 ($\varnothing$ 220 nm)(top) and Degradex PLGA (bottom) nanoparticles deposited on a silicon wafer and coated with the APTES plasma polymer. (c) AFM topography of the 2-steps process on silicon substrate. (d) Radial power spectral density $W_r(k)$ of 2-steps processes on silicon with different chemical functionalities overlapped with the fitted k-correlation model curve. For comparison are added the $W_r(k)$ of the T1 and PCL substrates and of the APTES plasma polymer on Si and PCL substrates.

Hence the two plasma polymers offer two different surface chemistries and are not influ-

enced by substrate material or morphology. In fig. 2c, the topographies obtained for the two fixing layers are shown. Both surfaces showed an average height of about 500 nm respect to the substrate and a root mean square roughness (R_{rms}) of 180 nm, which confirms that the particles are distributed on average in 2 levels. At smaller scale scanning range the single particle as roughness building blocks could be observed (Supporting Information fig. S8). In order to study the roughness as a function of the spatial frequency, the radial power spectral density ($W_r(k)$) of the roughness, which corresponds to the 2D Fourier transform of the auto-correlation function integrated over all the angles, was calculated (fig. 2d). For the analysis AFM measurements at different scanning range from 5×5 up to 80×80 μm were recorded. The $W_r(k)$ of the samples with MMA and APTES coatings on the nano-particles fully overlap and is characterised in a log-log scale by a linear slope at higher spatial frequencies and then saturates forward lower frequencies.

This behaviour is characteristic of a self-affine structure of the surface[55] and is related to the production process of the roughness. The silica nanoparticles are the building block of the growing interface, and added one by one produce a fractal structure that is usually represented by the following scaling form:

$$\sigma^2(L,h) = L^{\alpha} f(h/L^{\alpha/\beta}) \tag{1}$$

where σ is the variance in the surface height, h is the deposit thickness and L is lateral size of the system in simulation models, all parameters in unit of the building block[56]. The lateral size in real surfaces can be associated to the correlation length $\xi_{\parallel}$, which depends at the same time on h, slowly increasing as the thickness increases. The scaling function $f(x)$ has a constant value for $x \gg 1$ and has the form $f(x) \sim x^{\beta}$ for $x \to 0$. Therefore, as the particles deposition process proceeds, the roughness of the surface rises up to a certain saturation value[57]: the higher is β the faster is the increase, on the other side $\xi_{\parallel}$ acts as threshold limit for this scaling expansion. The $\xi_{\parallel}$ is related to the relaxation processes during the growth as the interactions among the spheres[58].

The $W_r(k)$, reflecting the same features of eq. 1, can be used to evaluate the different scaling parameters. By fitting the linear part in the log-log scale with the function $W_r(k) \propto k^{-2(\beta+1)}$, the β exponent is retrieved together with the fractal dimension as $D_f = 3 - \beta$[59]. For all the deposited structures, we obtained $\beta = 0.45 \pm 0.05$ and $D_f = 2.55 \pm 0.05$, which are in agreement with the 3D simulations where diffusion-limited aggregation mechanism and no long-range interactions are considered[60]. These hypotheses well fit with our deposition mechanism where the use of ethanol and of a mild heating avoid surface wetting limiting the possibility of a long-range rearrangement. Therefore, taking into account the self-affine surface behaviour, the topography evolves following the scaling law $h(bx) \sim b^{D_f-2}h(x) \sim \sqrt{b}h(x)$, which means, for example, that a lateral scaling of a factor 4 in building block units leads to a factor 2 in height. At the same time, the roughness is related to mean deposit thickness as $\sigma \sim \langle h \rangle^{\beta} \sim \sqrt{\langle h \rangle}$[56].

The whole $W_r(k)$ with its saturated part can be studied modelling an isotropic autocorrelation function, which allows to introduce a correlation length. The k-correlation model[61] considers for the autocorrelation function a continuous transition model between exponential and gaussian. For the radial power spectral density the corresponding fitting equation is:

$$W_r(k) = \frac{\sigma^2 \xi_{\parallel}(\xi_{\parallel} f)}{(1 + (\xi_{\parallel} f)^2)^{\beta + \frac{3}{2}}} \tag{2}$$

where for $\beta = 0$ the autocorrelation function decreases exponentially, while for $\beta \gg 0$ it get close to a gaussian behaviour. The advantage of this k-correlation model is the possibility to identify $\xi_{\parallel}$.

In fig. 2d the fitting result is presented with a $\xi_{\parallel} \simeq 800nm$, which is close to 3 units of building block and it is expected to slowly increase with layer growth[57]. For lateral frequencies lower than $\xi_{\parallel}^{-1}$ the roughness reached a saturation value and was constant. The $W_r(k)$ functions of the PCL films and the T1 titanium substrate, as well as the positive polarity coating on PCL and on silicon substrates, are presented for comparison. It can be observed that in the submicron lateral range both substrates had roughness more than an

order of magnitude lower and that even the effect of the coatings was irrelevant. Moreover the roughness introduced by the fixing coatings is visible only on the Si substrate that is nearly atomically flat and it showed an overall R_{rms} of about 4 nm. Therefore, the developed process selectively controls the roughness in the submicron range by the choice of the building blocks size, which defines the fractal basic unit. By setting the desired spatial roughness as $\xi_{\parallel}$, the use of building blocks few times smaller than $\xi_{\parallel}$ allowed us to reach efficiently the designed value only with one or few layers of particles deposited. This ensures the possibility to fix the particles on the surface. At the same time, the roughness on the vertical scale propagates as the square root. It has to be considered that the increase of the number of layers increases the $\xi_{\parallel}$ and the overall roughness, but does not influence the roughness height on the lateral scale lower than $\xi_{\parallel}$. Therefore, the choice of the building block size close to the required spatial roughness maximises also the roughness height in the same range.

The nano-structures have been obtained on 2D substrates, but in principle they can be obtained also on 3D geometries as just room temperature spraying techniques at atmospheric pressure are involved. Moreover, as feasible outlook, the proposed process can be implemented directly on a 3D printing bio-polymer platform in order to locally functionalise and control the nanostructures.

3.2 Osteoblasts growth

The biological response *in vitro* on titanium substrates was characterised using the widely diffused pre-osteoblastic murine cell line MC3T3-E1[62]. The substrates considered had two different surface finishing: T1 was after the machining process with a R_{rms} of the order of a hundred of nanometres, while T2 was after a following sand blasting - acid etch process (SBAE) and presented a $R_{rms} > 2$ μm. Therefore, the direct comparison between the results on T1 and T2 allows to highlight the effect of the microscale roughness and its interaction with the deposited nanostructured coating.

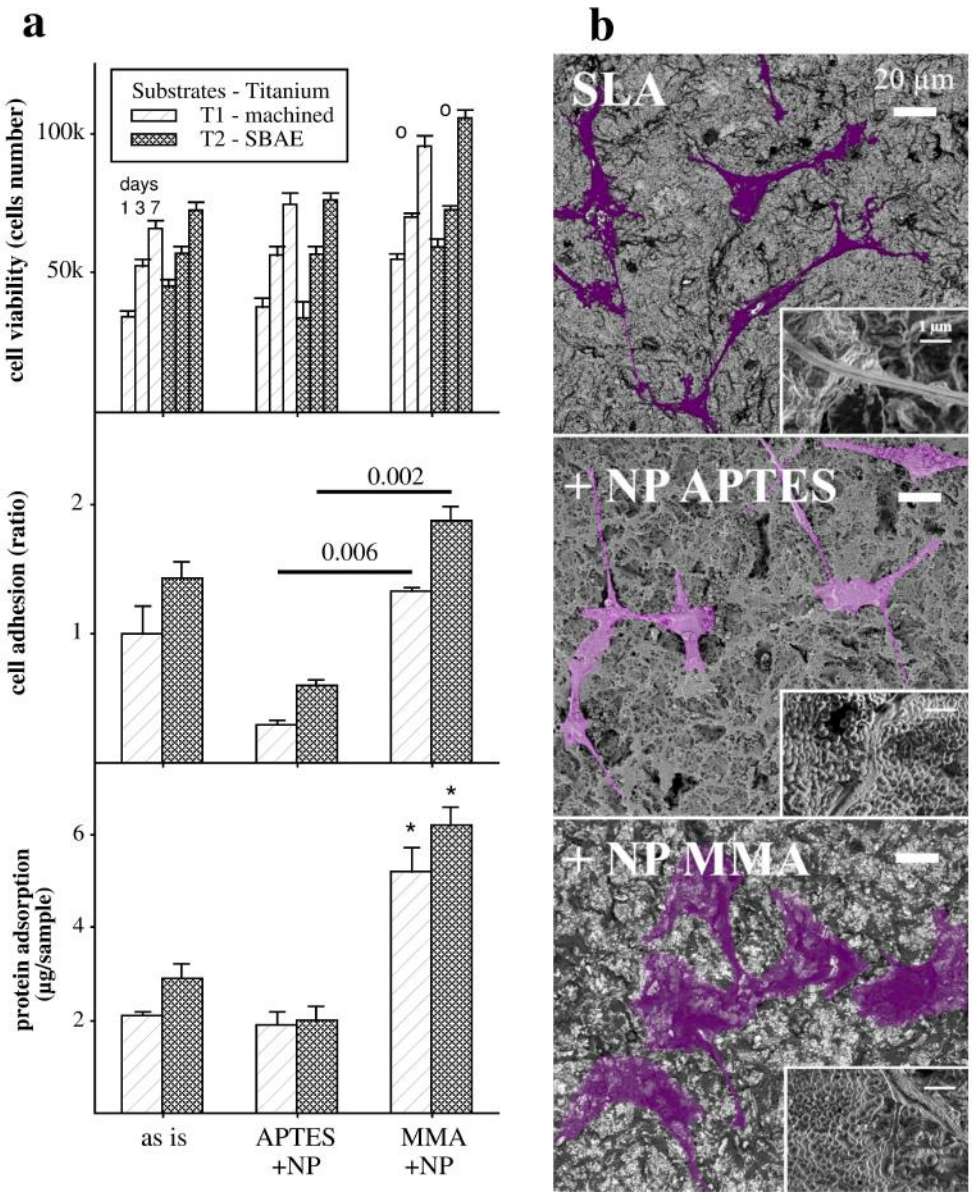

Figure 3: (a) Results on viability (1, 3, 7 days), protein adsorption and cells adhesion for titanium T1 and T2 substrates as is and with the deposited nanostructured layer with the two different chemical functionalisations: amino groups starting from APTES and carboxyl groups using MMA ($*$ 0.001 *vs* as is; $\circ$ 0.01 *vs* as is); (b) SEM images with backscattering detector to highlight cells morphologies on T2 substrates as is and coated (in the insets, higher magnification images obtained with the in-lens detector).

At first the cell viability was evaluated by CellTiter-Glo assay[63]. The T1 and T2 substrates showed, with and without the coatings, similar results (fig. 3a). This comparison highlights that the microroughness in this case seemed not to influence the survival of the

cells in all the assessed time points, which increased along time consistently with a non cytotoxic behaviour of all the considered surfaces[64]. The addition of the nanostructured coatings on both T1 and T2 substrates did not affect the cells viability, which was even enhanced by the deposition of the nucleophilic functionalities.

A similar trend with no differences between T1 ad T2 substrates could be observed for protein adsorption, with and without the coatings. This behaviour was expected, indeed, since the surface chemistry is known to mainly influence the protein adsorption and folding[2]. The main difference is in fact related to the threefold enhancement of the adsorption on surfaces with carboxylic functionality. The same behaviour was found on flat titanium substrates applying the same plasma polymers without nanostructures[10]. In that case, the increase was only of about the 50%, therefore the enhancement of this effect could be a first evidence of a synergistic action of nanoscale roughness and chemistry.

Cells adhesion seems to be more sensitive to the various chemical and morphological effects. In fact, the formation of the focal adhesion points and proteins interaction with the surface is the starting point of a communication chain that guides to cells proliferation and differentiation. As pointed out in Mussano et al.[10] using the same functional coatings and the same cell line, the amino groups on the surface led to a reduction of the proliferation and an improvement in osteogenesis differentiation verified by cell spreading and tapered morphology and by osteocalcin detection. On the other side, the carboxyl groups led to a higher proliferation and to chondrogenic differentiation. These results are widely supported in literature using as well other surface functionalisation techniques[9,13]. Also in fig. 3a we can observe the positive effect on cell adhesion of the nucleophilic surfaces opposed to an evident repression on electrophilic samples. The positive effect of the carboxyl groups is of the same order of magnitude of what obtained on flat substrates, therefore is mainly due to a surface chemistry effect. On the other side, the negative effect of the amino groups was strongly accentuated by the nanoscale roughness. Cells growth is always in competition with the differentiation. Therefore, since the nanoroughess with the selected spatial frequencies is

expected to strongly influence the formation of the focal adhesion points, it is not surprising that we observed less proliferation[21,22].

In cell adhesion, we observed also a significant difference between T1 and T2 substrates in all the conditions considered. This effect can be interpreted as general improvement of the adhesion due to the microscale roughness or maybe to a relaxation of the osteogenesis constrain, since the microroughess can enhance chondrogenesis[34]. Unfortunately, a deeper investigation using confocal reflection microscopy was not possible due to the high retention of the fluorescent probes used for the imaging by the nanostructured coating.

Cell morphology was assessed by SEM (fig. 3b). The surface roughness of the T2 substrates, with the addition of the insulating nanoparticles and plasma polymers hindered a clear view of the cells that were more easily detected on T1 substrates (Supporting Information fig. S10). Cells were spread with a rounded shape in the samples with the carboxylic functionalisation. On the other side, the not treated and the amino functionalised surfaces showed elongated structures with branching filipodia, which usually characterise osteoblasts. The highest degree of branching was observed on T2 substrate with the nanostructured coating and electrophilic functionalisation, thus showing the additional effect of nano- and micro-scale roughness to the chemistry guiding information.

The synergistic effect of chemistry and nanoroughness was even more evident observing the filipodia and the focal adhesion points (Supporting Information fig. S9). The filipodia were able to sense the roughness at the nanoscale[16]. Also in the as-worked sample the focal adhesion point originated from a surface defect of less than 100 nm and the filipodia has similar size. As microroughness was added in T2 samples the filipodia appeared thicker like ribbons about 350 nm wide, which stuck on higher and sharper asperities. As the nanoparticles and a compliant surface chemistry were added the filipodia fully laid on surfaces with the same size of the roughness' domes about 300 nm in diameter and were barely distinguishable from the coating.

However, the possibility to add surface functional groups and nano-roughness indepen-

dently, with respect to standard commercial SBAE titanium substrates, enabled to further guide cell proliferation, leading for example to an improvement of 20% on cells adhesion.

3.3 Fibroblasts growth

In order to show the potentialities of the presented approach to reproduce chemistry and surface roughness on sensitive biopolymers, the same coatings deposited on titanium was deposited on PCL films (m_p 60°C, T_g -60°C). In order to test their biological response, primary fibroblasts was selected as cells similar to osteoblasts but expected to be less sensitive to nano-topography[24,38]. No damage or change in surface morphology of the PCL film was observed after plasma polymers deposition.

In fig. 4a the results of the *in vitro* tests are shown. Cell viability was evaluated at 24 hrs of culture measuring the metabolic activity by MTT test[44]; the samples with the nucleophilic functionalities showed similar behaviour to untreated PCL, while the amine groups demonstrated a more than fivefold improvement. Electrophilic plasma polymers, in fact, present a similar chemistry with respect to the ester linkage of caprolactone, while in literature it is already known that amine groups enhance fibroblasts proliferation[65]. However, the effect of the nanoparticles was particularly evident, since they improved the viability of one order of magnitude with respect to PCL and of about a factor 2 with respect to a flat surface with the same chemistry.

Cell adhesion was evaluated following 15 min of culture. The results (fig. 4a) showed a dependence on both roughness and chemistry. In particular, the roughness allowed to improve the attachment of the cell to the surface even if with negative polarity: for both ammine and carboxylic functionalities, the nano-topography enhanced the adhesion of more than 2-fold, similarly to viability results. This effect is due to the known capability of the filipodia to sense the surface at the nanoscale[16]: as they run into some nano-topographic features wide enough to allow the spatial clustering of 7-8 integrins, focal points can mature leading to cell adhesion to the surface[18]. Viability in this case is a consequence of the

adhesion since, the selected cells need a support for their survival.

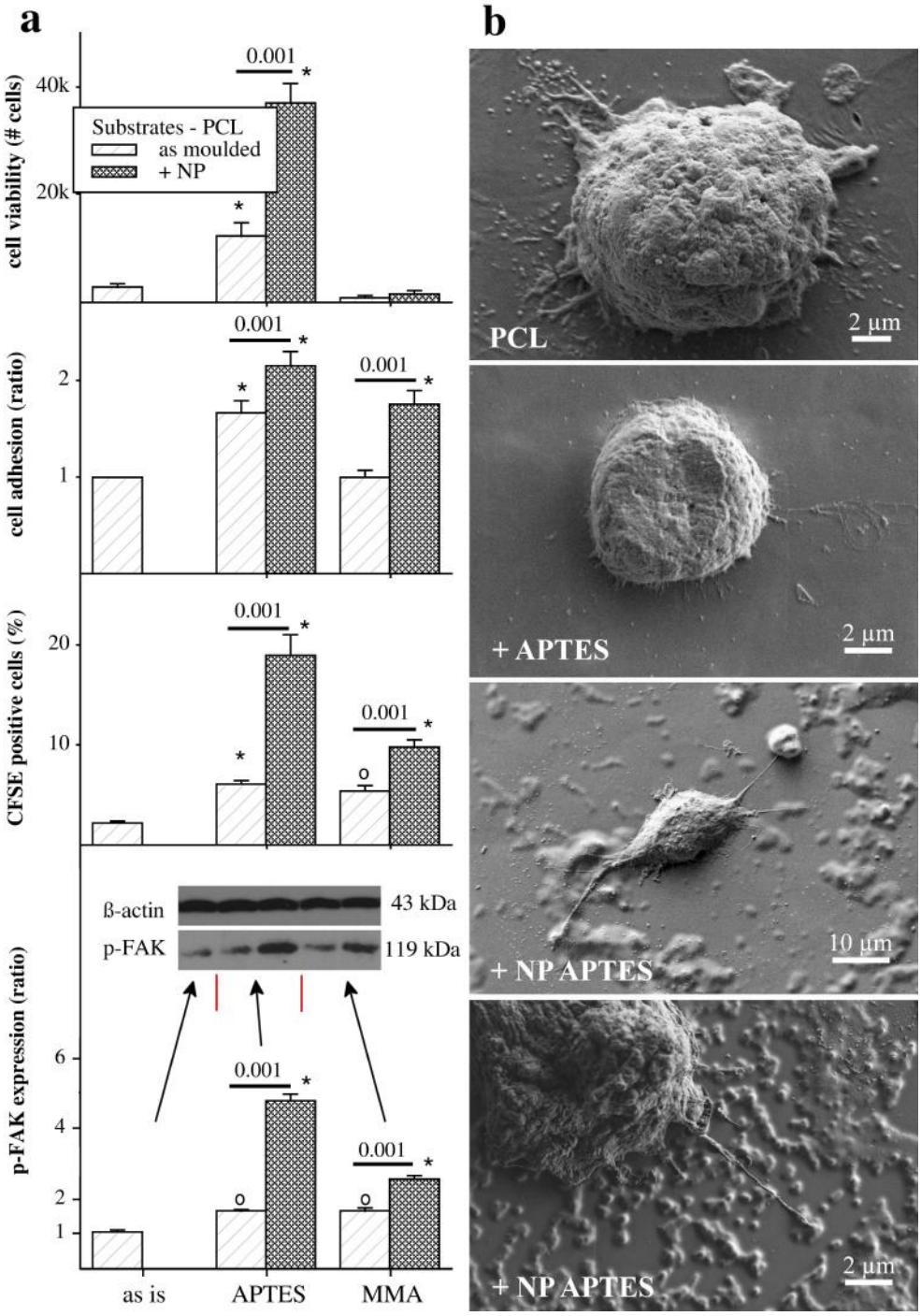

Figure 4: (a) Results of cell viability, cell adhesion (ratio *vs* as is), proliferation, and p-FAK expression (Western blot analysis and quantification, ratio *vs* as is)[66] for PCL films as is, with the plasma polymer coatings and with both plasma polymer and nanoparticles ($*$ 0.001 *vs* as is; $\circ$ 0.01 *vs* as is); (b) SEM images to highlight cells morphologies on PCL films as is, with the amine plasma polymer and with nanoparticle and plasma polymer.

A synergistic behaviour of roughness and chemistry can be observed again for proliferation and p-FAK expression(fig. 4 and Supporting Information figs. S11 and S12). The surface chemistry with both polarities alone led to a small increase in cells growth rate with respect to untreated PCL. As the roughness was added, cells replicated faster on the electrophilic surfaces, where the proliferation increased 3-fold and by a factor 10 compared to the not coated PCL substrate. The addition of the nanoscale morphology, therefore, led always to remarkable improvements, and the effect is probably enhanced by the spherical shape of the asperities that increases the mechanical stress in the focal adhesion points[19]. Mechanotrasduction stimuli generate cytoskeletal tension and effects on the inner cell structure integrating assembly and disassembly of cytoskeletal filaments with transduction of nuclear imputes. Thus, roughness of supports promote expression of proteins (i.e. YAP/TAZ; TWIST1) involved in long time effects and able to modulate the NF-kB pathway for cell survival and proliferation[67]. It has to be considered that the PCL films have been moulded on atomically flat single crystal silicon wafers, where cells are not supported to find preferential anchoring for the maturation of the focal adhesion points. Then, the presence of the nanoparticles is the only source of information for them on the surface apart from PCL chemistry. Conversely, the titanium substrates T1 and T2 were already presenting a certain roughness in the submicron range.

The morphology of the grown cells was observed by SEM (fig. 4b). Cells on the PCL film without the addition of nanostructures appeared in a rounded shape for all the surface chemical functionalities. On the surfaces with the amine groups the filipodia seemed to protrude for longer distances as can be expected also by their higher adhesion. As the nanoparticles were added the cells on the surface with carboxyl groups remained rounded while with the amine groups cells showed a strong elongation with the filipodia sensing the surface for distances comparable to cell size, seeking the nanoroughess for anchoring. It has to be highlighted that even if the nanoparticles on the surface offer to the cells also micrometre scale defects, the focal adhesion point matures on the single nanoparticles

edges. The dimension and density of the fixed nanoparticles on the surface is therefore a key parameter that can be carefully controlled by the current deposition method.

On the two substrates for the two cells type considered the surface chemistry and topography effect partially diverge. The results therefore highlight the importance to control the different parameters separately. Moreover, as pointed out in the introduction, in order to consider the synergistic effect of scaffold properties to guide cells growth, other parameters has to be considered such as, for example, scaffolds geometry and stiffness.

3.4 Fluorophore release

Since the release of active molecules is a relevant topic in medical implants we have investigated the potential use of the nanoparticles as release agents. Nanoparticles can be considered, indeed, not only as building blocks for the nanoscale roughness construction, but also as smart carriers. Inorganic materials for calcium or strontium ions release can substitute the silica nanospheres with no critical risk for process implementation. However, the low temperature of the process allows for even higher flexibility; for example biodegradable polymer particles can be used and if a bioactive molecule was previously embedded then it is released while the polymer dissolves in the body fluids[68]. Since the plasma process do not alter the biopolymer, no change in the active molecule can be expected. In order to verify this statement, we have considered a fluorescent dye as model drug embedded in a bioresorbable polymer. Therefore, PLGA DEGRADEX® particles, with an average diameter of 200 nm and with an embedded red fluorophore was fixed on silicon and titanium T1 surfaces by a plasma polymer starting from MMA precursor. For particles spraying three different dispersions in ethanol were prepared mixing PLGA DEGRADEX® with silica nanoparticles in 3 different concentrations (1%, 10%, 100%).

The deposited coating on the silicon substrate can be observed in fig. 5d. The size distribution of the PLGA nanoparticles was quite wide and goes from micrometre agglomerates (Supporting Information fig. S13) that could not be fixed on the surface by the 150 nm thick

coating, up to submicrometres or tens of nanometres in diameter. The coating appeared always with the same morphology with voids defects close to the particles due to the self-shadowing effect. In the cross section (fig. 5e) a PLGA particle can be distinguished below the coating, demonstrating that it was not damaged or melted during the plasma process.

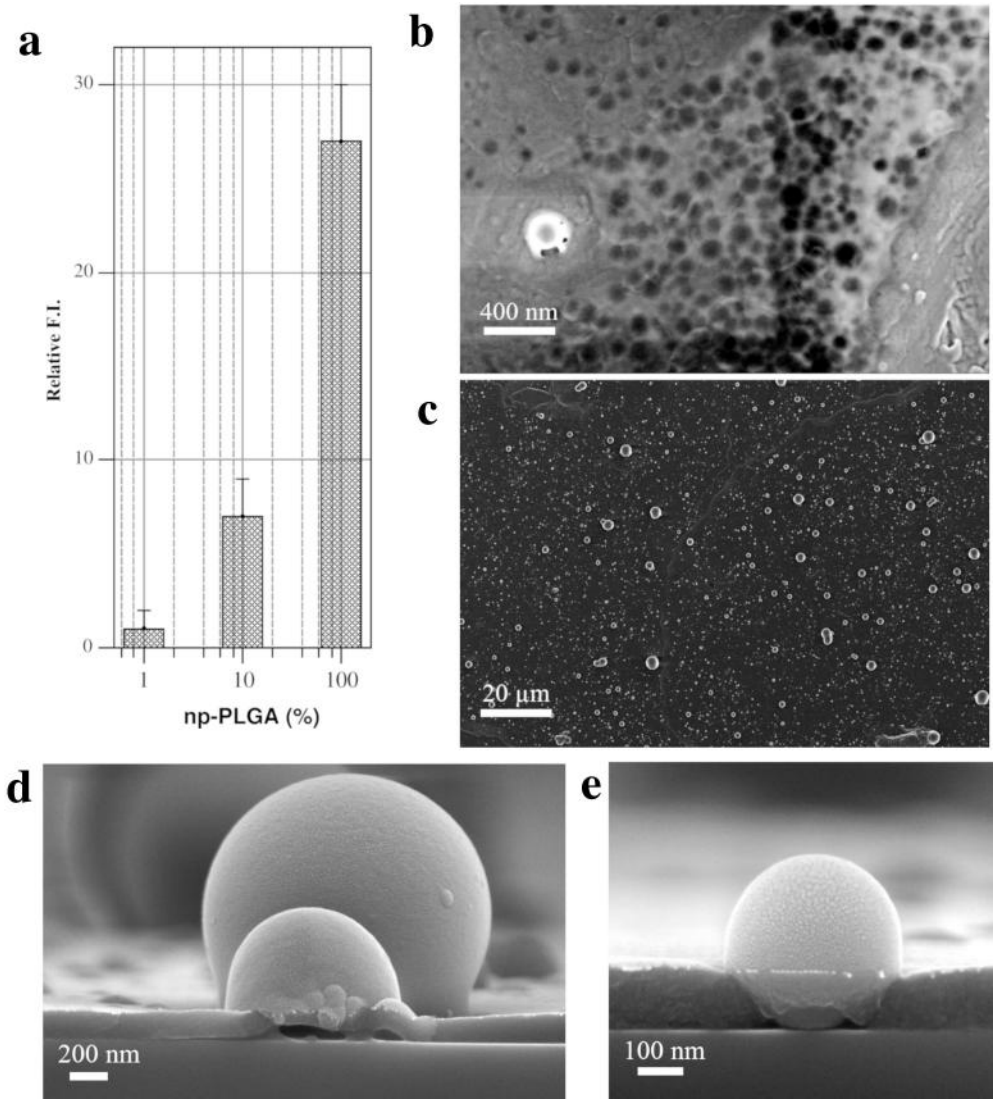

Figure 5: (a) Relative fluorescence intensity (F.I.) in the solution after 5 days of osteoblasts culture spraying by aerosol the silica and PLGA nanoparticles in the same dispersion at different concentrations; (b,c) SEM images of the T1 surfaces with 100% PLGA particles fixed with an MMA layer, before (c) and after (b) the cells culture. (d,e) SEM cross sections of the coating with the PLGA particles which highlighted the presence of the particles under the domes (e) and the voids in the fixing coating due to particle shadowing (d).

The coated T1 substrate was used for pre-osteoblastic murine cells line culture for 5

days and then it was washed. The PLGA nanoparticles are characterised by a two-step mechanism of release of the active molecules: a burst release that is related to random scission of the polymer chains followed by diffusion and a slower process due to polymer dissolution. Therefore, for both mechanisms, the smaller is the particle faster is the delivery. For the few hundreds nanometre size a burst release can be expected within the first 100 hours[69]. In order to verify the release, in our case, the fluorophore was quantified in the culture media for the three different concentrations. The results, presented in fig. 5a, did not scale exactly as a function of the PLGA particles concentration. This behaviour was probably due to the presence of micro-aggregates, which were most likely at higher concentrations and could not be fixed on the surface efficiently. The expected trend and fluorophore integrity was maintained demonstrating the protective role of the polymer.

The surface of the sample was observed by SEM before and after cell culture. In fig. 5b,c the images relative to the sample with PLGA concentration of 100% are presented. In the top view of the as deposited surface (fig. 5c) the macro-aggregates of several micrometres as well as the smaller particles randomly distributed are clearly visible. After cell culture (fig. 5b), the particles were dissolved in the fluids and on voids in the coating remained on the surface of the T1 substrates. In the image 5b an isolated nanoparticle with still the top coating can be observed, at the same time presenting some opening in correspondence with the voids defects around the particle. The image therefore suggests that those voids are the preferential communication path between the embedded nanoparticles and the culture media. Their closing by increasing the coating thickness could therefore allow for a permeability control, slowing down the delivery of the active molecules.

4 Conclusions

A process to deposit a nanostructured coating is presented and here validated as proof-of-concept, demonstrating the possibility to design the surface feature of scaffolds and implants

facing independently nanoscale roughness, surface chemistry and delivery of active agents.

The procedure consists of two steps: an aerosol deposition of a nanoparticles dispersion, followed by a plasma polymer coating using an APPJ. The deposition of the nanoparticles was used to control the nanoscale morphology on flat and micro-rough surfaces: size and density of the nanoparticle deposited can be used to tune spatial and vertical roughness. The APPJ deposits a stable fixing layer that envelops the particles and at the same time offers the desired surface functionalities, depending on the chosen precursor. The fixing coating presented a void defect around the particle, which can be used as preferential path for the release of active agents if nanocarriers are used to create the roughness.

The proposed approach was tested *in vitro* with pre-osteoblasts murine cells line on titanium substrates for potential applications as dental implants and with skin fibroblasts on PCL for tissue regeneration scaffolds. On titanium substrates, it was possible to highlight the synergistic effect of chemistry and morphology in enhancing the selectivity of the differentiation process. In particular, cell proliferation was enhanced of about the 20% even compared to SBAE substrate. Also, on the PCL substrates the nanostructures improved fibroblasts adhesion. The possibility to control morphology and chemistry separately on PCL allowed to highlight their synergistic effect with elongated cells seeking the nanoparticles for the focal adhesion maturation on amino functionalities. This synergistic effect enhanced fibroblast proliferation by a factor 10 compared to smooth PCL substrate. In the end, bioresorbable fluorescent nanoparticles was usedto create the roughness and the release of their undamaged red fluorophore in the cell culture media was checked.

The nanostructured coating deposition is performed at temperatures lower than 50°C; it allows for the treatment of sensitive material, such as biopolymers, but also similarly on a variety of substrates including metallic, polymeric, insulating or conductive.

The technology can be further developed, for example, by the introduction in the resorbable nanoparticles active molecules such as growth factors or antibiotics or by the implementation of conductive coatings deposited by APPJ. From the technological point of

view, the here proposed deposition process is based on spraying technologies at an equivalent rate of 0.5 mm·s^{-1}, which is compatible with the speed range of the extrusion printing technologies (0.01-150 mm·s^{-1})[70]. Our approach can therefore be potentially implemented on 3D systems in order to obtain a localised control of roughness, surface chemistry and drug release.

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Supporting Information Available

The following file is available free of charge.

- `SI_Nanobio.pdf`: AFM of the untreated substrates, APPJ scheme and description, plasma process temperature and electrical characterisation, silica nanoparticles SEM and size distribution, plasma process parameters for the coatings deposition, cross section SEM images of the nanostructured coating, AFM surface morphologies of the nanostructured coatings, SEM images of the filipodia on the different substrates, cell morphology by SEM images on T1 substrates, fibroblast proliferation assays, size distribution of PLGA DEGRADEX® particles.

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