Carbohydrate Polymers 184 (2018) 408–417

Contents lists available at ScienceDirect

Carbohydrate Polymers

journal homepage: www.elsevier.com/locate/carbpol

Freeze dried chitosan acetate dressings with glycosaminoglycans and traxenamic acid

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ARTICLE INFO

Keywords:
Chitosan
Glycosaminoglycans
Tranexamic acid
Hemostasis
Wound healing

ABSTRACT

Bleeding control plays an important role to increase survival in the early phase after a traumatic event.

The aim of present work was the development of hemostatic sponge-like dressings based on chitosan, in association with glycosaminoglycans (GAG) (chondroitin sulfate or hyaluronic acid) and the improvement of their hemostatic performance by loading tranexamic acid (TA). The dressings were prepared by lyophilizazion and were characterized for mechanical, hydration, bioadhesion properties and morphology. Moreover, FTIR analysis was performed to understand the interactions between the different polyelectrolytes present in the dressings. Clotting was investigated *in vitro* by using rat whole blood. Moreover, *in vitro* biocompatibility and proliferation were evaluated towards fibroblasts. *Ex vivo* proliferation properties were assessed by using human skin.

All the dressings were characterised by mechanical, hydration and bioadhesion properties suitable to be applied on bleeding wounds and to absorb bleeding or wound exudate, avoiding tissue dehydration. TA release was fast; TA and chitosan showed a synergic effect to speed up clotting. The dressings were biocompatible and able to sustain cell proliferation *in vitro* and *ex vivo* in human skin. In conclusion, sponge-like dressings based on chitosan and GAG and loaded with TA are an effective tool to enhance hemostasis and healing in bleeding wounds.

1. Introduction

Since 2001 (at the beginning of Afghanistan and Iraq military conflicts), many medical advances in military trauma care have been made to decrease morbidity and mortality and in particular, efforts have been made to control internal and external hemorrhage. However massive bleeding remains the leading cause of combat death and the second leading cause of death after traumatic brain injury in the civilian sector (Bennet, 2017; Champion, Bellamy, Roberts, & Leppaniemi, 2003). Moreover, in surgery the reduction of bleeding can positively influence patient's prognosis (Li et al., 2016). Thus, bleeding control plays an important role not only to increase survival but also to prevent the development of multiple organ failure in the early phase after a traumatic event (Shields & Crowley, 2014).

Advanced topical hemostatic devices (including bandages, fibrin glue, liquids, powders, gels, and scaffolds – generally referred to as dressings) have been developed to control hemorrhages, to reduce mortality and prevent further complications (Seon et al., 2017). These dressings can be grouped into three classes by mechanism of action: including blood factor concentrators, which concentrate blood cellular and protein components to promote clot formation, bioadhesive agents having strong adherence to tissues and procoagulant supplementary, which provides local concentration of procoagulat blood factors (Granville-Chapman, Jacobs, & Midwinter, 2011). Furthermore, dressings cover wounds preventing risk of infections (Seon et al., 2017).

Many types of natural polysaccharides have been proposed as dressing biomaterials. In particular, chitosan (CH), a linear cationic polysaccharide based on glucosamine and *N*-acetyl glucosamine, obtained by partial chitin deacetylation, is well known as hemostatic agent. CH is able to enhance coagulation, by both hemagglutination

Abbreviations: CH, chitosan; GAGs, glycosaminoglycans; ECM, extracellular matrix; HA, hyaluronic acid; CS, chondroitin sulfate; TA, tranexamic acid; GM, growth medium; g, glycerol; m, mannitol; I_s, absorbance of sample; I_r, absorbance of reference value; NHDFs, normal human dermal fibroblasts from juvenile foreskin; FCS, foetal calf serum; MTT, (3-(4,5-dimethylthiazol-2-yl)-2,5-diphenyltetrazolium bromide); PEC, polyelectrolyte complex

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https://doi.org/10.1016/j.carbpol.2017.12.066
Received 9 August 2017; Received in revised form 17 November 2017; Accepted 24 December 2017
Available online 28 December 2017
0144-8617/

and promoting platelet activation. In presence of blood, CH forms a solid clot due to thrombin and platelet activation, fibrin polymerization and fibrin fibers cross-linking by FXIIIa (Marchand, Rivard, Sun, & Hoemann, 2009; Iliescu, Hoemann, Shive, Chenite, & Buschmann, 2008) and CH is able to stabilize clot structure preventing its lysis (Hoemann et al., 2007). Moreover, it possesses other crucial properties fundamental in wound healing: it is biodegradable and characterized by antimicrobial properties. Furthermore, CH supports extracellular matrix regeneration stimulating granulation tissue formation, activating fibroblasts (Francesko & Tzanov, 2011) and enhancing their proliferation and migration (Sandri et al., 2011) in the early phases of healing. Moreover, it is able to enhance remodeling process increasing also angiogenesis.

Glycosaminoglycans (GAGs) play a crucial role in different stages of skin tissue regeneration and maturation, being important components of extracellular matrix (ECM). Moreover, GAGs are able to bind proteins including several chemokines and growth factors. The polysaccharidic structure of glycosaminoglycans as well as the presence, position and number of sulfate groups within the polymer chain plays an important role on these GAG-protein interactions. Among glycosaminoglycans, chondroitin sulfate (CS) and hyaluronic acid (HA), are naturally present in mammals and have an important role in the healing process.

CS is a sulfated polysaccharide based on *N*-acetylgalactosamine-glucuronic acid disaccharide units. It is able to interact with different important positively charged biological molecules, such as growth factors, chemokines, cytokines and adhesion molecules, involved in tissue healing and it prolongs their activities by charge–charge interactions, avoiding their enzymatic degradation (Silbert & Sugumaran, 2002; Yamada & Sugahara, 2008). HA is a linear polysaccharide composed of a dimeric repeating unit of D-glucuronic acid and *N*-acetyl-D-glucosamine and, during the tissue injuries, it takes part to the wound healing and coagulation process forming, together with fibrin, a matrix support for fibroblast migration and proliferation (Chen & Abatangelo, 1999; Jiang, Liang, & Noble, 2007).

The polysaccharides based systems should be more biocompatible rather than protein ones. In particular collagen, one of the major components of ECM seems the best candidate to obtain dressings/scaffolds for wound reparation but in literature it is reported as immunogenic material due to possible helical-recognition by antibodies, due to 3D intact triple helix conformation, peculiar aminoacid sequence able to start antibody recognition, presence of non-helical terminal regions (telopeptides) (Lynn, Yannas, & Bonfield, 2004; Olsen et al., 2003). Moreover, it could have concerns of species-to-species transmissible diseases (xenozoonoses) (Cataldo, Ursini, Lilla, & Angelini, 2008).

Given this premises, the aim of present work was the development of hemostatic sponge-like dressings based on chitosan (CH), in association with chondroitin sulfate (CS) and hyaluronic acid (HA) and loaded with tranexamic acid (TA), to be used in case of massive bleeding.

Tranexamic acid (TA) is a synthetic derivative of the aminoacid lysine and acts as antifibrinolytic agent by reversibly blocking lysine sites on plasminogen molecules (Dunn & Goa, 1999). In particular, TA is able to inhibit fibrinolysis by displacing plasminogen from fibrin and by avoiding its degradation (Ker, Edwards, Rerei, Shakur, & Roberts, 2012).

Sponge-like dressings should have advantages in application with respect to powdery hemostatic agents, especially in the treatment of intra-abdominal solid organ bleedings or injuries: if hemostatic agents are based on granules (microparticles) can leave residue in the lumen of the vessel and may occlude distal arterial flow. Moreover intraluminal dissemination of the clot, resulting in distal thrombosis, may occur (Khoshmohabat, Paydar, Kazemi, & Dalfardi, 2016).

TA loaded sponge-like dressings were prepared by freeze-drying and were characterized by hydration, mechanical, bioadhesion properties and morphology. Moreover, FTIR analysis was performed to understand the interactions between the different polyelectrolytes present in the dressing. The release profile of TA was evaluated in isotonic solution (NaCl 0.9%) by using Franz diffusion cells. The hemostatic performance of the systems was evaluated by means of a dynamic whole-blood clotting test. Furthermore, *in vitro* cytocompatibility and proliferation tests were assessed by using NHDFs (normal human dermal fibroblasts from juvenile foreskin). Finally, proliferation properties of TA loaded sponge-like dressings were evaluated by means of an *ex vivo* test on human skin.

2. Experimental part

2.1. Materials

The following materials were used: Chitosan (CH) low MW 251000 Da, deacetylation degree 98%, maximum charge density (number of positively charged functional groups per repeated unit): 0.98 (ChitoClear, Siiiglufjordur-Iceland); Hyaluronic Acid (HA) low MW 212000 Da; maximum charge density (number of negatively charged functional groups per repeated unit): 0.5 (Bioiberica, Barenz, Italy); Chondroitin sodium sulfate bovine 100 EP (CS) low MW 14000 Da, mixture of A (chondroitin 4 sulfate) and C (chondroitin 6 sulfate); maximum charge density (number of negatively charged functional groups per repeated unit): 1 (Bioiberica, Barenz, Italy); *trans*-4 (amino-methyl) cyclohexanecarboxyilic acid (Sigma Aldrich, Italy); Glycerol (g) 30° Be (Carlo Erba, Italy); D-Mannitol (Fluka, France).

2.2. Methods

2.2.1. Sponge-like dressing preparation

CH was hydrated in acetic acid 1% w/w (glacial acetic acid, Sigma Aldrich, Italy) under gentle stirring at room temperature. Acetate was used as counterion to obtain chitosan salification in mild pH conditions and it was preferred to hydrochloric acid because chitosan fragmentation could occur in HCl environment also at low molarity (Sabnis & Block, 2000).

HA and CS were hydrated in distilled water under gentle stirring at room temperature. CH and CS or HA polymeric solutions were mixed 1:1 weight ratio to obtain polymeric mixtures. Systems containing CH alone were also prepared in distilled water. Final acetic acid concentration in the polymeric systems was 0.5% w/w. CH 1 – HA and CH 1 – CS mixtures and CH 1, were subjected to zeta potential evaluation. Measurements were carried out at 25 °C by means of a Malvern Zetasizer Nano ZS90 (Malvern Ltd., UK). HA and CS solutions, prepared in the same concentrations used for CH 1 – HA and CH 1 – CS mixtures but without CH were also tested.

In two series of preparations of CH 1 – HA and CH 1 – CS mixtures and CH 1, glycerol or mannitol were added as lyoprotectants. Polymeric solutions appeared transparent, free from visible precipitate. As for TA loaded systems, drug was added to the polymeric mixture at 0.75% w/w. Two ml of each mixture were poured into each well of 12-well plate (well area 3.8 cm^2). In the TA loaded systems the drug concentration was 4 mg/cm^2. All the polymeric solutions were frozen at −40 °C overnight and freeze-dried (Heto 15, Analitica De Mori, Italy) for 24 h. All the dressings prepared were 5 mm thick (sd ± 0.52).

CH, HA and CS were at 5 mg/cm^2 concentration in CH 1 dressings; CH was at 10 mg/cm^2 concentration in CH 2 dressings, glycerol was at 1.3 mg/cm^2 while mannitol was at 10.5 mg/cm^2 concentration.

Dressings were re-solubilized in distilled water to check the eventual residues of acetic acid (2 g of distilled water per each dressing). pH of each sample was measured by means of a pHmeter (pH 210 Microprocessor, Hanna Instruments, Italy) and the acetic acid excess was titrated by means of acid-base titration, by using NaOH 0.1 M (Carlo Erba, Italy) and phenolphthatelin as indicator.

2.2.2. Sponge-like dressing characterization

2.2.2.1. Penetrometry measurements. Sponge-like dressings were subjected to penetrometry measurements by means of a texture analyzer (TA.XT plus, Stable Microsystems, ENCO, Italy), equipped with a cylinder probe (10 mm, P10), to evaluate the resistance to compression. Each dressing was placed onto the apparatus base and the probe was lowered at 1 mm/s speed. Maximum force of resistance to compression (Fmax) was determined as the maximum force required to penetrate into the sponge up to 2.5 mm.

2.2.2.2. Hydration measurements. Dressings were placed on a 0.45 μm membrane (HA, Millipore, Italy) that covered a container filled with pH 7.2 phosphate buffer (USP), simulating wound exudate. The liquid could freely cross the membrane dependently on the capability of each system to absorb liquid. At dry state (before the beginning if the hydration test) and at prefixed times during hydration (after the contact between each dressing and the membrane), the dressings were weighted and the amount of liquid absorbed was normalized by the weight of the dried dressing.

2.2.2.3. Bioadhesion measurements. Bioadhesion measurements were performed using texture analyzer (TA.XT plus, Stable Microsystems, ENCO, Italy) equipped with a 1 kg load cell, a cylinder probe having a diameter of 10 mm (P10) and the measuring system A/MUC (adhesion test system) (Sandri et al., 2013; Sandri et al., 2012; Szucs et al., 2008). The measuring system A/MUC consists of a support in which a biological substrate can be fixed. In this case, the biological substrate was egg shell membrane (Tao et al., 2009), chosen to mimic damaged skin, wetted with 100 μl of isotonic saline solution (NaCl 0.9% w/v). To obtain the membrane, the egg shell was placed in a 0.5 M HCl solution for 1 h. Sponge like dressings, having diameter of 10 mm, were stuck with cyanoacrylate glue to the cylinder probe and hydrated with 100 μl of pH 7.2 phosphate buffer (USP). The sample and the biological substrate were put in contact under a preload of 6 N for 3 min. The cylinder probe was then moved upward at a prefixed speed of 2.5 mm/s up to the complete separation of the bioadhesive interface (egg shell membrane-sample). The maximum force of detachment (Fmax) was recorded as a function of displacement.

Blank measurements were performed by using a filter paper disc wetted with 100 μl of isotonic saline solution (NaCl 0.9% w/v) instead of biological substrate: this allowed to evaluate cohesive properties of the sample. As comparison, two physical mixtures (CH-HApm and CH-CSpm) having the same polymer composition of sponge-like dressings were evaluate. Moreover the bioadhesion of CS and HA as powders (CSp and HAp) were compared to understand the contribution of GAG to bioadhesion. In these latter cases, the same experimental conditions (hydration and sample weight) were used.

2.2.2.4. Morphology. Dressing morphology was analyzed by means of scanning electron microscopy (SEM, Tescan, Mira3XMU, ARVEDI Center, University of Pavia, Italy). Samples were sputtered by means of graphite deposition under vacuum.

2.2.2.5. FT-IR measurements. FT-IR spectra were obtained using a Nicolet FT-IR iS10 Spectrometer (Nicolet, Madison, WI, USA) equipped with ATR (Attenuated Total Reflectance) sampling accessory (Smart iTR with ZnSe plate) by co-adding 256 scans in the 4000–650 cm^{-1} range at 4 cm^{-1} resolution.

2.2.2.6. TA release measurements. TA release measurements were performed *in vitro* by using Franz cells (vertical glass diffusion cell, orifice diameter 20 mm, PermeGear Inc., USA). A cellulose acetate membrane (pore size 0.45 μm, diameter 25 mm) was placed between the donor and the receptor chambers. The cells were thermostatedat 32 °C by means of water jackets. Isotonic solution (NaCl 0.9% w/v) was used as receiving phase. Each dressing was placed in the donor chamber and wetted with 100 μl of isotonic solution. At prefixed times, 500 μl of receiving phase were collected and the amount of drug released was assayed by means of a spectrophotometric method (Ansari, Raza, & Rehman, 2005). This method is based on the colorimetric detection of the Ruhemann's purple resulting from ninhydrin and TA reaction via oxidation deamination of TA primary amino group, followed by the condensation of the reduced ninhydrin in the basic medium at pH 8.0. TA stock solution was prepared in NaCl 0.9% w/v at 1 mg/ml concentration and a calibration curve was prepared with TA concentrations ranging from 1 mg/ml to 0.1 mg/ml. 1 ml of ninhydrin reagent (0.2% w/w ninhydrin (Carlo Erba, Italy) in methanol) and 0.5 ml of 20 mM phosphate buffer at pH 8.0 were added to 200 μl of each sample. Samples were heated at 90 °C for 20 min in a shaking bath, subsequently they were cooled at room temperature and their absorbance was read at 565 nm wavelength (Lamba 25, Perkin Elmer, Italy). The method was linear in the concentration range from 1 mg/ml to 0.1 mg/ml with R^2 higher than 0.995.

2.2.2.7. In vitro dynamic whole-blood clotting. The hemostatic activity of sponge-like dressings was evaluated using rat whole blood pooled from 6 male rats (Wistar 200–250 g). All animal experiments were carried out in full compliance with the standard international ethical guidelines (European Communities Council Directive 86/609/EEC) and approved by Italian Health Ministry (D.L. 116/92). The study protocol was approved by the Local Institutional Ethics Committee of the University of Pavia for the use of animals. The whole blood was supplemented with 10% v/v of acid-citrate-dextrose (ACDC, 38 mM citric acid/75 mM trisodum citrate/100 mM dextrose) to avoid the coagulation during the storage at 4 °C (Quan et al., 2015; Ong, Wu, Moochhala, Tan, & Lu, 2008; Sudheesh Kumar et al., 2012). Just before experiments, the anticoagulant activity of ACDC was inhibited by adding 100 μl of saturated $CaCl_2$ solution to 50 μl of blood.

50 μl volume of fresh blood was dropped onto unloaded and TA loaded sponge dressings (5 mm in diameter) placed in 50 ml glass beaker. After prefixed contact times, 10 ml distilled water was slowly poured in beaker without disturbing the clotted blood. The beaker was gently shaken for 2 min to suspend free red blood cells, not entrapped in the clot. The absorbance of each resulting sample, was assayed by means of spectrophotometric detection at 542 nm wavelength (Lamba 25, Perkin Elmer, Italy). As reference, fresh blood was considered.

The content of hemoglobin was quantified by the following equation:

$$\text{Hemoglobin absorbance } \% = I_s/I_r \times 100$$

where I_s is the absorbance of the resulting sample, and I_r is the absorbance of the reference (blood or blood/TA (Ong et al., 2008; Sudheesh Kumar et al., 2012)).

2.2.2.8. In vitro biocompatibility and proliferation. NHDFs (normal human dermal fibroblasts from juvenile foreskin, Promocell GmbH, Germany) were used between the 2^{nd} and 5^{th} passage, for all the experiments.

Fibroblasts were grown in presence of Dulbecco's modified Eagle medium (Sigma, Italy) and supplemented with 10% fetal calf serum (FCS, Euroclone, Italy) with 200 IU/ml penicillin, and with 0.2 mg/ml streptomycin (Sigma, Italy) and kept at 37° C in a 5% CO_2 atmosphere with 95% relative humidity (RH).

Fibroblasts were seeded in each well of 96-well plates (area 0.34 cm^2) at a seeding density of 10^5 cells/cm^2. Cells were grown 24 h to obtain sub-confluence.

3.5×10^4 cells/well (area 0.34 cm^2) were seeded in a 96-well plate and grown to confluence for 24 h.

After 24 h cells were washed with saline solution and placed in contact with 200 μl of CH, CH–CS and CH–HA polymeric mixtures (prepared as reported in the section 2.2.1.) at 1:20 (CH, CS or HA at

500 µg/ml, solubilized in cell growth medium) and 1:50 dilution with growth medium (GM) (CH, CS or HA at 200 µg/ml). After dilution, all the samples were transparent without visible precipitate. TA having the same concentration as in the loaded polymeric mixtures and GM were used as control. Cell substrates were incubated for 3 h (biocompatibility test) and for 24 h (proliferation test) with the samples, then the medium was removed and the MTT test was performed. Briefly, MTT test is based on the activity of mitochondrial dehydrogenases of vital cells that convert MTT in formazan (Sandri et al., 2017). At this purpose, 50 µl of MTT solution (Sigma Aldrich, Italy) at 2.5 mg/ml concentration in HBSS (Hank's Buffered Salt Solution) pH 7.4 was put in contact with each cell substrate for 3 h. The reagent was removed from each well, and the substrates were washed with 200 µl of PBS. After the removal of PBS, 100 µl of DMSO was put in each well, and the absorbance was assayed at 570 nm by means of an ELISA plate reader (Imark Absorbance Reader, Biorad, Italy), with a reference wavelength of 690 nm. Cell viability was calculated as % ratio between the absorbance of each sample and the absorbance of cell substrate maintained in contact with growth medium.

2.3.1. Ex vivo proliferation on human skin biopsy. To analyze the effects of dressings on wound healing, an *ex vivo* human skin model was used to reproduce *in vivo* physiological conditions (Mori et al., 2016). Adult healthy human skin biopsies were clinically obtained from surgery for breast reduction of different age donors (range 35–60 years old), after obtaining informed consent. Surgical biopsy (about 4 × 8 cm) was cut into smaller fragments (about 0.6 × 0.6 cm) and a skin circular portion (including epidermis and dermis) was centrally removed with aseptic circular punch (diameter: 3 mm). Each fragment was placed into Transwell® inserts (growing area: 0.33 cm^2, Corning® Costar®, Sigma, Italy) in 24-well-plate (membrane pore size: 0.4 µm) and circular portions of each dressing, having the same diameter of punch, were applied in correspondence of skin lesion.

The samples were maintained in culture in presence of Dulbecco's modified Eagle medium (Sigma, Italy) and supplemented with 10% fetal calf serum (FCS, Euroclone, Italy) with 200 IU/ml penicillin, and with 0.2 mg/ml streptomycin (Sigma, Italy) and kept at 37 °C in a 5% CO_2 atmosphere with 95% relative humidity (RH).

The medium (900 µl/each well) was added into well basal compartment to maintain skin viability, avoiding skin dehydration and ensuring an adequate supply of nutrients through hypodermis. After different times (24 and 72 h and 7 days) of culture, the biopsies were fixed with 4% paraformaldehyde in 0.1 M phosphate buffer, pH 7.4, for 24 h, and processed for histological analysis (Riva et al., 2007). Briefly, the fragments were dehydrated through graded concentrations of ethanol and embedded in paraffin. 8 µm sections were obtained by means of a microtome (Leitz, Germany), rehydrated and processed for immunohistochemical reactions for bromodeoxyuridine (BrdU) incorporation. The sections were incubated with primary antibody anti-BrdU (GE Healthcare, UK) overnight at appropriate dilution (1:100), the primary antibody anti-BrdU was reacted by using MACH 1 Universal HRP-Polymer Detection kit (Biocare Medical, USA). At this purpose the sections were reacted with MACH 1 Mouse probe for 15 min and subsequently with HRP-Polymer (horseradish peroxidase polymer) for 30 min and finally with Biocare's Betazoid DAB for 5 min. Finally, the sections were dehydrated and mounted with DPX mounting medium (a mixture of distyrene, a plasticizer, dissolved in toluene-xylene, Sigma, I).

Skin sections obtained in the middle of the lesions were observed at the magnification of 20× under a light microscope Axiophot (Zeiss, Germany) equipped with a digital camera. Image J program was used to analyze images, counting cell nuclear positivity in a section area larger 2 mm at right and left side of the lesions.

2.3.2. Statistical analysis. Statistical differences were evaluated by means of a non-parametric test: Mann Whitney (Wilcolxon) W test, (Stat Graphics 5.0, Statistical Graphics Corporation, MD, USA). Differences were considered significant at $p < 0.05$; only significant differences are evidenced in the figures as asterisks.

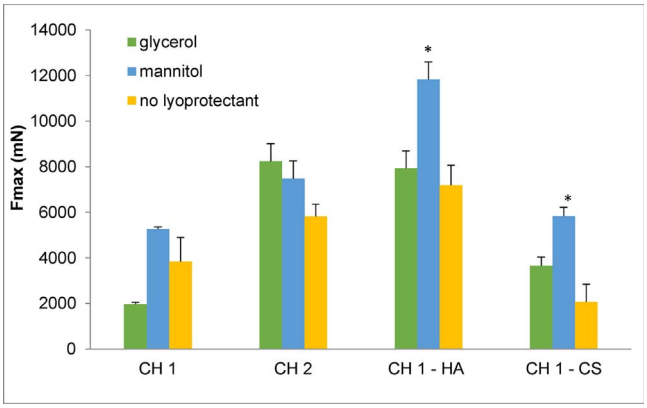

Fig. 1. Mechanical resistance as maximum force of penetration (Fmax, mN) of the dressings without lyoprotectant and with glycerol or mannitol, as lyoprotectants (mean values ± sd; n = 6).

3. Results and discussion

3.1. Sponge-like dressing technological properties

3.1.1. Mechanical properties

Fig. 1 reports the results of mechanical resistance as maximum force of penetration of the dressings without lyoprotectant and with glycerol or mannitol, as lyoprotectants.

The increase of chitosan amount in the dressings increased the mechanical resistance of the formulations: stiffness was significantly greater when lyoprotectants (glycerol or mannitol) were included in the formulations. The presence of HA in the dressings caused a significantly increase in the mechanical resistance and HA based formulation with mannitol, as lyoprotectant, was characterized by the higher Fmax value to indicate a more rigid structure. This behavior could probably due to the high molecular weight of HA, with respect to CS, that could dramatically strengthen the tridimensional network of the system. It is conceivable that HA interacted with chitosan during the preparation procedure, forming a polyelectrolyte complex (PEC) between chitosan and hyaluronate. Since both CH and HA have similar molecular weight the interaction between the two polymers should be able to form a highly crosslinked structure.

On the contrary, the presence of CS in the formulation did not significantly change the stiffness of the dressings. This is probably due to the resulting PEC formed by an interaction between CS and CH: CS is characterized by a molecular weight 18 folds lower than that of CH, moreover, sulfate groups of CS (more acid with respect to carboxylic group) could cause a coiled structure less prone to polymer chain entanglements. This probably rendered the dressing structure less stiff.

As reported in literature, structure and stability of PECs strongly depend on different parameters such as concentrations of each polyelectrolyte, charge molar ratio, charge density, molar masses and chain flexibility but depend also on extrinsic parameters as pH, ionic strength and/or temperature (Kabanov, 2005; Feng, Leduc, & Pelton, 2008; Le Cerf et al., 2014).

According to all these parameters, mixing oppositely charged polyelectrolytes could give two types of PECs dependently of charge stoichiometry: soluble PECs, leading to stable and transparent solutions, and insoluble PECs, with or without precipitation. The soluble PECs are obtained with an excess of anionic or cationic charges (Feng et al., 2008; Kabanov & Zezin, 1984; Schatz, Domard, Viton, Pichot, & Delair, 2004) and this was the case of CH-HA and CH-CS. In fact, charge density (positive) is 0.006 mol/g for chitosan, 0.0013 mol/g for HA

(negative) and 0.002 mol/g for CS (negative) with a clear excess of positive charges, in both cases. The charge molar ratio of CH–HA was 4.6 while that of CH–CS was 3.

Moreover Denunziere, Ferrier, & Domard (1996) reported interaction between CS and CH was independent of CS sulfate substitution on 4 or 6 carbons.

This was confirmed by zeta potential evaluation. CH 1 had a positive zeta potential of 53.2 ± 1.9 (mean value ± sd; n = 3) while HA and CS, as solutions, showed negative zeta potentials of −27.9 ± 0.9 and −16.8 ± 1.2, respectively (mean value ± sd; n = 3). CH 1 – HA and CH 1 – CS were characterized by positive zeta potentials of 39.3 ± 0.3 and 25.3 ± 0.8, respectively (mean value ± sd; n = 3). These results support that the interaction between CH and GAGs did not cause a neutralization of the two polyelectrolytes mixed (CH and HA or CS) and an excess of positive charges was present.

Moreover lyophilization caused acetic acid evaporation. In particular, acetic acid was added in excess, to form chitosan acetate: 350 µmoli of acetic acid were added to 110 µmoli of chitosan per each dressing prepared, starting from chitosan at 1% w/w. The titration of acetic acid residues after lyophilization revealed that there was a residues of 5 ± 1 µmoli per each systems independently of the addition of HA or CS. This is conceivably due to the almost total removal of acetic acid in excess (not involved in chitosan salification).

3.1.2. Hydration properties

Fig. 2 reports buffer (pH 7.2 phosphate buffer, to mimic wound exudates) taken up as function of time for all the dressings prepared: (a) without lyoprotectant, (b) with glycerol and (c) with mannitol, as lyoprotectans.

Considering all the compositions with and without lyoprotectants, chitosan based dressings were characterized by higher buffer taken up vs time profile with respect to those based on mixtures of chitosan and GAG; moreover chitosan dressings based on higher CH amount, showed higher hydration profile even not significantly different respect to hydration capability of dressing based on lower CH amount. Dressings based on mixtures of chitosan and GAG were characterized by significantly lower hydration profiles, independently of both the presence and type of lyoprotectants. All the systems showed a high hydration capacity in the first hour: it is conceivable that chitosan was likely to create a lower pH environment at the dressing/liquid interface and this caused a high liquid absorption probably due chitosan gelation. Subsequently, chitosan poor solubility at neutral pH (pKa 6.5) prevailed over its buffering properties and liquid absorption stopped (Rossi, Sandri, Ferrari, Bonferoni, & Caramella, 2003a,b). The presence of GAG further decreased system hydration probably due to the interpolymer interaction between chitosan and hyaluronic acid or chondroitin sulfate (Rossi et al., 2003a,b).

Glycerol did not markedly change the hydration properties of CH and CH in association with GAG. On the contrary, mannitol substantially decreased the hydration capacity of the systems: this could be related to stiff structure of the dressings, which could slow down and impair liquid penetration into the dressings.

Hydration behavior is a crucial point in dressings to ensure hemostasis, to absorb wound exudate and to avoid wound bed dehydration, enhancing granulation phase and healing.

3.1.3. Bioadhesion properties

Fig. 3 reports maximum force of bioadhesion (mN) for all the dressings prepared: (a) without lyoprotectant, (b) with glycerol and (c) with mannitol, as lyoprotectans.

As for dressings without lyoprotectants (Fig. 3a), chitosan concentration influenced bioadhesive behavior: CH 2 sponge-like dressing (containing higher chitosan amount) did not show bioadhesive propensity (Fmax measured without biological substrate was not

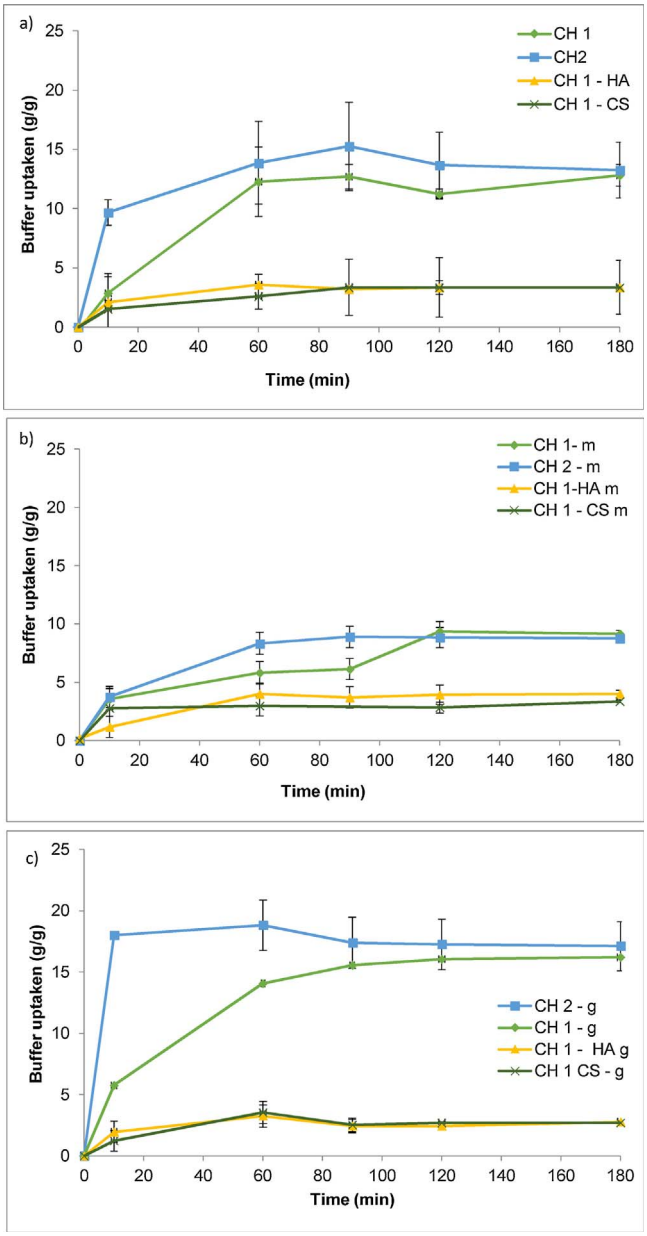

Fig. 2. Liquid taken up (pH 7.2 phosphate buffer) as function of time for all the dressings prepared: (a) without lyoprotectant, (b) with mannitol and (c) with glycerol, as lyoprotectans (mean values ± sd; n = 6).

significantly different with respect to the value in presence of biological substrate – egg shell membrane) while CH 1 dressing (containing chitosan at lower amount) was characterized by good bioadhesive properties. Such a behavior is probably due to a different degree of entanglement between chitosan polymeric chains. In particular, the lowest chitosan concentration the lowest the entanglement and the highest the capability to form a bioadhesive joint with the biological substrate (Rossi, Ferrari, Bonferoni, & Caramella, 2001).

Since zeta potential analysis confirmed that there was not a complete neutralization of CH with GAGs and both CH-HA and CH-CS had positive zeta potentials, the presence of GAG in dressings significantly enhanced the bioadhesive properties. The interaction between chitosan and GAG probably assisted bioadhesive joint consolidation, considering a higher concentration of carboxylic and OH residues in these systems compared to that of CH 1. The highest presence of these groups able to

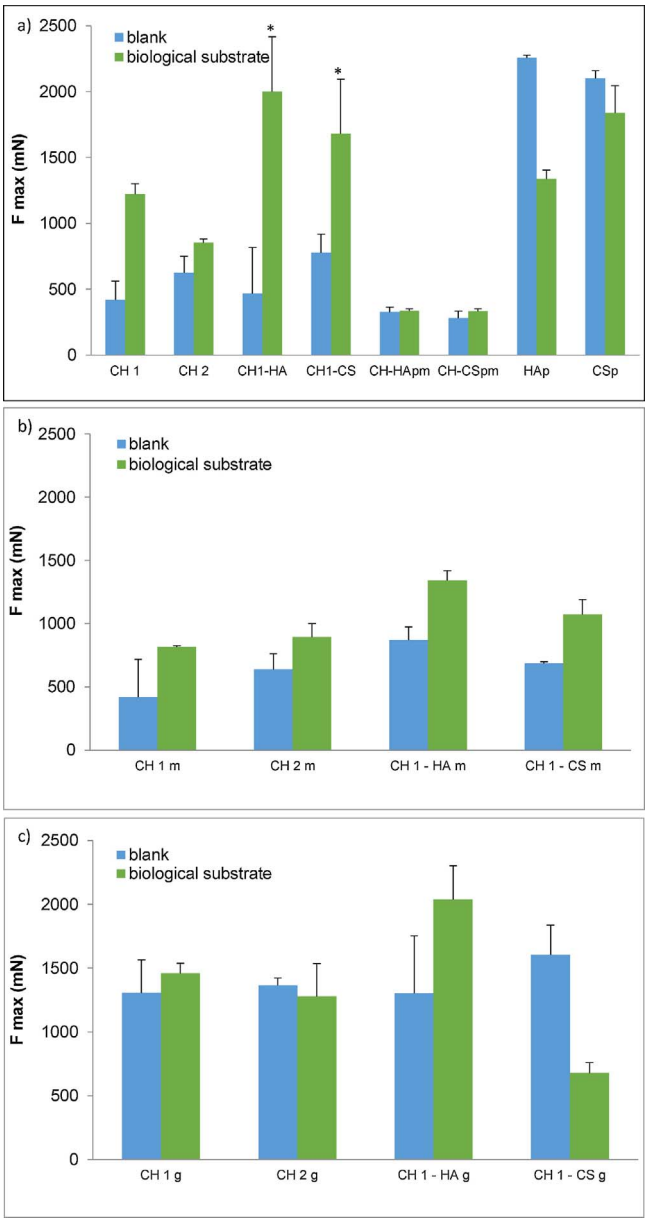

Fig. 3. Maximum force of bioadhesion, Fmax (mN) of all the dressings prepared: (a) without lyoprotectant, (b) with glycerol and (c) with mannitol, as lyoprotectans (mean values ± sd; n = 6).

form hydrogen bonds could consolidate bioadhesive properties of chitosan, that are mainly due to its positive charge density. Chitosan and GAG physical mixtures (CH-HApm, and CH-CSpm) did not show bioadhesive properties: these are easily inferred by the negligible solubility properties of chitosan at pH 7.2 phosphate buffer (hydration medium chosen to mimic wound bed exudate). Moreover CS or HA, as powders, were also considered and they showed a lack of bioadhesion in those conditions: this was probably due to slower hydration of HA during the test with respect to the lyophilized dressings and to the poor bioadhesive strenght of CS mainly caused by its low molecular weight.

In presence of lyoprotectans, both mannitol and glycerol (Fig. 3b and c, respectively), the bioadhesive properties were less pronounced and were significant only for HA and CS dressings with mannitol. The presence of lyoprotectans, especially of mannitol, that increased system stiffness and probably decreased polymer chain mobility, could impair polymer capability to interact with the biological substrate.

Also bioadhesion is a crucial characteristic that should favor an intimate and prolonged contact between wound dressings and lesion, favoring blood absorption and avoiding formulation detachment to increase the hemostatic potential.

3.1.4. Morphology

To understand if the system tridimensional networks could affect hydration and bioadhesion properties, morphology was assessed.

Fig. 4 reports SEM images of transversal sections of CH 1, CH 1- HA and CH 1 – CS unloaded dressings and TA loaded dressings without lyoprotectans, and CH 1, CH 1- HA and CH 1 – CS dressings with glycerol and with mannitol.

CH 1 dressing showed a beehive structure with polyhedric cavities interconnected by pores having oval or round shapes. The cavities had dimensions of about 500 μm while the pores ranged from 50 to 200 μm. The presence of GAG did not change the system morphology while it markedly decreased the cavity dimensions that were around 200–300 μm and furthermore the pore diameters that were of about 50 μm. This difference in system porosity conceivably contributed to their hydration behavior. The TA loading did not modify the structure of the dressings.

The presence of glycerol as lyoprotectant did not substantially change dressing structure while the presence of mannitol caused a partially loss of polyhedric cavities and the structure resembled much more randomly organized sheets without marked modifications of the porosity. This modification could be responsible to the system behavior upon hydration and bioadhesion.

Since lyoprotectans did not markedly improve the technological properties of the dressings and considering mechanical, hydration and bioadhesive properties, lyoprotectant free systems prepared using chitosan at 1% w/w were considered for the further characterizations.

3.1.5. FT-IR analysis

Fig. 5 reports the FT-IR spectra of CH 1, CH 1 – HA and CH 1 – CS dressings and of CH powder (in the inset) as comparison.

The three spectra shared many common features both in the number and position of the peaks and seemed to be dominated by the FT-IR pattern of CH (free CH as powder), where the region around 1630 and 1540 cm^{-1} is typical of the amide I and amide II stretching and bending modes (Radhakumary, Antony, &, Sreenivasan, 2011; Jayakumar, Prabhakaran, Nair, & Tamura, 2010). Interestingly, the position of these peaks shifted to lower frequency with respect to free CH (i.e. not interacting with acetate ions) shown Fig. 5 inset. In particular, the −C═O stretching of the amide I in free CH moved from 1652 to 1634 cm^{-1} when it interacted with acetate ions and it remained in the same position even in the CH 1 – HA and CH 1 – CS systems. Moreover, a new peak around 1404 cm^{-1} (correlated to the −C−H− bending) appeared in the spectra of CH 1 dressing with respect to free CH and was found also in the spectra of CH 1 − HA and CH 1 − CS. The two mentioned spectral features have been correlated to charge–charge neutralization (salification) which most probably occurred between CH and acetate ions and dominated the spectra of all the three systems shown in Fig. 5 (Shanti Krishna, Radhakumary, & Sreenivasan, 2015).

This suggests that, in presence of acetate ions, the charge–charge interaction, involving chitosan amino groups, was preferentially between CH and acetate and therefore the further addition of HA and CS mostly interact with chitosan through less strong bonding such as polar interactions. Such results are in agreement with zeta potential values determined.

3.1.6. In vitro drug release properties

Fig. 6 reports TA release profiles as function of time obtained for TA loaded dressings without lyoprotectans.

All the dressings were characterized by the same release behavior. TA is a water soluble drug (solubility: 167 mg/ml) and considering the experimental conditions, the maximum concentration reached in the receiving phase was 1 mg/ml. In these conditions, TA release from all

the dressings was relatively fast: the 50% of TA was released in about 30 min and reached quantitative and plateau value within about 3 h. This behavior should deliver TA at the application site, in the bleeding lesion, to perform its procoagulant function as quickly as possible and moreover, to sustain hemostatic activity for prolonged time to stabilize clot formation.

3.2. Sponge-like dressing biopharmaceutical properties

3.2.1. In vitro dynamic whole-blood clotting

Fig. 7 shows % hemoglobin absorbance profiles as function of time evaluated for (a) unloaded dressings and (b) TA loaded formulations.

A higher degree of hemoglobin absorbance indicates a slower clot formation rate.

Whole blood clotting formation was relatively slow and the hemoglobin absorbance reached the minimum level after about 420 s. A faster clot formation occurred in presence of all the dressings and 30 s after the contact between blood and formulations the % of free hemoglobin was around 20–50%. The minimum level of hemoglobin absorbance was reached after about 300 s. Chitosan capability to bind and aggregate platelets causing agglutination of erythrocytes and activation of hemostasis was predominant in all the formulation considered also in presence of GAG.

TA is a synthetic analogue of lysine and its antifibrinolytic activity is related to reversible bindings with four to five lysine receptor sites on plasminogen or plasmin: this prevents plasmin from binding to and degrading fibrin and preserves clot structure. TA showed a synergic activity with chitosan in enhancing clot formation: all the formulations reached the minimum in hemoglobin absorbance in about 180 s, presenting a better clotting performance with respect to unloaded dressings. TA CH 1 dressing allowed to maintain a % of hemoglobin at 420 s not significantly different from initial value, while CH 1 dressing determined a significant increase in % of hemoglobin comparing the initial value with respect to the final one (at 420 s). TA CH1 – HA and TA CH1 – CS dressings proved to maintain clot integrity stable from the beginning up to 420 s, while the same compositions, but unloaded, showed a partial clot modification at 90 s with a sharp increase in % of hemoglobin.

The presence of GAG slightly decreased the procoagulant activity of the dressings. In fact, HA and CS are reported in literature as anticoagulant materials. Fragments of HA having a molecular weight lower that 500 kDa, as it is the case, cause clot structure susceptible to mechanical deformations and softening (Komorowicz et al., 2017). CS is a sulfated glycosaminoglycan with heparin-like activity (Pandolfi & Hedner, 1984). On this basis, the procogulant acitivity of the systems seems exclusively due to chitosan and its synergic effect with TA.

3.2.2. In vitro and ex vivo biocompatibility and proliferation

Fig. 8 reports % biocompatibility (a) and % of proliferation (b) for all unloaded or TA loaded dressings (polymeric mixtures used to prepare dressings) towards fibroblasts (NHDF) for 3, 24 h and 7 days of contact times, respectively, (*in vitro*) and the amounts of proliferating nuclei/area (c) (positively stained for BrdU) counted for all the unloaded and TA loaded dressings (*ex vivo*).

Independently of the concentrations considered, all the dressings were characterized by good biocompatibility properties towards fibroblasts also in presence of TA and the cell viability was not significantly different with respect to growth medium (GM, standard growth conditions).

Similarly, after 24 h, the lower dressing concentrations determined the higher proliferation properties that was more evident in presence of tranexamic acid.

These results put in evidence that the presence of the hemostatic drug, TA, did not interfere with cell growth in the experimental conditions considered. Similar results were observed by Cholewinsski, Dietrich, Flanagan, Schmitz-Rode, and Jockenhoevel (2009): carotid artery derived cells maintained their morphology without toxic reaction

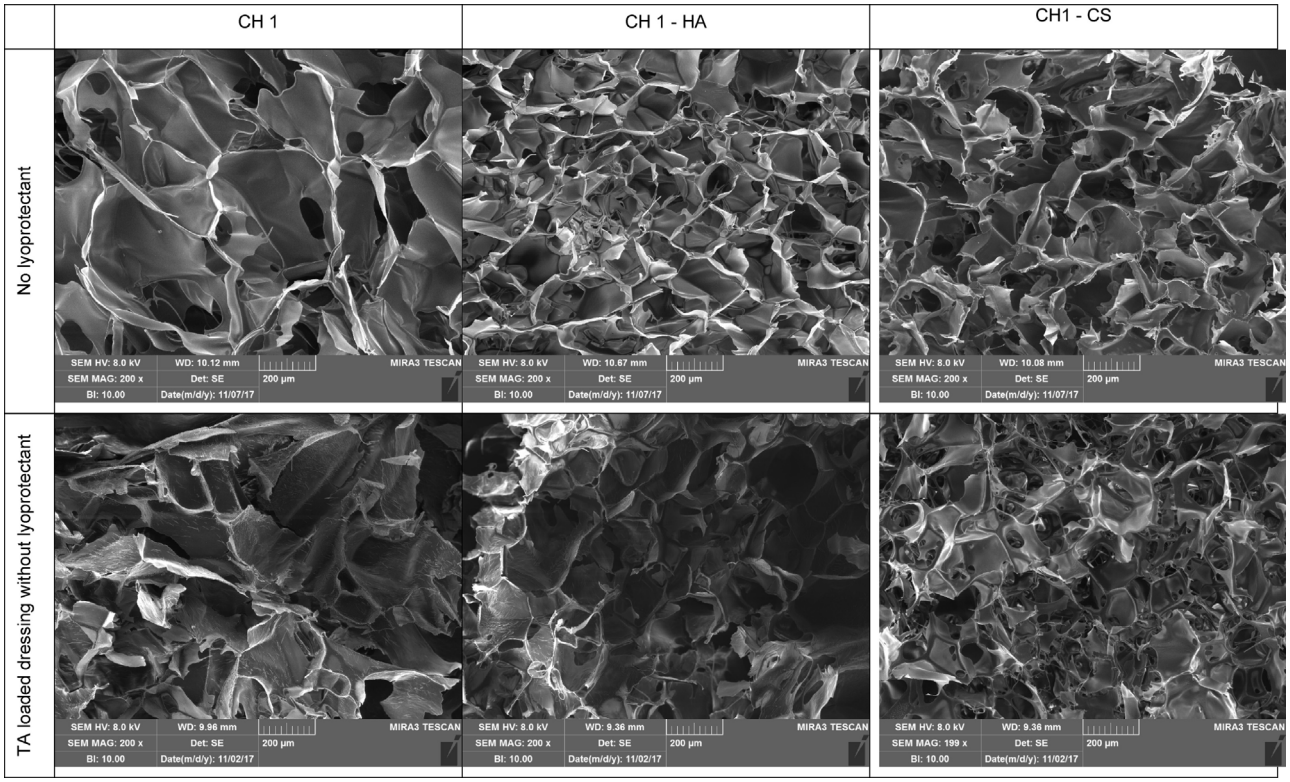

Fig. 4. SEM microphotographs of transversal sections of CH 1, CH 1 – HA and CH 1 – CS unloaded dressings and TA loaded dressings without lyoprotectans, dressings with glycerol and with mannitol.

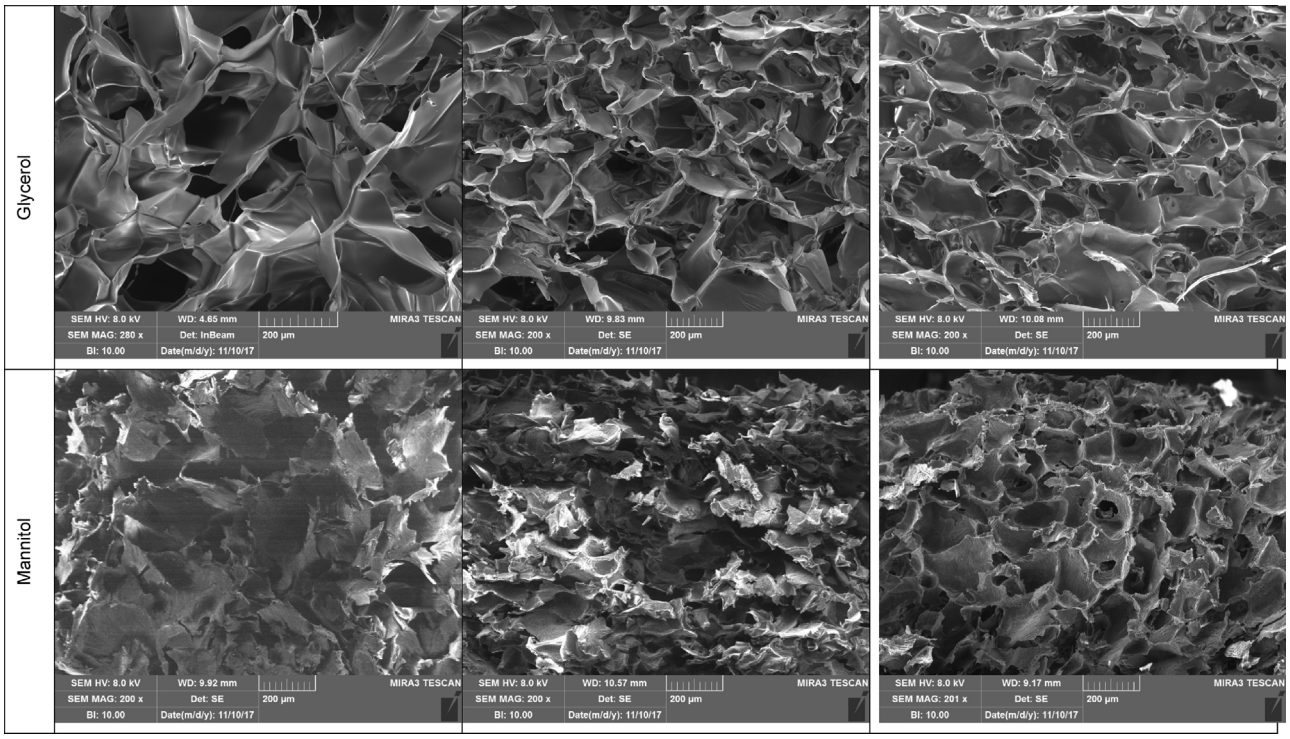

Fig. 4. (*continued*)

in presence of TA.

Ex vivo skin cell proliferation (GM) was maximum in the first 24 h and decreased after 72 h and 7 days, except for CS based system that had constant proliferation activity (Fig. 8(c)). Unloaded dressings were able to sustain cell proliferation up to 7 days demonstrating a capability to enhance cell proliferation after tissue damage. TA loaded dressings were characterized by the same proliferating properties as unloaded one, considering CH 1 and CH1 – HA dressings. As for CH1 – CS dressing the presence of TA significantly increased the system proliferation properties after 24 h and 7 days and allowed to obtain the better performance, with a prolonged effect up to one week. This probably was caused by the synergic effect of TA and CS on cell growth: CS was reported as able to enhance fibroblasts and endothelial cells proliferation (Sandri et al., 2015) while TA could stabilize fibrin clot without impairment of cell growth and proliferation (Cholewinsski et al., 2009). At this purpose fibrin clot was conceivably formed just after skin biopsies and it could retain growth factors released by platelets during normal physiological process, on this perspective TA, as fibrinolytic drug, could preserve the clot integrity assisting cell proliferation (Wolberg and Campbell, 2008).

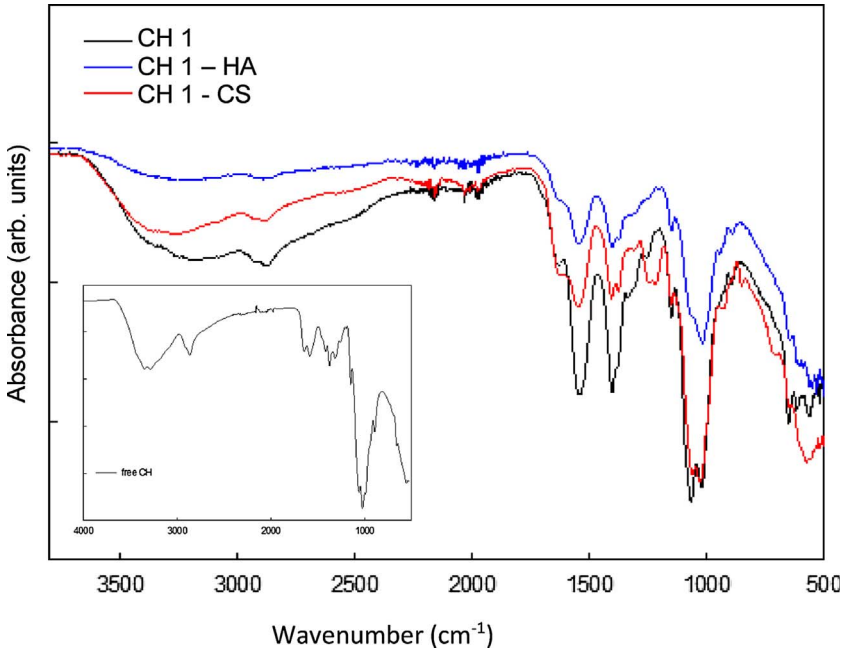

Fig. 5. FT-IR spectra of CH 1, CH 1 – HA and CH 1 – CSdressings. In the inset the FTIR spectrum of CH powder (free CH) is reported as comparison.

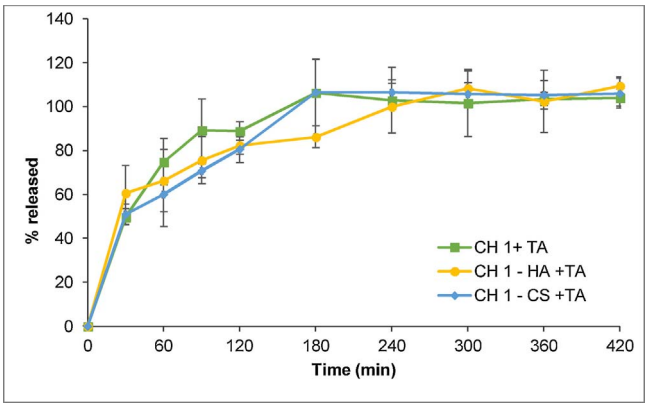

Fig. 6. TA release profiles as function of time obtained for TA loaded dressings without lyoprotectans (mean values ± sd; n = 6).

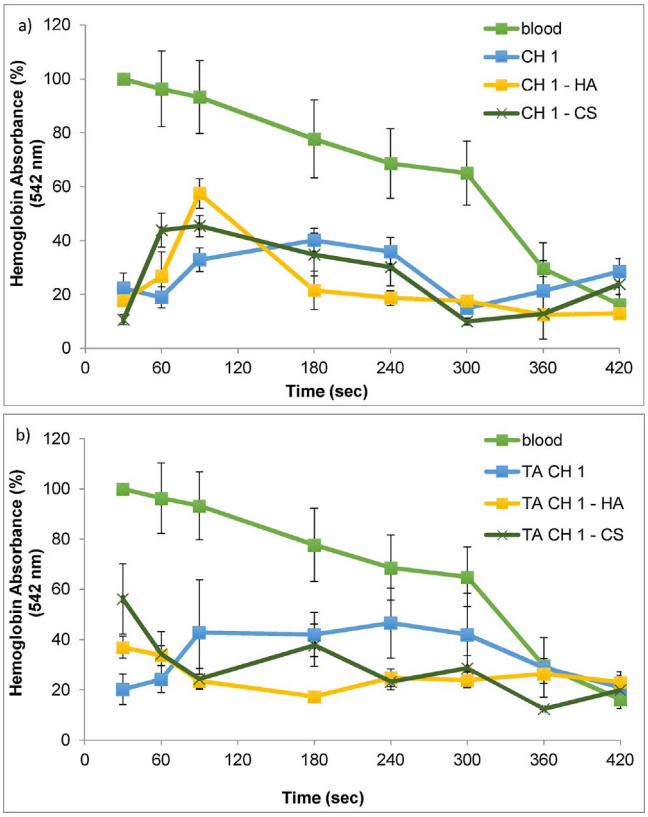

Fig. 7. % Hemoglobin absorbance profiles as function of time evaluated for (a) unloaded dressings and (b) TA loaded formulations (mean values ± sd; n = 6).

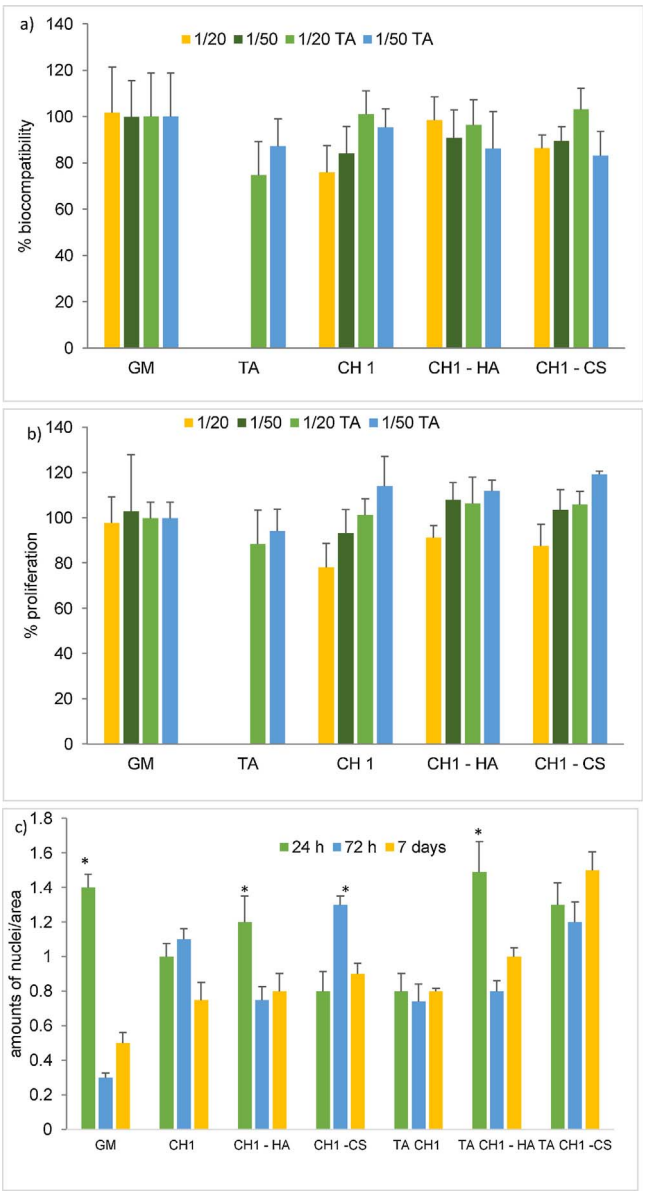

Fig. 8. % Biocompatibility (a) and % proliferation (b) of all unloaded or TA loaded dressings (polymeric mixtures used to prepare dressings) towards fibroblasts (NHDF) for 3 and 24 h of contact times, respectively (*in vitro*); amounts of proliferating nuclei (c) (positively stained for BrdU) for all the unloaded and TA loaded dressings after 24, 72 h and 7 days of contact times (*ex vivo*) (mean values ± sd; n = 8).

4. Conclusions

Sponge-like dressings based on chitosan and chondroitin sulfate or hyaluronic acid were prepared by lyophilization and loaded with tranexamic acid. The presence of GAG conceivably caused the occurrence of either chitosan and hyaluronic acid or chitosan and chondroitin sulfate interactions. The presence of glycosaminoglycans decreased hydration properties of the systems in a buffer simulating wound exudate – pH 7.2 – and this was primarily due to dressing morphology. Chitosan based dressings had a beehive structure with polyhedric cavities of about 500 μm, interconnected by pores having oval or round shapes (50–200 μm) while dressings based on chitosan in association with GAG markedly decreased the cavity dimensions (200–300 μm) with smaller pore diameters (about 50 μm). Liquid absorption is a key point in dressings to ensure hemostasis, to control wound bed hydration, enhancing grunulation phase and healing.

The presence of GAG in dressings significantly enhanced dressing bioadhesive properties considering as biological substrate egg-shell membrane to mimic wound bed. The association of chitosan with GAG allowed the consolidation of bioadhesive joint probably via hydrogen bonds that strengthen chitosan bioadhesion behavior, mainly due to charge–charge interaction between chitosan and the biological substrate. This is a functional property since bioadhesion is fundamental to favor an intimate and prolonged contact between wound dressing and lesion, avoiding formulation detachment and increasing hemostatic potential.

TA release was fast to allow procoagulant function as quickly as possible and to sustain hemostatic activity for prolonged time, to stabilize clot, moreover, TA presented a synergic effect with chitosan to speed up clotting formation.

In vitro and *ex vivo* evaluations on fibroblasts and human skin, respectively, evidenced that the developed dressings enhanced cell proliferation.

Bartley (2013) suggested that the association of chitosan with TA should improve hemostatic control and should lead to improved clinical outcomes in internal surgical wounds. In this paper, sponge-like dressings based on chitosan and GAG and loaded with TA demonstrated to combine both hemostasis and proliferation properties and it seemed promising in control bleeding and healing in wounds as well as in abdominal surgery.

Acknowledgments

Dr. F Saporito wishes to thank ABOCA Società Agricola S.p.A. for her PhD grant. The authors wish to thank Prof. C. Boselli (Department of Drug Science, University of Pavia, Italy) for the rat blood collection, Prof. G. Nicoletti (Plastic and Reconstructive Surgery Unit, Department of Clinical Surgical Diagnostic and Pediatric Sciences, University of Pavia, Salvatore Maugeri Research and Care Institute, Pavia, Italy) for the human skin biopsy and Dr. I. Tredici (Centro Arvedi, University of Pavia, Italy) for SEM measurements.

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