Wear Characteristics of a Novel Bearing System for Artificial Knee Joint*

(Micro-Pocket-Covered Femoral Component and Tibial Poro-Elastic-Hydrated Cartilage)

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This paper presents an experimental investigation upon the wear characteristics of a novel bearing system proposed for the artificial knee joints. Such bearing system is original in the meaning that micro-pockets are machined on surface of the rigid femoral component; a poro-elastic-hydrated layer, called artificial cartilage, covers the rigid tibial substrate. Under compression, the fluid exudes from the artificial cartilage, fills and pressurizes the micro-pockets. In this way, a poro-elasto-hydrodynamic regime of lubrication is developed. Wear tests are performed with different types of artificial cartilages made in PVA hydrogel, articulating against smooth and micro-pocket-covered femoral components. Influence of the number of walking cycles, femoral porosity, depth of the micro-pockets, water content and polymerization degree of the PVA hydrogel upon its wear factor is illustrated. For wear tests performed until 10^6 walking cycles, in the case of PVA hydrogel with 12 300 polymerization degree and 77% water content versus smooth component, the wear factor is on order of 10^{-6} – 10^{-5} mm³/Nm; for the same artificial cartilage against a femoral component uniformly covered by identical micro-pockets of 50 μ m depth, on 15% of its surface, the wear factor drops in the range 10^{-7} – 10^{-6} mm³/Nm, i.e., the wear factor reduces with 80%. The highest wear reduction due to the micro-pockets presence is obtained in the case of PVA hydrogel with 77% water content, i.e., for a fluid amount similar to that of the natural cartilage.

Key Words: Biomechanics, Artificial Knee Joint, Wear, Micro-Pocket-Covered Component, Artificial Cartilage, Biomaterial, Polyvinyl Alcohol (PVA) Hydrogel

1. Introduction

The goals of artificial knee joint design include relief of pain, restoration of function, durability for the life of the patient and reliability⁽¹⁾. As the surgery of joint replacement has been applied to younger and more active patients, current limitations of artificial joints durability are related to the wear of their

components⁽²⁾. One of the main reasons of failure of the artificial joint (e.g., knee, hip) fixation into the host bone is the cellular reaction against wear debris, which causes the bone resorption around the implant^{(2),(3)}. Therefore, in order to improve the durability of artificial knee joint, it is necessary to diminish the tissue adverse reaction to wear debris. This target may be achieved by reduction of the amount of debris generated or/and modification of the wear particles characteristics, e.g., material, size and shape⁽⁴⁾.

The majority of wear particles are generated at the articulating surfaces because of asperity contact in a mixed or boundary lubrication regime experienced in current knee and hip joint replacements⁽⁴⁾. In principle, the increasing of the artificial knee lifetime can be achieved by enhancement of a fluid film, which

* Received 8th January, 2003 (No. 03-4005)

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prevents the articular surfaces to come into contact. The classical hydrodynamic lubrication mechanisms⁽⁵⁾ demand a continuous, high-speed relative motion between the tibial plateau and the femoral component in order to provide a substantial load carrying pressure. Unfortunately, such conditions could never met in vivo situations, where knee motions are at best intermittent in nature. Only hydrostatic bearings can successfully operate under low-speed and severe loading conditions. The pumping of lubricant generates the fluid film and a hydrostatic pocket amplifies this effect. For natural joints a self-pressurized hydrostatic mechanism, termed weeping lubrication⁽⁶⁾, was proposed: a hydrostatic pressure develops in the articular cartilage under loading, which pushes the fluid out of the cartilage into the cracks separating the contacting micro-asperities of opposing articular layers. It was experimentally proved⁽⁷⁾ that, in the case of natural cartilage, the hydrostatic pressure of the interstitial fluid carried above 90% of the total applied force for hundreds of seconds after loading. Since the force supported by the solid matrix skeleton is about 10% of the total applied load, the friction force on it is relatively reduced. Developing such mechanisms of lubrication in the case of artificial joints will allow them to operate with reduced friction and wear. In order to achieve this task, a novel bearing system for the artificial knee joint (Fig. 1) was proposed⁽⁸⁾⁻⁽¹⁰⁾. Such bearing system is original in the meaning that micro-pockets (Fig. 1) are machined on the surface of the rigid femoral component; a poro-elastic-hydrated layer, called artificial cartilage, made in polyvinyl alcohol (PVA) hydrogel, covers the rigid tibial substrate. Under compression, the fluid exudes from the artificial cartilage, fills and pressurizes the femoral micro-pockets. In this way, a poro-elasto-hydrodynamic regime of lubrication is developed. In comparison with the case of smooth femoral component (conventional design), the friction

coefficient reduces in the presence of the micro-pockets⁽⁸⁾⁻⁽¹⁰⁾. Thus, the theoretical investigations stressed that after micro-pockets pressurization, reduction of the friction coefficient became proportional to the percentage of the femoral surface covered by micro-pockets^{(9),(10)}. Assuming that lower friction would result in lower wear, the proposed bearing system seems to be able to reduce the amount of wear debris.

Purpose of the present paper is to experimentally check the theoretical predictions concerning the wear characteristics of the proposed bearing system. Wear tests are performed with different types of artificial cartilage made in PVA hydrogel, articulating against smooth and micro-pocket-covered femoral components. Previously, wear experiments for smooth femoral component against PVA hydrogels were reported⁽¹¹⁾⁻⁽¹⁴⁾. For instance, in our preliminary study⁽¹⁴⁾ wear measurements were performed for a smooth femoral component sliding versus artificial cartilage from PVA hydrogel, in order to obtain the reference wear data for the present work. Thus, investigations presented here represent the first attempt done in understanding the influence of a micro-porous structure machined on the rigid surface of the femoral component, articulating against PVA hydrogel. The influence of the number of walking cycles, femoral porosity (percentage of the femoral surface covered by micro-pockets), depth of the micro-pockets, water content and polymerization degree of the PVA hydrogel upon its wear factor is illustrated. Wear factors obtained for different micro-pocket-covered femoral components are compared with the reference data of the smooth component, and the benefit of the micro-pockets on the wear factor reduction is estimated.

2. Materials and Methods

2.1 Artificial articular cartilage

Artificial articular cartilage is made in PVA hydrogels with various water contents ϕ_f and polymerization degrees n (see Table 1). The cyclic freezing/thawing method⁽¹⁷⁾ was used in order to

Fig. 1 Novel bearing system for artificial knee joint

Table 1 PVA hydrogels employed as artificial cartilage for wear experiments

Symbolization of PVA hydrogel	Polymerization degree of PVA, n [-]	Water content, ϕ_f [%]	Preparation method
18PVAH85	1800	85	Cyclic freezing/thawing
18PVAH81	1800	81	Cyclic freezing/thawing
18PVAH77	1800	77	Cyclic freezing/thawing
18PVAH50	1800	50	Annealing
45PVAH77	4500	77	Cyclic freezing/thawing
45PVAH45	4500	45	Annealing
123PVAH77	12300	77	Cyclic freezing/thawing

prepare PVA hydrogels with high water content (77, 81 and 85%), similar with the fluid amount of the natural articular cartilage (about 78%⁽¹⁶⁾). On the other hand, PVA hydrogels with moderate water contents (45 and 50%) were manufactured employing the annealing method⁽¹³⁾. Only hydrophilic substrate⁽¹⁴⁾ from glass was used during the preparation process of the hydrogel.

2.2 Femoral component

Femoral component is considered as a circular plate of diameter $D=25$ mm and 15 mm thickness, made in stainless steel (SUS304). After micro-pockets machining, its surface was mirror finished in order to achieve the same mean roughness, $R_a=0.02$ μm , as the actual knee prosthesis⁽⁴⁾ and as the smooth component described in Ref. (14). Five different types of femoral components called *A*, *B*, *C*, *D* and *E* (Table 2 and Fig. 2) were employed. Component *A* has no micro-pockets on its surface and it is called smooth femoral component. Micro-pockets occur as uniformly and continuously distributed pores in a rectangular grid of pitch L (Fig. 2) on the surface of the components *B*, *C*, *D* and *E*.

Table 2 Femoral components used in wear experiments

Surface of the femoral component	Type	Micro-pocket			Femoral porosity ϕ_2^f [%]
		Diameter, $2R_p$ [μm]	Depth, h_2 [μm]	Pitch, L [μm]	
Smooth	<i>A</i>	-	-	-	0
Micro-pocket-covered	<i>B</i>	100	25	325	7.5
Micro-pocket-covered	<i>C</i>	100	25	230	15
Micro-pocket-covered	<i>D</i>	100	50	325	7.5
Micro-pocket-covered	<i>E</i>	100	50	230	15

(a) *B* and *D* components (b) *B* and *D* components (optical view) (SEM view)

(c) *C* and *E* components (d) SEM view of a micro-pocket (optical view)

Fig. 2 Optical microscopy view (a, c) and SEM images (b, d) of the surface of micro-pocket-covered components

The micro-pockets geometrical parameters (depth, radius, surface distribution and edge radius), have been established to reduce the local contact stresses, to assure low friction forces and to minimize the damage of the poro-elastic-hydrated cartilage⁽¹⁰⁾, made in PVA hydrogel. From a theoretical standpoint, in order to reduce the solid-solid interactions between the micro-pocket-covered component and PVA hydrogel cartilage, one might consider that the higher is the porosity ϕ_2^f , the lower is the effective friction coefficient⁽¹⁰⁾ and implicitly the wear. However, when practical applications are considered, for micro-pockets diameter larger than the grid pitch ($2R_p \geq L$), which corresponds to $\phi_2^f \geq 0.79$, the micro-pockets are no more distinctive pores and the surface resembles to a kind of “cheese-grater”, ripping the compliant surface apart and increasing the wear. A similar risk might occur if the edge radius of the micro-pockets is not carefully selected. On the other hand, a proper edge radius might induce benefic hydrodynamic lifting effects⁽¹⁰⁾.

The micro-pocket-covered components have been manufactured using the relatively accessible technology of fine mechanical drilling, which assures a maximum porosity on the component surface of about 20% ($\phi_2^f \leq 0.2$). Under this condition the distance between the closest adjoining micro-pockets is large enough to justify the assumption of negligible interaction between them⁽¹⁵⁾. During manufacturing process attention was paid to round off the edge of the micro-pockets. The polished edges radii of the micro-pockets can be seen on the SEM microscopy photo (Fig. 2(b)) as bright circles surrounding each micro-pocket.

All femoral components *B*, *C*, *D* and *E* have machined on the surface identical micro-pockets of diameter $2R_p=100$ μm . Components *B* and *D* have the same femoral porosity, i.e., $\phi_2^f=7.5\%$. Femoral porosity of the components *C* and *E* is $\phi_2^f=15\%$, i.e., two times higher than that of the components *B* and *D*. Components *E* and *D* have the same depth $h_2=50$ μm , which is two times higher than that of components *B* and *C* (see Table 2).

2.3 Lubricant

Lubricant used was deionized water instead of periprosthetic liquid.

2.4 Test rig and the anterior-posterior (A-P) displacement pattern

Wear tests were carried on a test rig, which is able to simulate the anterior-posterior displacement pattern of the knee motion during normal walking⁽¹⁸⁾. A loading screw produces a compressive load, which is applied upon the vertical axis on the femoral component via a lamella spring. Detailed description of

the test rig and its operation mode is given in detail in reference⁽¹⁴⁾.

2.5 Loading

In different human activities, the knee joint is time varying loaded^{(3),(14)}. In this study the loading was simplified and only a constant applied normal load $F_n=490$ N was considered, on the basis of the arguments given in Ref.(14).

2.6 Number of walking cycles

In this work wear experiments were performed until $N=10^5 - 10^6$ walking cycles. This number of walking cycles is not enough, since the ISO/WD 14243-000 standard⁽¹⁸⁾ recommended $5 \cdot 10^6$ cycles (about 5 years of functioning in the human body). However, tests presented here are not industrial (maker tests); these experiments are conducted to preliminarily emphasize the influence of the micro-pockets upon the wear performances of PVA hydrogel cartilage, in the frame of the proposed novel bearing system for artificial knee joints.

2.7 Wear factor evaluation

In the case of a poro-elastic-hydrated material one should differentiate the mass of the exuded water from the worn mass. Wear mass of the PVA hydrogel sample was determined by an accessible weighting method^{(14),(19)}, by employing an electronic balance with a resolution of ± 0.1 mg. In this method, first the hydrogel test sample is weighted before the wear test. Wear experiments are performed under similar conditions with those for the reference case of smooth femoral component⁽¹⁴⁾. Thus, PVA hydrogel is fixed into the water tank of the wear test rig and it is loaded at constant force. Experiments were carried-out until $N=10^5 - 10^6$ walking cycles, but after every 3 600 walking cycles (1 hour functioning), wear measurements were done. Thus, after every one hour of wear tests, the PVA hydrogel is unloaded and introduced in water for rehydration for about 24 hours. During the rehydration process, at equal intervals of one hour, the PVA hydrogel sample is removed from water is blotted with absorbent tissue and weighted. In this way, one determines the difference between the PVA hydrogel mass before and after wear experiment, at a certain moment during the rehydration process. When full rehydration was achieved, this difference rests constant, and represents exactly the worn mass m_w of the PVA hydrogel sample^{(14),(19)}. In every test configuration, corresponding to a certain type of femoral component against a specific kind of PVA hydrogel, the wear test is repeated for 3 test samples and the mean value of the PVA hydrogel wear mass is determined. Once the average value of the PVA hydrogel wear mass m_w is established, the wear factor can be calculated

as^{(13),(14)}:

$$f_w = m_w / (\rho F_n d_T), \quad (1)$$

where $\rho=1.004 \cdot 10^{-3}$ g/mm³ is the PVA hydrogel density, $F_n=490$ N is the applied normal load and $d_T=21.812$ mm is the total sliding distance⁽¹⁴⁾.

PVA hydrogel wear factors obtained for different micro-pocket-covered femoral components **B**, **C**, **D** and **E** are compared with those for the reference case (smooth component **A**) and the benefit of the micro-pockets upon the wear factor reduction is evaluated. For each micro-pocket-covered component one estimates the reduction of the wear factor $S_{R-pocket}$ relative to the smooth component:

$$S_{R-pocket} = [f_w(\phi_2^f=0) - f_w(\phi_2^f \neq 0)] / f_w(\phi_2^f=0) \cdot 100 [\%], \quad (2)$$

where $f_w(\phi_2^f \neq 0)$ is the PVA hydrogel wear factor obtained for micro-pocket-covered component and $f_w(\phi_2^f=0)$ is the wear factor for the reference case.

3. Results and Discussions

Figures 3, 4, 5 and 6 illustrate the wear factor variation f_w versus the number of walking cycles for the micro-pocket-covered femoral components **B**, **C**, **D** and **E** against the following artificial cartilages made in PVA hydrogel: 18PVAH50, 18PVAH77, 18PVAH81, 18PVAH85, 45PVAH45 and 45PVAH77. Thickness of the hydrogel layer is $h_1=2.8$ mm and the applied pressure $p_a=1$ MPa. In the case of the micro-pocket-covered components **B**, **C**, **D** and **E** (see Figs. 3 - 6) a similar tendency of the wear factor variation versus the number of walking cycles as for the smooth femoral component **A** can be observed⁽¹⁴⁾. Thus, in the initial, i.e., running-in period, the PVA hydrogel wear factor maintains almost constant; then, it decreases until a second constant level, which is kept up to surface damages are initiated; after that the wear factor sharply increases with the number of walking cycles.

In order to emphasize the influence of the femoral micro-pockets presence upon the wear factor, Fig. 7 presents the variation of the wear factor for 18PVAH77 versus the number of walking cycles, considering micro-pockets covered components **B**, **C**, **D** and **E** in comparison with the smooth component **A**. Thus, when micro-pockets are machined on the femoral component surface, which articulates against a poro-elastic-hydrated material, the wear factor reduces. Additionally, the moment when surface damages are initiated and the wear factor begins to sharply increase is delayed in comparison with the case of the smooth component **A**. These results successfully prove the benefit of micro-pockets machined on the femoral component surface, concerning the wear performances of the artificial knee joint.

Fig. 3 Variation of the wear factor versus the number of walking cycles for micro-pocket-covered femoral component **B** against PVA hydrogel ($\phi_2^f=7.5\%$; $h_2=25\text{ }\mu\text{m}$; $p_a=1\text{ MPa}$)

Fig. 6 Variation of the wear factor versus the number of walking cycles for micro-pocket-covered femoral component **E** against PVA hydrogel ($\phi_2^f=15\%$; $h_2=50\text{ }\mu\text{m}$; $p_a=1\text{ MPa}$)

Fig. 4 Variation of the wear factor versus the number of walking cycles for micro-pocket-covered femoral component **C** against PVA hydrogel ($\phi_2^f=15\%$; $h_2=25\text{ }\mu\text{m}$; $p_a=1\text{ MPa}$)

Fig. 7 Variation of the wear factor versus the number of walking cycles for **B**, **C**, **D** and **E** micro-pocket-covered components in comparison with the smooth component **A** ($\phi_2^f=0, 7.5$ and 15% ; 18PVAH77; $p_a=1\text{ MPa}$)

Fig. 5 Variation of the wear factor versus the number of walking cycles for micro-pocket-covered femoral component **D** against PVA hydrogel ($\phi_2^f=7.5\%$; $h_2=50\text{ }\mu\text{m}$; $p_a=1\text{ MPa}$)

Fig. 8 Variation of the wear factor versus the number of walking cycles for **E** micro-pocket-covered component in comparison with the smooth component **A** ($\phi_2^f=0$ and 15% ; 123PVAH77; $p_a=1\text{ MPa}$)

One concludes that the proposed novel bearing system for artificial knee joints is superior to the usual smooth surfaces.

Next, Fig. 8 illustrates the variation of the wear factor for 123PVAH77 versus the number of walking cycles, considering micro-pocket covered component **E** in comparison with the smooth component **A**. Note that the 123PVAH77 assured the lowest wear factor in tests against the smooth component **A**⁽¹⁴⁾. On the other hand, the micro-pocket covered component **E** produced the lowest wear factor when articulated against 18PVAH77 (Fig. 7). Thus, for the best available combination of PVA hydrogel and femoral component, wear tests were performed until $N=10^6$ walking cycles. While, for 123PVAH77 versus smooth component **A** the wear factor is on order of $10^{-6} - 10^{-5}$ mm³/Nm, in the case of 123PVAH77 against micro-pocket covered component **E** the wear factor drops in the range $10^{-7} - 10^{-6}$ mm³/Nm, i.e., the wear factor reduces with 80%.

Wear factor obtained for each of the micro-pocket-covered component **B**, **C**, **D** and **E** was compared with that obtained in the reference case of a smooth femoral component **A**⁽¹⁴⁾. Using Eq. (2) the wear factor reduction $S_{R-pocket}$ relative to the smooth component was evaluated. Thus, Fig. 9 illustrates the wear factor reduction $S_{R-pocket}$ for **B** and **C** components, which have the same micro-pockets depth $h_2=25\ \mu\text{m}$, but different femoral porosities. One observes that the reduction of wear factor $S_{R-pocket}$ is more accentuated when the femoral porosity ϕ_2^f increases. For instance in case of the artificial cartilage 45PVAH77 the wear factor reduction was 44.2% for $\phi_2^f=7.5\%$ (**B** component), and 66.7% when the femoral porosity increased to $\phi_2^f=15\%$ (**C** component). For all types of PVA hydrogels, with various water contents and polymerization degrees, augmentation of the femoral porosity ϕ_2^f produced the increasing of the wear factor reduction. These results are in agreement with the theoretical findings^{(9),(10)}: reducing the fraction of the solid-solid contact area, by increasing the femoral porosity, the effective friction coefficient is proportionally reduced.

Next, Fig. 10 shows the wear factor reduction $S_{R-pocket}$ for **D** and **E** components, having the same micro-pockets depth $h_2=50\ \mu\text{m}$ but different femoral porosities. The same general tendency (see for comparison Figs. 9 and 10) can be observed: augmentation of the femoral porosity ϕ_2^f produces the increasing of the wear factor reduction. For instance, in the case of artificial cartilage 45PVAH77 the wear factor reduction was 58.1% for $\phi_2^f=7.5\%$ (**D** component), and 84.1% when the femoral porosity was $\phi_2^f=15\%$ (**E** component). In case of 45PVAH45, which has the

same polymerization degree but smaller water content, the wear factor reduction increases from 30.1% to 47.2%, when the femoral porosity varies from 7.5% (**D** component) to 15% (**E** component). However, for all types of PVA hydrogels the wear factor reduction was more accentuated in the case of the **E** component, with higher femoral porosity, than for the **D** component.

From Figs. 9 and 10 one concludes that maintaining constant the micro-pockets depth h_2 and increasing the femoral porosity ϕ_2^f from 7.5% to 15%, the wear factor reduction increases with about 20–33%.

Figure 11 illustrates the wear factor reduction $S_{R-pocket}$ in presence of the femoral micro-pockets (**B**

Fig. 9 Influence of the femoral porosity on the wear factor reduction for **B** and **C** micro-pocket-covered components ($\phi_2^f=7.5$ and 15% ; $h_2=25\ \mu\text{m}$; $N=3\ 600$; $p_a=1\ \text{MPa}$; Reference case: **A** component)

Fig. 10 Influence of the femoral porosity on the wear factor reduction for **D** and **E** micro-pocket-covered components ($\phi_2^f=7.5$ and 15% ; $h_2=50\ \mu\text{m}$; $N=3\ 600$; $p_a=1\ \text{MPa}$; Reference case: **A** component)

and *C* components) for the hydrogels 18PVAH50, 18PVAH77, 18PVAH81, 18PVAH85, 45PVAH45 and 45PVAH77. Maintaining constant the polymerization degree, both for low femoral porosity ($\phi_2^f=7.5\%$ for *B* component) and high femoral porosity ($\phi_2^f=15\%$ for *C* component), the reduction of the wear factor due to the micro-pockets presence was more important for hydrogels with high water contents ($\phi_1^f=77, 81$ and 85%) than for hydrogels with moderate water contents ($\phi_1^f=45$ and 50%). In the case of the hydrogels with $n=1800$ against *C* component ($\phi_2^f=15\%$) the wear factor reduction was 36% for 18PVAH50, 67% for 18PVAH77, 61.1% for 18PVAH81 and 53.6% for 18PVAH85. One observes that the wear factor reduction $S_{R-pocket}$ first increases with the increasing of the PVA hydrogel water content and then decreases. In the case of PVA hydrogel with moderate water con-

tents, the quantity of exuded water is smaller and longer time interval is necessary for the filling and pressurization of the micro-pockets. For this reason, the wear factor reduction was relatively modest. One remarks that the PVA hydrogel with $\phi_1^f=77\%$, which has similar water content as the natural cartilage (about 78% ⁽¹⁶⁾), produces the highest reduction of the wear factor in the micro-pockets presence. Hydrogels with higher water content, i.e., $\phi_1^f=81$ and 85% , might have reduced mechanical strength and also too high permeability. For these reasons they maintain the fluid pressurization for shorter time intervals, and consequently reduction of the wear factor decreases.

Taking into account results presented in Figs. 9–11 one remarks that for two hydrogels, 18PVAH77 and 45PVAH77, which have the same water content $\phi_1^f=77\%$, but different polymerization degree $n=1800$ and 4500 , the wear factor reduction $S_{R-pocket}$ is almost the same. For instance, the wear factor reduction in the case of *B* component was 43.5% for 45PVAH77 and 42.7% for 18PVAH77 (Figs. 9 and 11). In general, for all components *B*, *C*, *D* and *E*, the wear factor reduction due to the micro-pockets is not sensitive to the polymerization degree of the PVA hydrogel.

Figure 12 illustrates the wear factor reduction $S_{R-pocket}$ for *B* and *D* components, which have the same femoral porosity $\phi_2^f=7.5\%$ but different micro-pockets depth $h_2=25$ and $50\mu\text{m}$. One observes that the reduction of wear factor $S_{R-pocket}$ is higher when the micro-pockets depth h_2 increases. For instance, in the case of the artificial cartilage 45PVAH77 the wear factor reduction was 43.5% for $h_2=25\mu\text{m}$ (*B* component), and 58.1% for $h_2=50\mu\text{m}$ (*D* component). For all types of PVA hydrogels the wear factor reduction

Fig. 11 Influence of the PVA hydrogel water content on the wear factor reduction for *B* and *C* micro-pocket-covered components ($\phi_2^f=7.5$ and 15% ; $h_2=25\mu\text{m}$; $N=3600$; $p_a=1\text{MPa}$; Reference case: *A* component)

Fig. 12 Influence of the micro-pocket-depth h_2 on the wear factor reduction $S_{R-pocket}$ for *B* and *D* micro-pocket-covered components ($\phi_2^f=7.5\%$; $h_2=25$ and $50\mu\text{m}$; $N=3600$; $p_a=1\text{MPa}$; Reference case: *A* component)

Fig. 13 Influence of the micro-pocket-depth h_2 on the wear factor reduction $S_{R-pocket}$ for *C* and *E* micro-pocket-covered components ($\phi_2^f=15\%$; $h_2=25$ and $50\mu\text{m}$; $N=3600$; $p_a=1\text{MPa}$; Reference case: *A* component)

was higher in the case of the **D** component (deeper micro-pockets), than that obtained for **B** component. The same effect, i.e., higher wear factor reduction for larger depth of the micro-pockets is also illustrated in Fig. 13, which presents the wear factor reduction $S_{R-pocket}$ for **C** and **E** components; they have the same femoral porosity $\phi_2^f=15\%$ but different micro-pockets depth $h_2=25$ and $50\ \mu\text{m}$. For instance, in the case of artificial cartilage made in 45PVAH77, the wear factor reduction was 66.7% for $h_2=25\ \mu\text{m}$ (**C** component) and 84.1% for $h_2=50\ \mu\text{m}$ (**E** component).

From Figs. 12 and 13 one concludes that maintaining constant the femoral porosity and increasing the micro-pockets depth h_2 from $25\ \mu\text{m}$ to $50\ \mu\text{m}$, the wear factor reduction increases with about 11–17%. Enhancement of the hydrostatic effects, and as result, decreasing of the wear factor, in the case of deeper micro-pockets might be considered responsible for this behavior.

4. Conclusions

This paper presents an experimental investigation upon the wear characteristics of a novel bearing system proposed for artificial knee joints. Such bearing system is original in the meaning that micro-pockets are machined on the surface of the rigid femoral component; a poro-elastic-hydrated layer, called artificial cartilage, covers the rigid tibial substrate.

Experimental results prove that wear of the poro-elastic-hydrated cartilage reduces in the case of micro-pocket-covered femoral component, in comparison with the reference case of a smooth component. Wear tests performed here sustain the theoretical predictions^{(9),(10)}, that the fluid, which exudes from the PVA hydrogel artificial cartilage, pressurizes the micro-pockets. Since a poro-elasto-hydrodynamic regime of lubrication is developed, the friction and wear are reduced. Thus, the proposed modified design of the artificial knee joint might increase its lifetime by reducing the amount of wear debris.

Wear factor reduces with the increasing of the femoral porosity (percentage of the femoral surface covered by micro-pockets) and the increasing of the micro-pockets depth. Wear factor is more sensitive to the femoral porosity variation than micro-pockets depth variation. These results agree with the theoretical findings^{(9),(10)}.

Wear reduction depends also on the water content of the artificial cartilage made in PVA hydrogel. For instance, in the case of a femoral component uniformly covered by identical micro-pockets of $25\ \mu\text{m}$, respectively $50\ \mu\text{m}$ depth, on 15% of its surface, the maximum wear factor reduction, 66.7%, respectively

84%, was obtained against hydrogel with 77% water content. This fluid amount is similar to that of the natural cartilage.

The lowest wear factor obtained in this work for 123PVAH77 against a femoral component uniformly covered by identical micro-pockets of $50\ \mu\text{m}$ depth, on 15% of its surface, is on order of $10^{-7} - 10^{-6}\ \text{mm}^3/\text{Nm}$. Laboratory tests in the case of UHMWPE (actual artificial knee joints) showed a wear factor on order of $10^{-9} - 10^{-6}\ \text{mm}^3/\text{Nm}$ ⁽⁴⁾, even long-term clinical studies for UHMWPE indicated wear factors greater than $10^{-6}\ \text{mm}^3/\text{Nm}$ ⁽⁴⁾. However, the UHMWPE wear debris, due to the cellular reaction against them, causes the bone resorption around the implant and failure of the artificial knee joint fixation into the host bone⁽⁴⁾. Even the durability of PVA hydrogel, as artificial cartilage, shown in this paper might be still insufficient for clinical use, the results obtained are encouraging. In order to improve the durability of PVA hydrogel, one might eventually employ thicker layers of artificial cartilage, which have better wear resistance. However, not only the wear volume plays an important role in the long-term survival of the artificial joints. Size, morphology and biological response to wear particles are also important factors. Since the PVA hydrogel wear particles are well tolerated by the living tissues⁽¹⁴⁾ they probably do not produce the bone resorption around the implant.

Further investigations are necessary in order to transform this idea of a novel bearing system for artificial joints into a real product, ready for clinical use. For instance, wear tests were performed in this paper on a simplified model (circular femoral component, constant applied pressure), since the purpose was to preliminarily prove the micro-pockets efficiency upon the wear reduction. Therefore, the next step requires experimental investigations considering the real geometry of the femoral and tibial components, the actual dynamic loading pattern of the knee joint and increasing the number of walking cycles.

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