J Mater Sci: Mater Med (2010) 21:147–154
DOI 10.1007/s10856-009-3863-5

Influence of dynamic load on friction behavior of human articular cartilage, stainless steel and polyvinyl alcohol hydrogel as artificial cartilage

Feng Li · Yonglin Su · Jianping Wang · Gang Wu · Chengtao Wang

Received: 23 December 2008 / Accepted: 24 August 2009 / Published online: 16 September 2009

Abstract Many biomaterials are being developed to be used for cartilage substitution and hemiarthroplasty implants. The lubrication property is a key feature of the artificial cartilage. The frictional behavior of human articular cartilage, stainless steel and polyvinyl alcohol (PVA) hydrogel were investigated under cartilage-on-PVA hydrogel contact, cartilage-on-cartilage contact and cartilage-on-stainless steel contact using pin-on-plate method. Tests under static load, cyclic load and 1 min load change were used to evaluate friction variations in reciprocating motion. The results showed that the lubrication property of cartilage-on-PVA hydrogel contact and cartilage-on-stainless steel contact were restored in both 1 min load change and cyclic load tests. The friction coefficient of PVA hydrogel decreased from 0.178 to 0.076 in 60 min, which was almost one-third of the value under static load in continuous sliding tests. In each test, the friction coefficient of cartilage-on-cartilage contact maintained far lower value than other contacts. It is indicated that a key feature of artificial cartilage is the biphasic lubrication properties.

1 Introduction

In human and animal joints, articular cartilage has excellent friction properties. The friction coefficient of articular cartilage is much smaller than that of man-made bearings [1, 2]. Two common reasons of cartilage injury are trauma and arthritis. During participation in sports, the acute and repetitive impact and torsional loading on joints may cause articular cartilage damage. Progressive joint degeneration can be induced by mechanical disruption of articular cartilage [3]. There are over 100 types of arthritis, such as osteoarthritis (OA) and rheumatoid arthritis. In the last 45 years, arthroplasty has been widely used to replace the arthritic joint with prosthesis [4]. A lot of particles wear off the gliding surface and induce osteolysis in everyday use. Then the loosening of artificial joints will occur and the revision surgery is required [5, 6]. Furthermore, lesion of hip or knee is limited to articular cartilage, but a large amount of cancellous bone is resected in the hip or knee arthroplasty. Therefore, many kinds of artificial cartilage, such as methyl methacrylate, poly (2-hydroxyethyl) methacrylate (polyHEMA) hydrogel and polyvinyl alcohol (PVA) hydrogel, have been developed to repair or replace damaged articular cartilage. PVA hydrogel is a kind of rubber-like gel which has three-dimensional network structure. Many vivo tests show that PVA hydrogel has good biocompatibility [7–9]. PVA hydrogel also has many extraordinary properties, such as low elastic modulus and high resilient [10].

Many studies have been made on the friction properties of PVA hydrogel [11, 12] and articular cartilage [13–16]. Covert et al. [11] have conducted friction tests to investigate the influence of independent variables to PVA hydrogel friction. These variables included temperature, lubrication viscosity, load and so on. Katta et al. [14] have investigated the friction properties of cartilage against cartilage at different contact stress levels under static and dynamic load conditions. They found that an increase in contact stress reduced the friction levels. In the long term reciprocating

F. Li · Y. Su · J. Wang · C. Wang (✉)
School of Mechanical Engineering, Shanghai Jiao Tong University, No.800 Dong Chuan Road, Shanghai, People's Republic of China 200240
e-mail: happywind@sjtu.edu.cn

G. Wu
College of Mechanical and Material Engineering, China Three Gorges University, Yichang Hubei 443002, China

motion friction tests of cartilage-on-metal contact, Forster and Fisher [13] found that the friction coefficients of the initial and repeat tests were both dependent on the loading time. In the research by Northwood et al. [15], the friction of biomaterials was compared in long term continuous sliding tests. According to previous studies, static load were widely used to investigate the friction properties of natural and artificial cartilage. In natural synovial joints, the loads on hips and knees under motion are dynamic applied [4]. It is necessary to study the long term friction properties of natural and artificial cartilage under dynamic load.

The purpose of this study is to investigate the frictional behavior of articular cartilage, stainless steel and PVA hydrogel as artificial cartilage under static and dynamic load in long term tests. Cartilage-on-PVA hydrogel contact and cartilage-on-cartilage contact and cartilage-on-stainless steel contact were used to evaluate friction variations in reciprocating motion pin-on-plate friction tests. Continuous sliding tests under constant load of 25 N were used. Two kinds of dynamic load were used in these tests, including cyclic load and 1 min load change.

2 Experimental details

2.1 Materials

Dimethyl sulfoxide (DMSO) and PVA were supplied by Sinopharm Chemical Reagent Co, Ltd. (SCRC), P. R. China. The viscosity-average degree of polymerization of PVA was 1750 ± 50. The saponification degree of PVA and the concentration of DMSO was 99%.

The solution of PVA was prepared in DMSO/water (80:20) mixed solvent. The concentration of PVA power in the mixed solvent was 15wt%. Subsequently the solution was stirred in autoclave at 110°C for 3 h to become viscous slurry. After degassing of the air bubbles, the viscous slurry was poured into stainless steel container and putted in a medical refrigerator for 24 h at −20°C. PVA hydrogel was obtained by thawing the viscous slurry. Then the PVA hydrogel was immersed in distilled water which was changed continuously at 20°C for 20 days to exchange the mixed solvent in the specimens [17]. Specimens were placed in Ringer' solution for 24 h before tests. The dimension of each specimen was 3 mm in thickness, 20 mm in width and 40 mm in length.

Human articular cartilage specimens used in this research were obtained from femoral condyle. Human knee joints without any injury were obtained by cadaver donation from Shanghai Changzheng hospital, P. R. China. Specimens were treated in accordance with accepted ethical practices. Cartilage pins were removed by hollow drill. Each pin sample was 9 mm in diameter and 6 mm in length. Rectangular plates, 11 mm in width and 20 mm in length, were obtained by a surgical hand saw. In the sawing process, surfaces of rectangular plates were chosen as flat as possible. The human knee joints and rectangular plates obtained were kept hydrated regularly during the sample extraction process with Ringer' solution. The subchondral bones of rectangular plates were 4 mm in thickness. Cartilage samples were stored in Ringer' solution and frozen at −20°C. The freezing processes do not affect the results of cartilage samples [13, 15, 16].

The stainless steel specimen was surgical grade 317L stainless steel, with a surface roughness of $Ra \approx 0.05$. The friction tests were run at room temperature. The tester was UMT-2 Micro-Tribometer (CETR INC., USA). Friction force is measured by strain gauge sensor and converted by software. A schematic diagram of the pin-on-plate friction apparatus is shown in Fig. 1. The cartilage pin was mounted on designed fixture which was connected to the strain gauge sensor. The cartilage and PVA hydrogel plates were mounted on the stainless steel chamber. The cartilage pin was reciprocated 2 mm s^{-1} over a distance of 4 mm in each test.

In continuous sliding tests, tests were conducted to evaluate friction variations in 60 min under constant load of 25 N (0.4 Mpa). The 1 min load change tests were conducted following the continuous sliding tests. After a continuous sliding test in 60 min, the load was systematically reduced to a constant load of 0.5 N for 1 min and the cartilage pin maintained a static state at the same time. Then a new reciprocating motion was performed in 30 min under the load of 25 N.

In cyclic load tests, a new load methodology was developed. Initially constant load of 25 N was applied to the cartilage pin for 1 s without moving. Next, the cartilage pin moved forward and then moved back. At last the load was reduced to a constant value of 0.5 N for 4 s and the cartilage pin stop moving. A schematic diagram of the cyclic load condition is shown in Fig. 2.

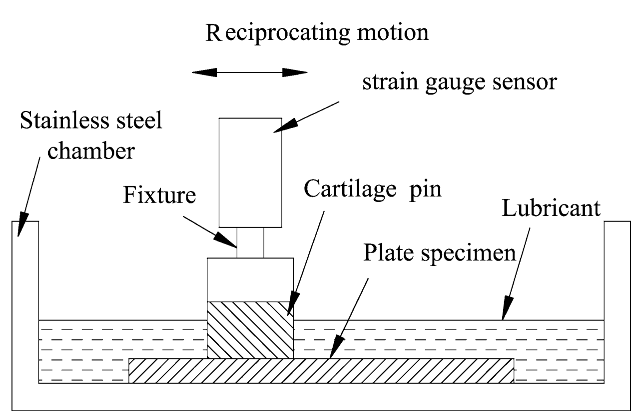

Fig. 1 Schematic diagram of the device for friction testing

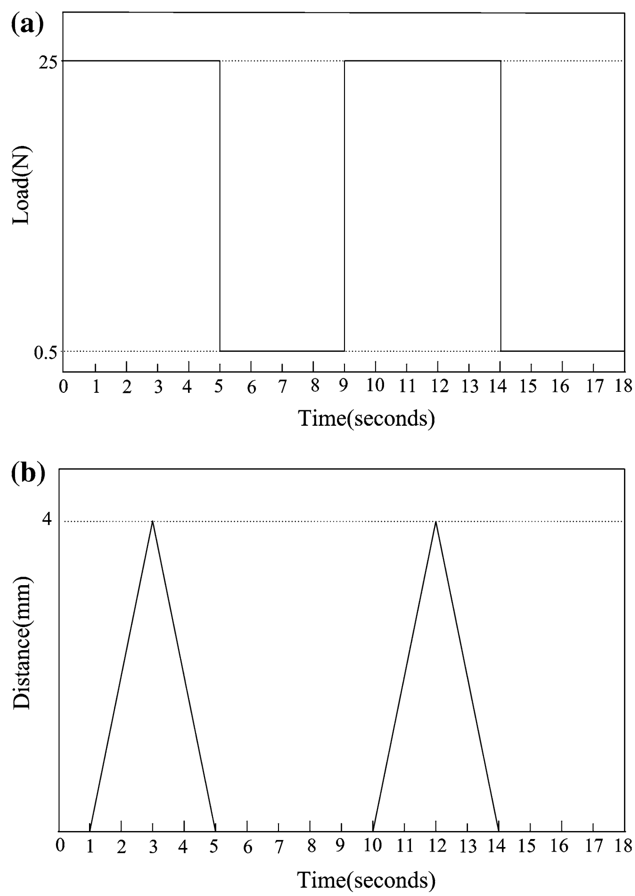

Fig. 2 Schematic diagram showing cyclic load (**a**) and sliding distance (**b**)

2.2 Measurement

The friction coefficient was the average value of each experimental configuration. The friction readings were taken every 4 min up to 60 min. Friction data were obtained from the last cycle of the data acquisition period. All tests were conducted using Ringer's solution as lubricant. A sample size of $n = 3$ were used in each experimental configuration.

2.3 Surface analysis

PVA hydrogel was used in the application of artificial cartilage. It is necessary to evaluate the surface morphology changes of cartilage-on-PVA hydrogel contact. In conventional scanning electron microscopy (SEM) a lot of preparations were required, including fixing, dehydration, drying and gold coating [18]. Articular cartilage contains 60–80% water, thus it is inconvenient to examine by SEM. Environmental scanning electron microscopy (ESEM, Philips XL30) was used for the examination in this study. No special preparation was necessary for the examination in ESEM. All samples were kept in their natural hydrated state during the examination. Surfaces of the cartilage pin and PVA hydrogel plate in cartilage-on-PVA hydrogel contact in continuous sliding tests were examined. The accelerating voltage, the spot size and the chamber pressure was 15 kV, 4.0–5.0 and ~5.0 Torr, respectively. The temperature of specimens was maintained at approximately 5–6°C.

3 Results and discussion

3.1 Continuous sliding tests in 60 min

Trends of the friction coefficient in continuous sliding tests in 60 min are shown in Fig. 3. The mean values and standard deviations of the friction coefficient were obtained from the three tests. It can be seen that the friction coefficient of cartilage-on-PVA hydrogel contact and cartilage-on-stainless steel contact gradually increased in 60 min. The friction coefficient of cartilage plates had no significant change during 60 min sliding tests. The friction rise of cartilage-on-PVA hydrogel contact and cartilage-on-stainless steel contact indicated the loss of the interstitial fluid load in the cartilage pin and PVA hydrogel [19–21]. In this study the friction coefficient of cartilage-on-stainless steel contact increased more quickly than cartilage-on-PVA hydrogel contact. In the 60 min sliding tests, the friction coefficient between the cartilage pin and PVA hydrogel plate was 0.178 compared with 0.266 of cartilage-on-stainless steel contact.

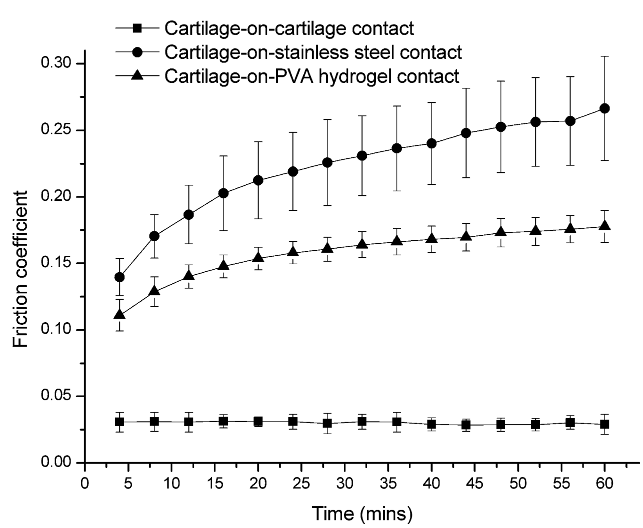

Fig. 3 Friction results of continuous sliding tests under static load

3.2 One minute load change tests

The change of the friction coefficient with the load change tests is shown in Fig. 4. After the constant load of 0.5 N in 1 min, a new continuous sliding test was performed in 30 min under constant load of 25 N. In the 30 min sliding test, there was an increase in the friction coefficient for cartilage-on-PVA hydrogel contact from 0.152 to 0.171. The cartilage-on-stainless steel contact produced an obvious increment in friction coefficient from 0.254 to 0.289. It is shown in Fig. 4 that the friction coefficient of cartilage-on-cartilage contact did not show remarkable increment in the friction coefficient.

3.3 Cyclic load tests in 60 min

General trends in the friction coefficient of cyclic load tests are depicted by the graph in Fig. 5. The friction coefficient of cartilage-on-stainless steel contact increased significantly in 60 min. The friction coefficient of cartilage-on-stainless steel contact rapidly increased in initial 20 min from 0.093 to 0.155. After 40 min in cyclic load tests, the friction coefficient of cartilage-on-stainless steel contact reached equilibrium value of approximately 0.159. The friction coefficient of cartilage-on-PVA hydrogel contact did not show a similar rise. The friction coefficient of cartilage-on-PVA hydrogel contact slightly increased from 0.046 to 0.058 in initial 20 min. Then there was no obvious increase in friction coefficient in the remaining time.

3.4 Comparison between continuous sliding tests and cyclic load tests

The friction coefficients of the two kinds of tests are shown in Fig. 6. In these tests, the friction coefficient of cartilage-on-cartilage contact had no significant change during the

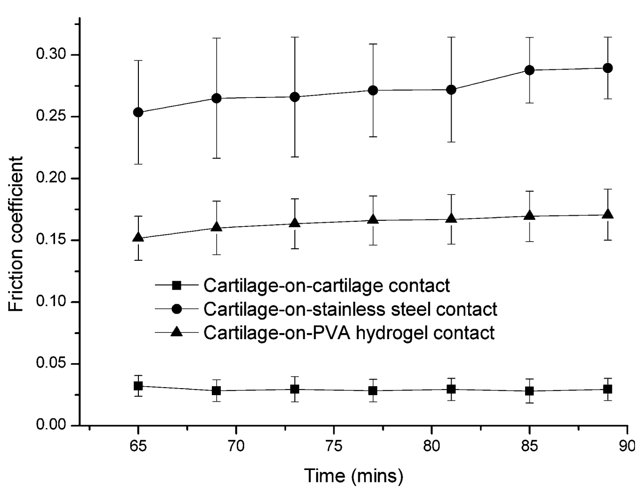

Fig. 4 Friction results of 1 min load change tests

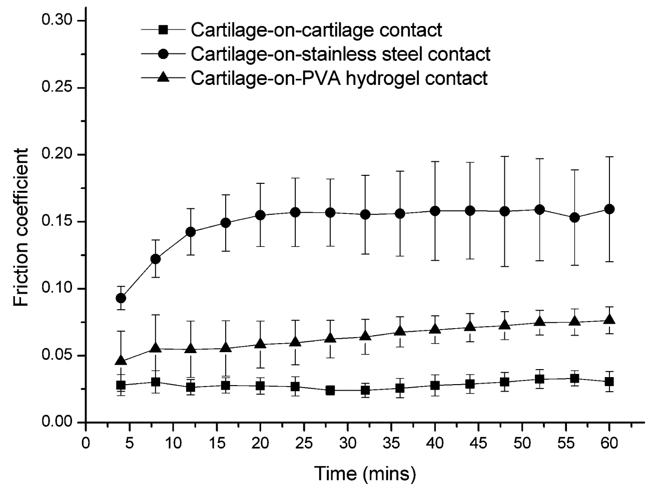

Fig. 5 Friction results of cyclic load tests

60 min of continuous sliding test, with a mean value of 0.029. The low friction coefficient was similar to the previous joint studies [22–24] and the cartilage pin studies [25, 26]. The friction level of cartilage-on-cartilage contact was far lower than that recorded for cartilage-on-stainless steel contact and cartilage-on-PVA hydrogel contact.

For stainless steel and PVA hydrogel, the cyclic load tests did result in a decrease in friction compared with the continuous sliding tests. It can be seen that the friction coefficient of cartilage-on-stainless steel contact decreased from 0.266 to 0.159 in 60 min in cyclic load tests, which was almost half of the value in continuous sliding tests. The average friction coefficient of cartilage-on-PVA hydrogel contact under cyclic load tests decreased from 0.178 to 0.076 in 60 min, which was almost one-third of the results in continuous sliding tests. In cartilage-on-PVA hydrogel contact, the increase rate of the friction coefficient under cyclic load in the initial 20 min is marked lower than that in continuous sliding tests. For cartilage-on-stainless steel contact, the increase rate of the friction coefficient had no change in initial 20 min in continuous sliding tests and cyclic load tests.

Articular cartilage is a unique porous biological material with extremely low friction. The biphasic theory of articular cartilage has been put forward to explain its excellent friction property [27]. Articular cartilage is considered to be made up of fluid phase and solid phase. Fluid phase represents the interstitial fluid in its porous structure. Solid phase represents collagen fibers, proteoglycans, and other components. Articular cartilage has very low permeability which is ranging from 10^{-15} to 10^{-16} m^4/Ns, thus very high drag forces are induced when fluid phase is moving. The friction properties of articular cartilage are based on the load partition between fluid phase and solid phase. With load the fluid phase withstands very high stressed due

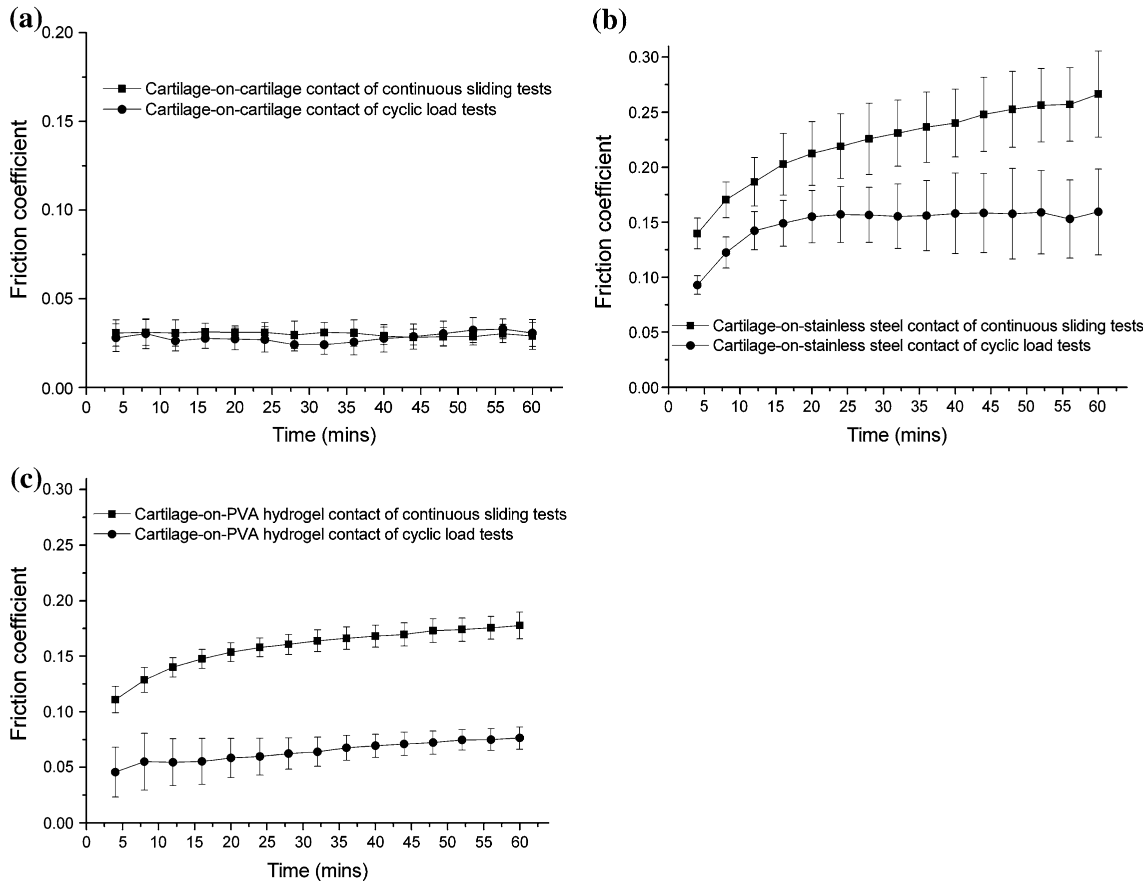

Fig. 6 Comparison of continuous sliding tests and cyclic load tests. **a** cartilage-on-cartilage contact; **b** cartilage-on-stainless steel contact; **c** cartilage-on-PVA hydrogel contact

to its high drag forces and the solid phase withstands low stress, thus low level of solid to solid contact occurs and the low friction coefficient exists. Krishnan et al. [21] have found that as the duration increases, the friction coefficient increases. The load support of the fluid phase decreases significantly at the same time. Furthermore, an amorphous and electron-dense layer called surface lamina on articular cartilage surface has been found in previous researches [28–30]. The surface lamina is a nonfibrous and membrane-like filamentous coat on articular cartilage surface. Boundary lubricants such as sulphated glycosaminoglycan (GAG), lubricin and phospholipids have been found in it. These boundary lubricants of surface lamina could maintain low friction in tests. Both the load support of fluid phase and boundary lubricants contribute to the good lubrication performance of the cartilage. In this study, the effects of boundary lubrication and biphasic lubrication for articular cartilage could not be isolated. Because all tests were performed with Ringer's solution as lubricant, there was no newly added boundary lubricant.

In this study, the friction coefficient of cartilage-on-cartilage contact maintained low value in all tests. During sliding tests, cartilage plate could keep the intrinsic biphasic properties effective due to the low permeability. The interstitial fluid phase would be rehydrated in each unload phase. As a result, the fluid phase would withstand most stress and low value of the friction coefficient was maintained. Moreover, the surface lamina contributed to maintain low friction.

Stainless steel is a single-phase material without any porous structure. In continuous sliding tests, the friction coefficient of cartilage-on-stainless steel contact increased all along in 60 min. However, the friction coefficient of cartilage-on-stainless steel contact under cyclic load reached an equilibrium value which is much lower. The continuous sliding condition showed great effect on the load partition between fluid phase and solid phase in the cartilage pin. The decrease of the friction coefficient in cyclic load tests was mostly attributed to the partial recovery of the biphasic lubrication mechanism in articular cartilage.

PVA hydrogel is a kind of viscoelastic material and has three-dimensional network structure. This kind of structure holds large amounts of water without dissolving [31, 32]. PVA hydrogel possesses a similar biphasic lubrication mechanism to natural articular cartilage, which have interstitial fluid pressure withstanding load. In continuous sliding and cyclic load tests, the interstitial fluid of the cartilage pin and PVA hydrogel continuously lost, and thus the level of solid to solid contact rise and the friction value increased. During cyclic load tests, the change of load encouraged the rehydration of the fluid phase. Because the fluid phase could maintain its load support by rehydration, the friction coefficient increased very slowly. Bell et al. [25] had investigated that parts of the cartilage plate can be unloading during continuous sliding tests and the friction coefficient remains low. In this study, the entire plate specimen was not loaded in the cyclic tests.

It is also indicated that the biphasic property of PVA hydrogel is not as effective as that of articular cartilage. The friction coefficient increased gradually with time from 0.047 to 0.076 under cyclic load due to the increment of solid phase load carriage in the PVA hydrogel. Furthermore, PVA hydrogel has no surface lamina which contains boundary lubricants. The boundary lubricant on the surface of the cartilage pin is less than that on the cartilage plate. Compared with cartilage-on-cartilage contact, cartilage-on-PVA hydrogel contact got little benefit from the boundary lubrication.

3.5 Comparison between continuous sliding tests and 1 min load change tests

The friction results of continuous sliding tests and 1 min load change tests are shown in Fig. 7. After the 1 min load change, the friction coefficient of cartilage-on-stainless steel contact and cartilage-on-PVA hydrogel contact decreased. It can be seen from Fig. 7b that the friction coefficient of cartilage-on-stainless steel contact decreased from 0.266 to 0.254. Compared with stainless steel, the friction coefficient between PVA hydrogel and cartilage has more quickly decreased from 0.178 to 0.152. It can be concluded that both cartilage-on-stainless steel contact and cartilage-on-PVA hydrogel contact could not recover to the initial friction level.

Forster and Fisher [13] have investigated entire load removal for 1 min upon the cartilage pin against stainless steel. They found that following 1 min period of load removal there is a sharp drop in the friction coefficient. In this study, the load reduced to a constant load of 0.5 N for 1 min. Furthermore, the effect of the 1 min load change for the cartilage pin against PVA hydrogel and the cartilage pin against cartilage plate were observed. The friction coefficient of cartilage-on-PVA hydrogel contact decreased more significantly than cartilage-on-stainless steel contact. The friction decrease of both stainless steel and PVA hydrogel was related to the intrinsic biphasic lubrication mechanism of articular cartilage and PVA hydrogel. However, following a 1 min period of load removal, there is no obvious change in the friction level of cartilage-on-cartilage contact.

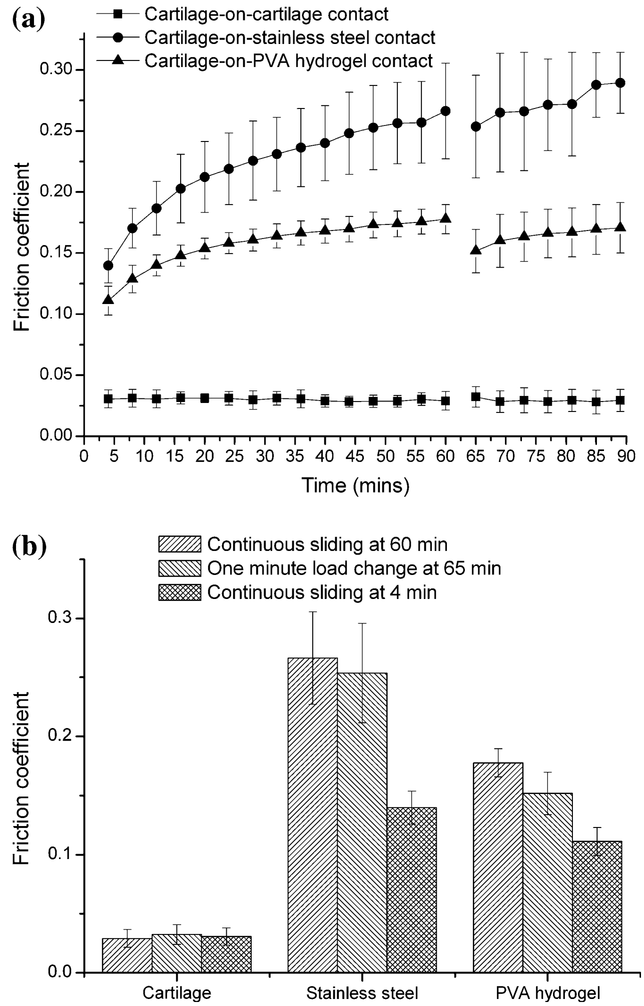

Fig. 7 Comparison of continuous sliding tests and 1 min load change tests. **a** Whole tests; **b** In 4 min

3.6 ESEM analysis of articular cartilage and PVA hydrogel

The surfaces of the cartilage pin and PVA hydrogel plate were examined by ESEM in Figs. 8, 9, 10 and 11. The ESEM examination of unworn specimen of articular cartilage revealed a smooth surface. There were no significant morphological features, such as scars or splits in the surface. Surface lamina can be seen on the top of articular cartilage. Some particles appear on the surface due to the preparation of samples. It can be seen from Fig. 9 that the surface of the unworn PVA hydrogel is smooth with some small holes. These small holes reveal the porous structure of PVA hydrogel.

After a continuous sliding test in 60 min and a 1 min load change test in 30 min, the ESEM of worn articular

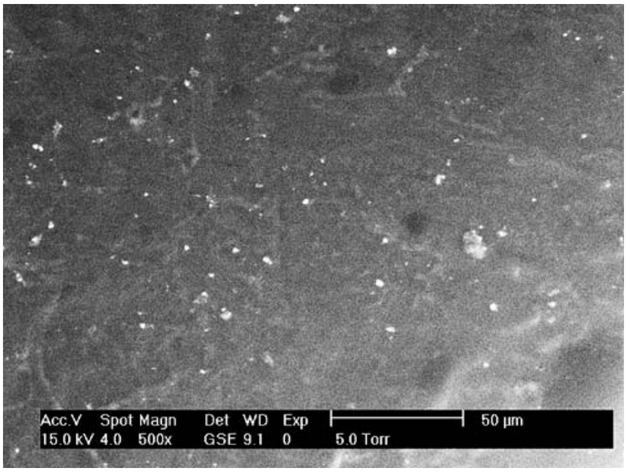

Fig. 8 ESEM image of unworn articular cartilage

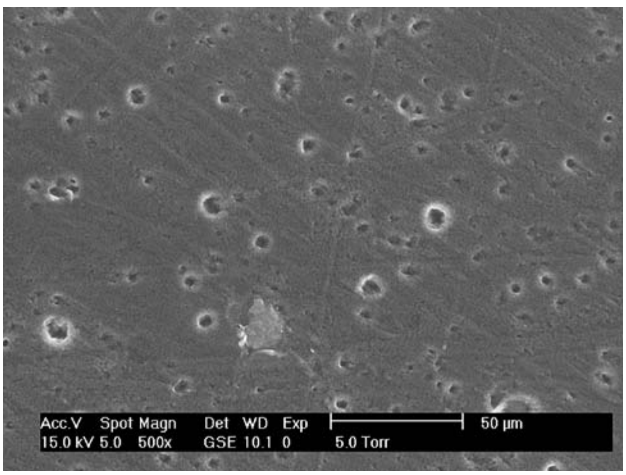

Fig. 9 ESEM image of unworn PVA hydrogel

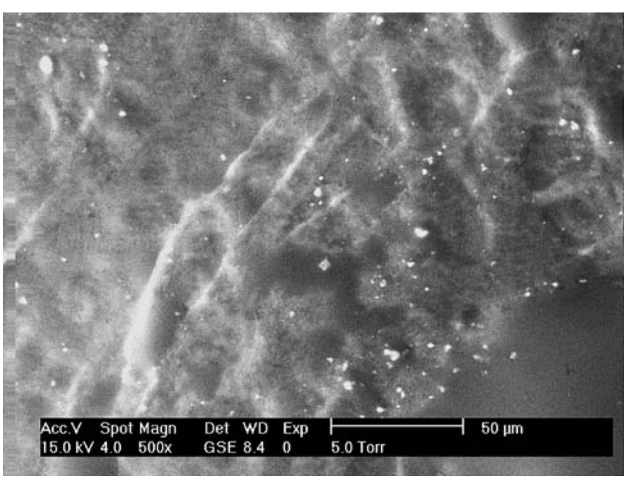

Fig. 10 ESEM image of worn articular cartilage

cartilage revealed signs of damage to the cartilage pin. Holes and scars were visible on the surface. Surface lamina worn off and some pitting appeared. It is shown that scars

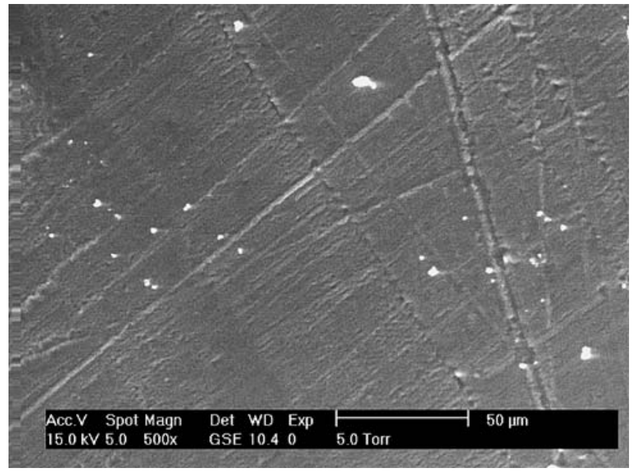

Fig. 11 ESEM image of worn PVA hydrogel

and small wear particles were found on PVA hydrogel surfaces in Fig. 11. The holes on the surface of unworn PVA hydrogel disappeared due to the high stress on the contact area.

In continuous sliding tests, the cartilage pin was loaded all the time. The interstitial fluid phase in cartilage is lost continually and the solid contact increased. Similarly, large amount of the fluid phase in PVA hydrogel was also extruded from the porous structure and most loads were supported by the solid phase in PVA hydrogel. As a result, the level of solid to solid contact rose and the surface of the cartilage and PVA hydrogel wore off.

4 Conclusions

The friction coefficient of cartilage-on-cartilage contact maintained far lower value than other contacts in all tests. In continuous sliding tests, the friction coefficient of cartilage-on-stainless steel contact and cartilage-on-PVA hydrogel contact had gradually increased. After a constant load of 0.5 N for 1 min, both cartilage-on-stainless steel contact and cartilage-on-PVA hydrogel contact have a drop in the friction coefficient. In cyclic load tests, cartilage-on-PVA hydrogel did not show similar increase rate and increased slightly. The friction coefficient of cartilage-on-stainless steel contact and cartilage-on-PVA hydrogel contact decreased significantly.

The friction behavior of articular cartilage and PVA hydrogel is associated with the intrinsic biphasic properties. It is indicated that the loss of interstitial fluid load support induced the friction rise. It is also found that the biphasic property of PVA hydrogel is not as effective as that of articular cartilage. The similarities in the friction response of cartilage-on-cartilage contact and cartilage-on-PVA hydrogel contact indicated that PVA hydrogel is a promising kind of artificial cartilage. It can be concluded that a key feature of artificial cartilage is the biphasic lubrication properties. Further studies are still necessary to optimize friction performance of the artificial cartilage.

Acknowledgments This research was supported by the national natural science foundation of China (No. 50535050). Articular cartilage donations by Shanghai Changzheng hospital were greatly appreciated. Many thanks to Prof. Zhang Dekun, Dr. Wang Shibo and Dr. Meng Baoxing of China University of Mining and Technology for their technical assistance.

References

1. Linn FC. Lubrication of animal joints. II The mechanism. J Biomechanics. 1968;1(3):193–205.
2. Wright V, Dowson D. Lubrication and cartilage. J Anat. 1976; 121(1):107–18.
3. Buckwalter, JA. Articular cartilage injuries. Clin Orthop Relat Res. 2002;402:21–37.
4. Unsworth A. Recent developments in the tribology of artificial joints. Tribol Int. 1995;28(7):485–95.
5. Benevenia J, et al. Pathologic supracondylar fracture due to osteolytic pseudotumor of knee following cementless total knee replacement. J Biomed Mater Res. 1998;43(4):473–7.
6. Engh GA, Dwyer KA, Hanes CK. Polyethylene wear of metal-backed tibial components in total and unicompartmental knee prostheses. J Bone Joint Surg B. 1992;74(1):9–17.
7. Delecrin, J, Oka M, Kumar P. Joint reactions against polymer particles: PVA-H versus UHMWPE. J Jpn Orthop Assoc. 1990;64.
8. Noguchi T, et al. Poly(vinyl alcohol) hydrogel as an artificial articular cartilage: evaluation of biocompatibility. J Appl Biomater. 1991;2(2):101–7.
9. Oka M, et al. Development of artificial articular cartilage. Proceedings of the Institution of Mechanical Engineers, Part H. J Eng Med. 2000;214(1):59–68.
10. Oka M, et al. Development of an artificial articular cartilage. Clin Mater. 1990;6(4):361–81.
11. Covert RJ, Ott RD, Ku DN. Friction characteristics of a potential articular cartilage biomaterial. Wear. 2003;255(7–12):1064–8.
12. Pan YS, Xiong DS, Ma RY. A study on the friction properties of poly(vinyl alcohol) hydrogel as articular cartilage against titanium alloy. Wear. 2007;262(7–8):1021–5.
13. Forster H, Fisher J. The influence of continuous sliding and subsequent surface wear on the friction of articular cartilage. Proceedings of the Institution of Mechanical Engineers, Part H. J Eng Med. 1999;213(4):329–45.
14. Katta J, et al. Effect of load variation on the friction properties of articular cartilage. Proceedings of the Institution of Mechanical Engineers, Part J. J Eng Tribol. 2007;221(3):175–81.
15. Northwood E, Fisher J. A multi-directional in vitro investigation into friction, damage and wear of innovative chondroplasty materials against articular cartilage. Clin Biomech. 2007;22(7):834–42.
16. Pickard JE, et al. Investigation into the effects of proteins and lipids on the frictional properties of articular cartilage. Biomaterials. 1998;19(19):1807–12.
17. Hyon SH, Cha WI, Ikada Y. Preparation of transparent poly(vinyl alcohol) hydrogel. Polym Bull. 1989;22(2):119–22.
18. Bloebaum RD, Wilson AS. The morphology of the surface of articular cartilage in adult rats. J Anat. 1980;131(2):333–46.
19. Caligaris M, Ateshian GA. Effects of sustained interstitial fluid pressurization under migrating contact area, and boundary lubrication by synovial fluid, on cartilage friction. Osteoarthritis Cartilage. 2008;16(10):1220–7.
20. Forster H, Fisher J. The influence of loading time and lubricant on the friction of articular cartilage. Proceedings of the Institution of Mechanical Engineers, Part H. J Eng Med. 1996;210(2):109–18.
21. Krishnan R, Kopacz M, Ateshian GA. Experimental verification of the role of interstitial fluid pressurization in cartilage lubrication. J Orthop Res. 2004;22(3):565–70.
22. Mabuchi K, et al. The effect of additive hyaluronic acid on animal joints with experimentally reduced lubricating ability. J Biomed Mater Res. 1994;28(8):865–70.
23. Roberts BJ, Unsworth A, Mian N. Modes of lubrication in human hip joints. Ann Rheum Dis. 1982;41(3):217–24.
24. Shirley Jones E. Joint lubrication. Lancet. 1936;227(5879):1043–5.
25. Bell CJ, Ingham E, Fisher J. Influence of hyaluronic acid on the time-dependent friction response of articular cartilage under different conditions. Proceedings of the Institution of Mechanical Engineers, Part H. J Eng Med. 2006;220(1):23–31.
26. Northwood E, Fisher J, Kowalski R. Investigation of the friction and surface degradation of innovative chondroplasty materials against articular cartilage. Proceedings of the Institution of Mechanical Engineers, Part H. J Eng Med. 2007;221(3):263–79.
27. Mow VC, Lai WM. Recent developments in synovial joint biomechanics. SIAM Rev. 1980;22(3):275–317.
28. Crockett R, et al. Imaging of the surface of human and bovine articular cartilage with ESEM and AFM. Tribol Lett. 2005;19(4): 311–7.
29. Kobayashi S, Yonekubo S, Kurogouchi Y. Cryoscanning electron microscopic study of the surface amorphous layer of articular cartilage. J Anat. 1995;187(2):429–44.
30. Orford CR, Gardner DL. Ultrastructural histochemistry of the surface lamina of normal articular cartilage. Histochem J. 1985; 17(2):223–33.
31. Anseth KS, Bowman CN, Brannon-Peppas L. Mechanical properties of hydrogels and their experimental determination. Biomaterials. 1996;17(17):1647–57.
32. Stammen JA, et al. Mechanical properties of a novel PVA hydrogel in shear and unconfined compression. Biomaterials. 2001;22 (8):799–806.