



Tribological characterization of polyvinyl alcohol hydrogel as substitute of articular cartilage

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ABSTRACT

Polyvinyl alcohol (PVA) hydrogels are considered as promising biomaterials to substitute damaged articular cartilage due to their excellent biocompatibility, high permeability to fluids and low friction coefficient against smooth and wet surfaces. The study aims to demonstrate that under tribological loading mimicking *in vivo* conditions, PVA hydrogel/natural articular cartilage consists in a particular combination of mating surfaces with friction and wear characteristics compatible with the application as articular joints. Hydrogel membranes having approximately 1 mm thickness were processed from PVA aqueous solutions (Aldrich Mw 89000–98000 g/mol, 99% hydrolyzed), with 10% (w/w) concentration of polymer in solution, to be used as tribological samples against bovine articular cartilage. In order to ensure the natural properties and resistance of the tissue, articular cartilage samples were collected from condyles and prepared with ~4 mm subchondral bone. Tribological tests were performed on a pin-on-plate tribometer with a linear reciprocating geometry in the presence of distilled water and phosphate buffered saline solution (PBS) at 37 °C. The reciprocating sliding frequency and stroke length were kept constant at 1 Hz and 8 mm, respectively. Contact pressures ranging from 1 MPa to 5 MPa were applied. For each experiment the friction coefficient was determined and the dominant wear mechanisms analyzed by SEM. The topography of hydrogel membranes in the hydrated state was evaluated with AFM microscopy for quantitative and qualitative analysis of worn surfaces. Results show that PVA hydrogel present excellent tribological performance against natural articular cartilage with very low friction coefficient values (≈ 0.02 to 0.05) combined with the preservation of both mating surfaces.

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1. Introduction

Despite its low vascularity and low mitotic activity of chondrocytes, articular cartilage is considered a highly efficient rubbing surface due to its low wear rate and friction coefficient [1–3]. Articular cartilage injuries and diseases affect a considerable percentage of population and are directly related to symptoms like severe pain and articular dysfunction. These pathologies treatments imply some undesirable drawbacks such as high costs, invasive surgeries, risk of disease transmission, wear and many

others [1,4,5]. A growing emphasis on synthetic polymers has led to the study of hydrogels as materials with high potential for regeneration or replacement of damaged articular cartilage. Hydrogels are characterized for being a cross-linked network of polymers and hydrophilic materials with high swelling capacity and great compliance, similar to articular cartilage which main component is water (between 70% and 80% of its total weight) [1,6–8]. The biphasic nature of cartilage is composed by a solid matrix and an interstitial fluid phase responsible for maintaining the natural joint lubrication. This property can be replicated in hydrogels, pointing out the benefits of using these materials to substitute damaged cartilage [9,10].

In this study, polyvinyl alcohol (PVA) was the eligible polymer since it stands out for its viscoelastic behaviour, hydrophilicity, chemical stability and biocompatibility [2,11–13]. PVA is used in

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many medical devices approved by FDA (Food and Drug Administration—USA) such as contact lenses, membranes, drug delivery systems and orthopedic devices [11,12,14–16] urging the relevance of this polymer on biomedical applications. In line with this, many authors have been studying PVA properties and applications particularly focusing on articular cartilage tissue engineering. Formerly and according to Ratner et al. [6], Bray and Merrill constructed a synthetic articular cartilage material based on PVA to replace a synovial joint. Ng et al. [12] studied the morphological and mechanical properties of macroporous PVA scaffolds for focal cartilage defect repair. Ma et al. [17] used polyvinylpyrrolidone (PVP)/PVA blend hydrogels to evaluate the influence of factors such as sliding speed, contact load and lubrication fluid on friction properties. The mentioned authors had in view the prosthesis durability increase and wear protection. Maiolo et al. [18] proposed the use of hydroxyapatite reinforced PVA hydrogel composites as a promising biomaterial to replace articular cartilage. Samples used presented low friction coefficient values together with a high resistance to compression, compatible with natural articular cartilage properties. The compressive and shear properties of PVA hydrogels were characterized by Stammen et al. [14] using both lower (75%) and higher (80%) water content formulations of the biomaterial. The three-dimensional network of polymer chains is responsible for water retention and can be controlled during hydrogel processing. The contribution of PVA hydrogels water content and swelling capacity was also demonstrated by Bavaresco. [19] to reduce the friction coefficient. The possible application of PVA hydrogels for knee joint reconstruction was assessed by Suciu et al. [20]. Those authors studied the wear characteristics of these compliant materials as artificial cartilage replacement assessing the influence of PVA layer thickness on wear. Other authors have been investigating PVA hydrogels for replacement of damaged cartilage focusing on the tribological behaviour of the polymer [21–23]. Notwithstanding, a better perception of the tribological response of PVA hydrogels in conditions mimicking the *in vivo* condition is still central to highlight the similarity of this material to articular cartilage.

The aim of this study was to investigate the potential of PVA hydrogel as artificial articular cartilage for articular joint replacement. To reach this main goal, the tribological behaviour of PVA hydrogel membranes was evaluated for lubricated sliding conditions against bovine articular cartilage using an oscillating pin-on-plate tribometer.

The friction coefficient evolution was analyzed throughout the test duration. The surface degradation by wear was characterized by Scanning Electron Microscopy (SEM) and the changes on surface topography were assessed by Atomic Force Microscopy (AFM).

2. Materials and methods

2.1. Synthesis of PVA hydrogel membranes

PVA hydrogel membranes were prepared from PVA aqueous solutions (Aldrich Mw 89000–98000 g/mol, 99% hydrolyzed, 10% (w/w)). Distilled water at 80 °C was added to the polymer and the final solution poured into 30 cm diameter Petri plates to dry. An acetalization was pursued for chemical treatment of all membranes by immersion in a solution of formaldehyde, sulfuric acid and anhydrous sodium (60:50:300 g). Afterwards, PVA membranes were submitted to ionizing radiation of electrons beam (dose of 25 kGy) using a Dynamiton electron beam accelerator (Institute of Energy and Nuclear Research, São Paulo, Brazil) for crosslinking of polymer network chains and for hydrogel

mechanical properties improvement. Final PVA membranes presented approximately 1 mm thickness.

2.2. Equilibrium water content measurements

For equilibrium water content (EWC) determination, PVA hydrogel membranes were first weighed and then immersed in distilled water at room temperature until reaching a constant weight. Then, samples were removed from the media and promptly weighed again. The difference between the weight of the hydrated hydrogel (W_{hh}) and the initial weight of dried hydrogel (W_{dh}) gives the EWC in percentage (Eq. (1)).

$$EWC (\%) = \frac{W_{hh} - W_{dh}}{W_{dh}} \times 100 \quad (1)$$

2.3. Cartilage sampling

Articular cartilage samples were harvested aseptically from knee joints of adult bovines. A square of tissue and subchondral bone were cut together and articular cartilage was clipped in order to leave only the needed surface area, as illustrated in Fig. 1. Subchondral bone plays a relevant role giving support during the articular joint movement. Keeping it is essential to reproduce the natural conditions of articular cartilage during tribological tests.

The collection was accomplished during the next 24 h right after slaughter, to keep tissues freshness. Samples were washed in distilled water for 5 min and kept at 4 °C until tribological tests execution.

2.4. Preparation of pins

Stainless steel pins were used on tribological tests to carry PVA hydrogel membranes. Pins were cut from a stainless steel 316L cylindrical substrate, each one with 8 mm diameter and 35 mm length. PVA membranes with 4 mm diameter were washed in distilled water and bonded to one surface extremity of the pin using a small drop of Loctite® 401. Coated pins were then submerged in distilled water to speed up the glue drying process and to keep hydration level of PVA hydrogel pellicles.

2.5. Tribological tests

Tribological tests were performed with the pair hydrogel/bovine articular cartilage for friction coefficient and wear evaluation. For reference, articular cartilage self-mated tests were also conducted. The trials were performed with a reciprocating pin-on-plate geometry using a Plint TE67/R tribometer (Phoenix Tribology Ltd, UK). Notwithstanding some limitations in perfectly mimicking the *in vivo* application in articular joints, as transient

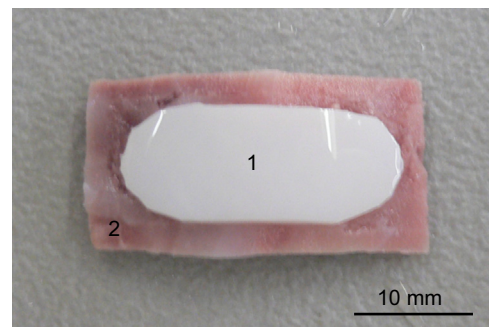


Fig. 1. Bovine articular cartilage sample with subchondral bone support. (1) articular cartilage; (2) subchondral bone.

kinematic loading and motion cycles including a cross shear component are not allowed by this test configuration, the simple pin-on-plate test geometry was adopted due to its suitability for a parametric characterization as followed in this work. Temperature was kept at $37\text{ }^{\circ}\text{C} \pm 1\text{ }^{\circ}\text{C}$, attempting to simulate the physiological condition of human body, and controlled by one thermocouple placed inside the lubricant bath. Distilled water and phosphate buffered saline (PBS) (Sigma-Aldrich) were used as lubricant fluids. Samples of articular cartilage with subchondral bone, prepared as exemplified in Fig. 1, were fixed to a stainless steel plate and the assembly was mounted inside an aluminum container where the natural tissue was permanently kept hydrated by submerging the sample in the lubricant medium, as shown in Fig. 2. Parameters such as frequency of the reciprocating motion of the plate and stroke length were fixed for all trials in 1 Hz and 8 mm, respectively. Contact pressure assumed the values of 1.0 MPa, 2.0 MPa, 3.5 MPa and 5.0 MPa. The main elements tangled in the tribological apparatus, like coated pin, plate of cartilage and lubricant fluid, and its positioning are schematically represented in Fig. 3. The duration of the reciprocating sliding in each tribological experiment was 2 h and the friction coefficient evolution was continuously monitored during the test. Worn samples were maintained hydrated after the tests in order to characterize surface topography by Atomic Force Microscopy (AFM) and thus, qualitatively and quantitatively, evaluate wear damage. Then, worn surfaces were dehydrated and gold-sputtered for examination by Scanning Electron Microscopy (SEM—FEI Nova 200) for morphological characterization and identification of wear mechanisms.

Articular cartilage was also tested against itself keeping the conditions described above, for reference friction coefficient.

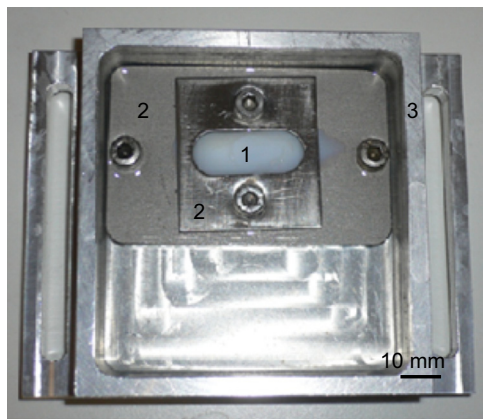


Fig. 2. Aluminum container used on tribological tests and assemble fixing the cartilage sample. (1) articular cartilage; (2) stainless steel plates fixing between them the articular cartilage sample; (3) aluminum container.

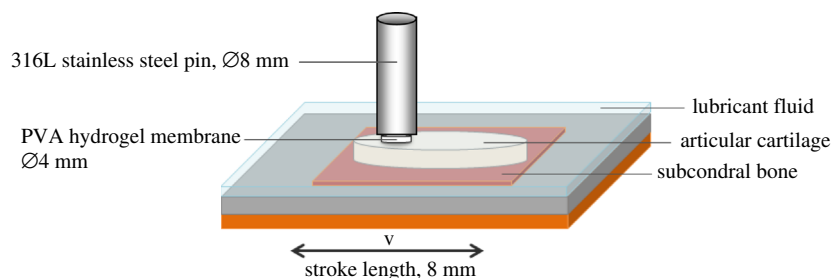


Fig. 3. Schematic representation of the mounting and positioning of the different elements during tribological tests.

2.6. Topography

Topographic characterization of PVA hydrogel at nanoscale should be accomplished in the hydrated state, as variation on the hydration condition of this kind of compliant material induces dramatic changes on the surface topography of the samples. Atomic Force Microscopy (AFM) offers the unique possibility to quantify the roughness of the surfaces at the nanometric level with high resolution, thus allowing to evaluate changes promoted by wear even if surface damage is almost negligible. In addition, this technique allows to evaluate the hydrogel surface in the hydrated state without need of preparation or dehydration. For all these reasons, AFM was the technique adopted to characterize the topography of the compliant samples considered in this study.

All the AFM examinations were carried out in the same room kept at $24\text{ }^{\circ}\text{C}$ and approximately 50% relative humidity. Observations were conducted in an aqueous environment (distilled water) using the liquid cell of the AFM (Nanoscope III, Digital Instruments, Santa Barbara, CA). AFM scans were performed in contact-mode using Si_3N_4 tips (NP-20, Digital Instruments, Santa Barbara, CA) and considering the central area of the wear track. Besides the topographic images, RMS values were obtained over an area of $10 \times 10\text{ }\mu\text{m}^2$ from roughness analysis facility of the Nanoscope III software.

3. Results and discussion

A similarity between hydrogel and articular cartilage properties is crucial to correctly mimic the natural cartilage behaviour. Hydrogels should possess appropriate mechanical and tribological properties and also provide the water content needed to support and relieve the daily stresses that are present in an articular joint. Cartilage water content is fundamentally responsible for lubrication of contact surfaces during sliding, ensuring a friction and wear protection through its capacity to exsude fluids. So, considering that a hydrogel is stable about its swelling capacity and is capable of absorbing a large amount of water [24], it seems to be a highly recommended material for damaged articular cartilage replacement, guaranteeing a very low friction coefficient and wear reduction. The released water contributes to shape a protective film between mating surfaces when compressed against each other. According to this, the water absorption capacity of PVA hydrogels was analyzed in order to determine its swelling capacity through the equilibrium water content in percentage (EWC (%)). Results showed a EWC of $32\% \pm 0.5$. This property directly influences the hydrogel behaviour during tribological tests and can be controlled during synthesis and cross-linking processes.

PVA unworn surfaces were analyzed by optical microscopy and by SEM looking for a hydrogel surface evaluation and possible surface artifacts due to the effects of drying. With the exception of some cracking in particular regions of the dehydrated samples, no

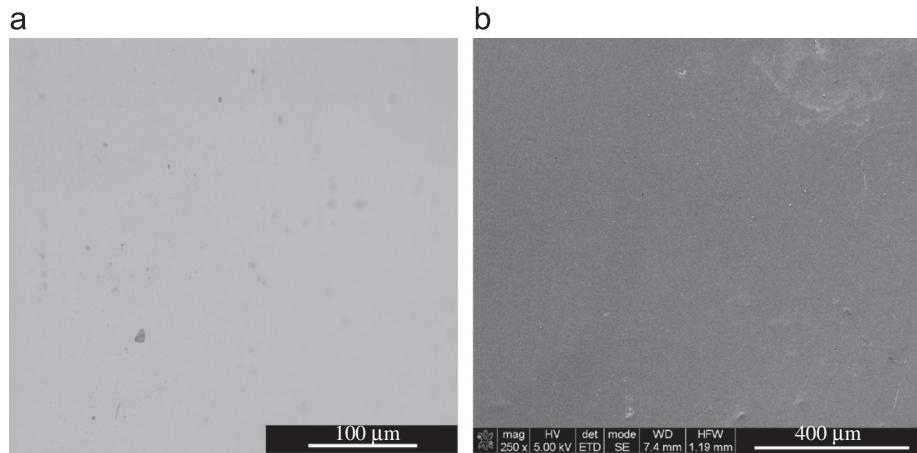


Fig. 4. Unworn surface of dehydrated PVA hydrogel: (a) optical micrograph; (b) SEM micrograph.

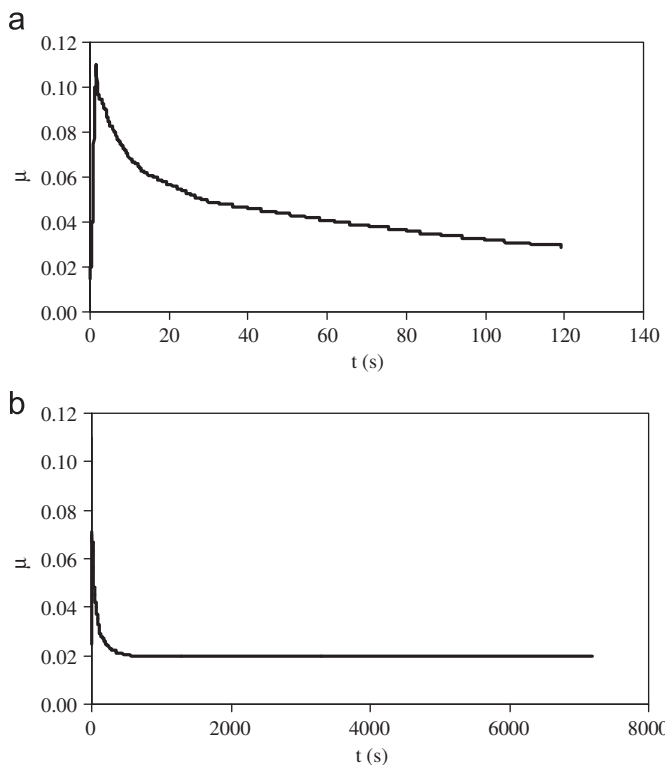


Fig. 5. Representative friction coefficient evolution curve for PVA against articular cartilage: (a) first two minutes of reciprocating sliding movement; (b) whole test duration (PBS, $p=5.0$ MPa).

significant change on surface morphology between hydrated and dehydrated samples was evidenced by optical microscopy. According to the observations of dehydrated samples by optical microscopy and SEM, PVA hydrogel revealed to be an amorphous material with a vitreous appearance and high compliance characterized by its smooth and monotonous surface, without any relevant detail (Fig. 4). This description is vital and very helpful for comparison with worn surfaces.

When tribological testes begin and sliding movement initiates, the mating surfaces are substantially in contact, partially separated by a very thin layer of lubricant fluid. This layer should not be enough to avoid an adhesion between the two compliant

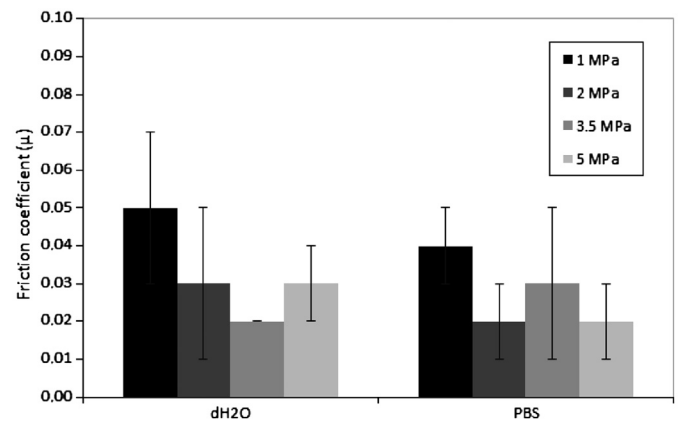


Fig. 6. Steady-state friction coefficient for the different values of contact pressure for PVA hydrogels tested against articular cartilage in distilled water (dH₂O) and PBS.

surfaces therefore contributing to the occurrence of a start-up friction with relatively high values of friction coefficient (μ), typically around 0.10. These values are justified by the rupture of asperity junctions formed between mating surfaces prior to sliding. The tendency is the friction coefficient to decrease to a steady-state regime during the immediate seconds after starting the test, characterized by very low friction values, in the range of 0.02 to 0.05. Fig. 5 shows the friction coefficient evolution curve during sliding of PVA against articular cartilage in presence of PBS for the highest contact pressure ($p=5.0$ MPa), being representative of the evolution of all frictional responses considered in this work. A detailed evolution of friction coefficient for the first two minutes of sliding is given in Fig. 5(a) showing the main characteristics of the typical running-in period, usually accomplished after ~ 500 cycles of reciprocating movement, whereas Fig. 5(b) presents the friction coefficient evolution for the whole test duration.

In order to have a reference value for friction coefficient, articular cartilage was self-tested applying a contact pressure of 1.0 MPa and using PBS as the lubricant fluid. This procedure would also be beneficial to make a comparison with values found in literature. Average values of friction coefficient ranging from 0.001 to 0.01 were achieved corroborating the values reported for other groups of study focused on tribological testing of articular cartilage [25–27].

Fig. 6 illustrates the average values of steady-state friction coefficient obtained for all applied contact pressures, for both lubricant fluids. According to Fig. 6, independently of the contact pressure and lubricant, the friction coefficient values tend to be very low never exceeding $\mu=0.07$ and being relatively close to the friction coefficient achieved for self-mated articular cartilage tested in PBS ($\mu \approx 0.01$). For both lubricant fluids there is a tendency for steady-state friction coefficient to decrease with the increase of contact pressure, excepting for 5.0 MPa in distilled water and 3.5 MPa with PBS. In these specific cases, the unexpected but slightly increase on friction coefficient could be related to an early mechanical degradation of the cartilage tissue due to differences on cartilage sample thickness or even on its positioning inside the container. When comparing both fluids, PBS shows lower friction coefficient values for contact pressure of 1.0 MPa ($\mu \approx 0.04$), 2.0 MPa and 5.0 MPa ($\mu \approx 0.02$) than distilled water. PBS viscosity (0.70 cP) is slightly higher than distilled water (0.68 cP) at 37 °C [28] and this could be responsible for a slower exudation of PBS from the hydrogel during sliding, extending the protection of mating surfaces for longer periods and then preserving from higher friction coefficient. The difference observed reinforces the advantage taken from using PBS on biotribological tests instead of distilled water.

Affording the very low steady-state friction coefficient values achieved, it was expected that PVA worn hydrogels maintain a high level of surface preservation and then a good wear response. SEM analysis allowed a morphological characterization of wear features of hydrogel contact surfaces. Fig. 7 shows the general appearance of PVA worn surfaces after sliding against articular cartilage in water (Fig. 7(a) and (b)) and in PBS (Fig. 7(c) and (d)) for limit contact pressures of 1.0 MPa (Fig. 7(a) and (c)) and

5.0 MPa (Fig. 7(b) and (d)). Micrographs of Fig. 7(a) and (c) are also representative of the worn surface appearance resulting from the reciprocating sliding under intermediate contact pressures (2.0 MPa and 3.5 MPa). From Fig. 7(a) and (c) it can be observed a great resemblance between worn surfaces resulting from sliding in both fluids for $p=1.0$ MPa, presenting smooth and homogeneous surfaces. For the maximum contact pressure (5.0 MPa) the level of superficial preservation is relatively kept for both lubricants (Fig. 7(b) and (d)), being visible some slightly marks aligned parallel to the sliding direction (horizontal in all micrographs). Besides this particularity, the PVA hydrogel contact surfaces seems to remain well preserved independently of the increase on contact pressure or lubricant used, corroborating the very low friction coefficient values ($0.02 < \mu < 0.05$) presented in Fig. 6.

Fig. 8 shows detailed views of PVA worn surfaces resulting from sliding against articular cartilage in PBS under $p=3.5$ MPa (Fig. 8(a)) and $p=5.0$ MPa (Fig. 8(b)). Fig. 8(a) is also representative of the worn surface appearance from tests under lower contact pressures ($p=1.0$ MPa and $p=2.0$ MPa) and no differences in the morphological wear features were evidenced between both lubricant media for contact pressure up to 3.5 MPa. Therefore, for this level of contact pressure ($1.0 \leq p \leq 3.5$ MPa), the preservation and the smooth appearance of PVA hydrogel contact surfaces are confirmed by SEM when observed in detailed view (magnification of $4000\times$). However, for $p=5.0$ MPa a modification on the morphological aspect of hydrogel contact surface is obvious (Fig. 8(b)). It can be observed from the detailed view a particular morphology revealing that the hydrogel starts to lose its smoothness, to evidence a rough appearance as a consequence of viscoelastic flow of the compliant surface. These features are not

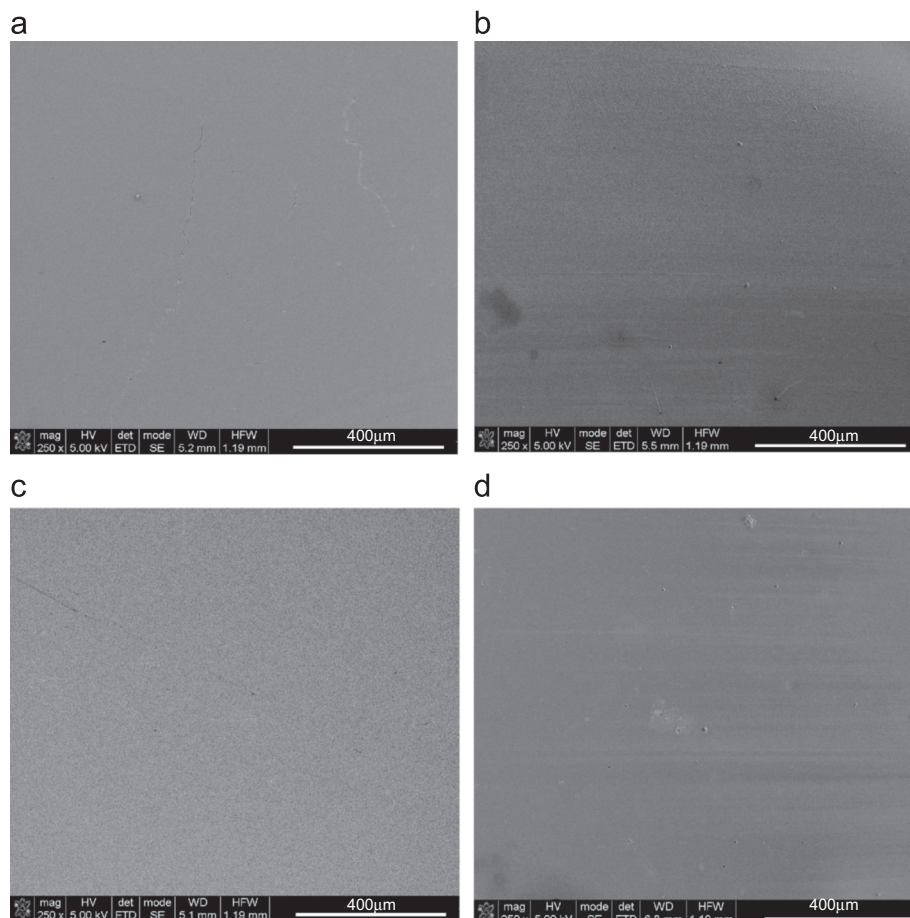


Fig. 7. General appearance of PVA worn surfaces after sliding against articular cartilage in distilled water (a, b) and in PBS (c, d) under $p=1.0$ MPa (a, c) and $p=5.0$ MPa (b, d).

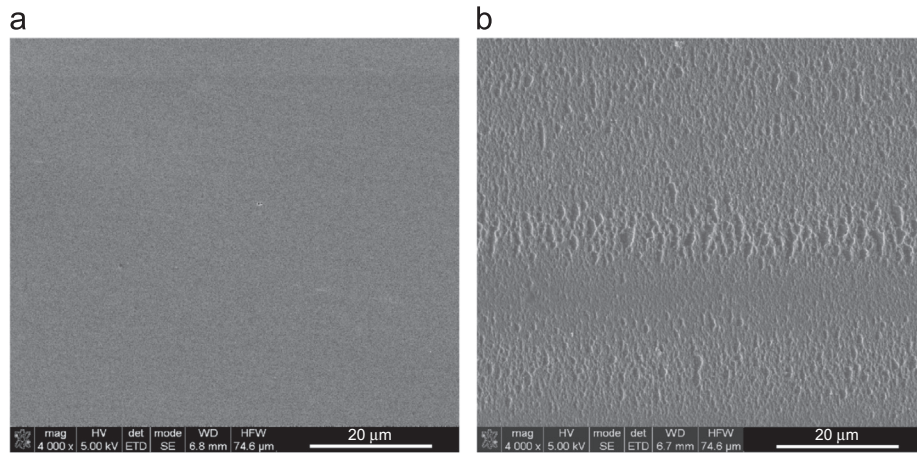


Fig. 8. Detail of PVA worn surfaces resulting from sliding against articular cartilage in PBS under $p=3.5$ MPa (a) and $p=5.0$ MPa (b).

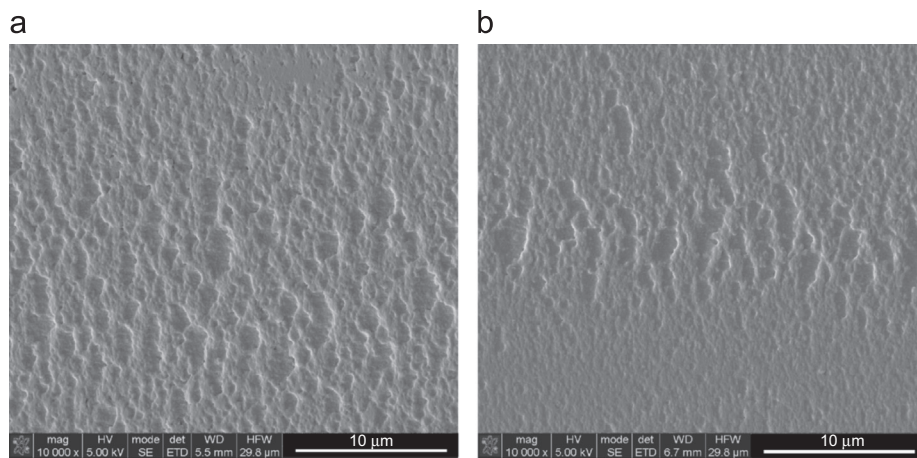


Fig. 9. SEM high magnification images of PVA worn surfaces after sliding against articular cartilage under $p=5.0$ MPa in distilled water (a) and PBS (b).

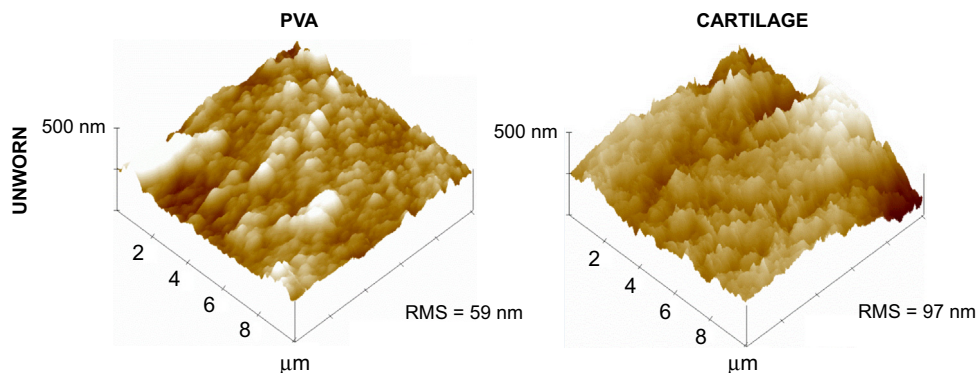


Fig. 10. Atomic force microscopy (AFM) scans ($10\ \mu\text{m} \times 10\ \mu\text{m}$) of unworn PVA and bovine articular cartilage and respective surface roughness values (RMS).

uniformly distributed by the contact surface, but as marks aligned parallel to the sliding direction. This explains the nature of the morphology evidenced by the general appearance of PVA contact surfaces presented in Fig. 7(b) and (d). In spite of the loss of smoothness of PVA surface after testing under $p=5$ MPa, this fact has no negative influence on the friction coefficient values, which were characterized by the lowest value for sliding in PBS ($\mu=0.02$), with only a slight increase to 0.03 in the presence of water (Fig. 6). This behavior is attributed to the compliant nature of the hydrogel surface that can compensate the marginal effect that the loose of

smoothness could exert in terms of the mechanical component of friction.

As previously mentioned, the contact surface of PVA tends to evidence similar wear features independently of the fluid used during sliding (distilled water or PBS). However, in order to investigate possible differences in the degree of surface damage between sliding in both media for the condition where the hydrogel surface starts to loose smoothness ($p=5.0$ MPa), high magnification SEM images are required, which are presented in Fig. 9. As it can be seen, the morphological wear features are

similar for sliding in both fluids. Though, in the case of distilled water the loss of smoothness denotes a more regular distribution by the apparent area of contact. This marginal difference on surface morphology corroborates the slightly increase of friction coefficient in the presence of distilled water for $p=5.0$ MPa (Fig. 6).

Atomic force microscopy (AFM) was the elected characterization technique to perform a detailed evaluation of the topography of all contact surfaces since it offers the possibility to quantify the surface roughness at a nanometric scale with high resolution. Besides that, this technique allows the use of hydrated samples, aspect of great relevance for PVA hydrogel pellicles which are characterized for its high compliance and water content. Thus, as previously mentioned, to perform the AFM technique samples were permanently kept hydrated with distilled water.

Root mean square (RMS or 'Rq') was the surface roughness parameter adopted in this work, which represents the standard deviation from the mean surface plane being expressed in nanometers. As in the present study it was not possible to measure the loss of material from the hydrogel tested surfaces by conventional methods, RMS values were considered as the base for the quantitative characterization of surface damage due to wear. Fig. 10 displays AFM scans of unworn reference samples, PVA hydrogel and bovine articular cartilage together with the respective RMS surface roughness measured. When comparing the RMS values of both samples, it is noticeable the difference between them: PVA hydrogel with a lower RMS value (RMS=59 nm) than articular cartilage (RMS=97 nm). The topographic features presented by the unworn PVA surface are characterized by round high-rise peaks, some of them with a height up to 350 nm, alternating with depressed valleys. This topographic characteristic can be related to replication of the molding glass surface and to non-uniformity in the spatial and temporal distribution of polymerization centers or crosslinking points during polymerization [29]. Taking in consideration that RMS=23 nm was reported for $p(\text{HEMA}+\text{MA})$ smooth surface of contact lens in hydrated state [29], it can be considered that the RMS value obtained in this work to characterize the unworn PVA surface is a consistent value. For the unworn cartilage surface, which at macroscopic scale is characterized by rough surface features denoting a non-uniform appearance, it is not surprise that, at nanometric level, this natural surface presents a rougher appearance than the hydrogel surface. As shown in Fig. 10, unworn articular cartilage shows relatively sharp peaks surrounded by deep depressions and valleys, which justify the corresponding high value of RMS.

Fig. 11 shows AFM scans of worn surfaces of PVA hydrogel after testing under different contact pressures (1.0 MPa, 3.5 MPa and 5.0 MPa) in the presence of PBS and presents the respective RMS surface roughness measured. Results presented in Fig. 11 are also representative of sliding in the presence of distilled water. According to RMS values, there is a clear decreasing of this topographic parameter with the decreasing of contact pressure, with RMS results from low to intermediate contact pressure (1.0 MPa to 3.5 MPa) being lower than the one corresponding to the unworn hydrogel surface. Corroborating these quantitative results, the topography of PVA worn surface is very smooth which is also in accordance with the morphological features obtained by SEM (Fig. 7(a) and (c)). The smoothening of the PVA hydrogel surface as a consequence of tribological loading under contact pressure up to 3.5 MPa against articular cartilage can be related to plastic deformation of this compliant material, significantly reducing the presence of peaks and valleys. Therefore, it can be inferred that for this level of contact pressure the surface of PVA hydrogel remains preserved in terms of wear and only shows a beneficial change on its topography.

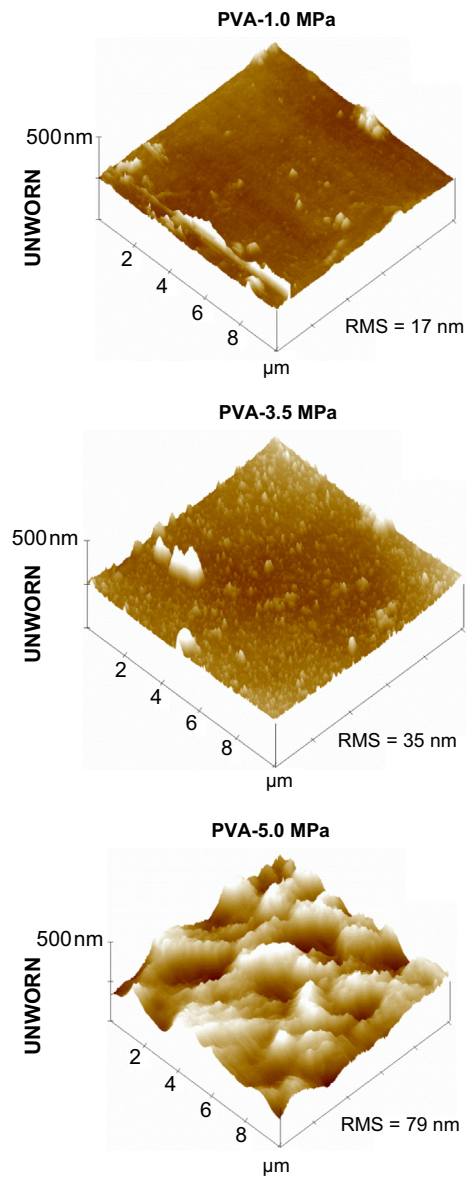


Fig. 11. Atomic force microscopy (AFM) scans ($10\ \mu\text{m} \times 10\ \mu\text{m}$) of worn PVA hydrogel after sliding against articular cartilage in the presence of PBS for different contact pressures and respective surface roughness values (RMS).

Under the maximum contact pressure, 5.0 MPa, AFM scan evidences significant change on PVA worn surface topography (Fig. 11). In this case, RMS value increases comparatively to the unworn surface ($79\ \text{nm} \gg 59\ \text{nm}$) with the contact surface presenting a rougher appearance. Again, AFM results are in accordance with the morphological features obtained by SEM (Fig. 7(b) and (d)). Therefore, as previously mentioned from SEM characterization, the topographic changes revealed by AFM scans for 5.0 MPa can be associated to the fact that for sliding under this high contact pressure condition PVA hydrogel starts loose its smoothness as consequence of viscoelastic flow. Nevertheless, there is no clear evidence that these topographic changes could result in material loss from the hydrogel surface.

For all test conditions no signs of surface damage were evidenced on the opposing articular cartilage contact surfaces. This fact, together with the reported friction and wear response of PVA under tribological loading mimicking to a certain extent *in vivo* conditions, demonstrates that this hydrogel has potential to be used as substitute of articular cartilage.

4. Conclusions

In this study the tribological properties of polyvinyl alcohol hydrogel were characterized in reciprocating sliding against articular cartilage under conditions mimicking the *in vivo* application. According to the achieved and discussed results, very low friction coefficient values were obtained (ranging from 0.02 to 0.05) with only marginal effect of contact pressure or lubricant fluid (PBS or distilled water). This remarkable frictional response together with an exceptional surface preservation of PVA hydrogel for the different test conditions, as proved by SEM and AFM analysis, demonstrates that this hydrogel has high potential to be used as substitute of articular cartilage.

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