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A Synthetic Hydrogel Composite with a Strength and Wear Resistance Greater than Cartilage

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Key hurdles for replacing damaged cartilage with an equivalent synthetic construct are the development of a hydrogel with a strength that exceeds that of cartilage and fixation of this hydrogel onto the surface of an articulating joint. This article describes the first hydrogel with a tensile and compressive strength (51 and 98 MPa) that exceeds those of cartilage (40 and 59 MPa), and the first attachment of a hydrogel to a metal backing with a shear strength (2.0 MPa) that exceeds that of cartilage on bone (1.2 MPa). The hydrogel strength is achieved through reinforcement of crystallized polyvinyl alcohol with bacterial cellulose. The high attachment strength is achieved by securing freeze-dried bacterial cellulose to a metal backing with an adhesive and a shape memory alloy clamp prior to infiltration and crystallization of the polyvinyl alcohol. The bacterial cellulose-reinforced polyvinyl alcohol is three times more wear resistant than cartilage over one million cycles and exhibits the same coefficient of friction. These advances in hydrogel strength and attachment enable the creation of a hydrogel-based implant for durable resurfacing of damaged articulating joints.

1. Introduction

Osteoarthritis (OA) is a degenerative joint disease that is a common cause of disability.^[1,2] OA leads to the formation of lesions in the articular cartilage that lines the ends of bones. Articular cartilage lesions most commonly occur in the knee and cause debilitating pain. Cartilage has a limited capacity

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for healing because it is avascular.^[3] One approach to treating cartilage lesions is to attempt to regrow cartilage with a technique such as microfracture or autologous chondrocyte implantation.[4-6] Unfortunately, these methods have high failure rates (25-50% at 10 years), prolonged rehabilitation times (>12 months), and show decreasing efficacy in patients older than 40-50 years.^[6,7] Implantation of fresh osteochondral allografts can speed recovery as grafts eliminate the need to regrow cartilage and, with a survivorship of 82% at 10 years, this approach is the most successful strategy for treatment of cartilage defects in the knee.^[8,9] Unfortunately, the small supply of fresh allografts limits the number of these procedures to $\approx 1\%$ of all cartilage repair surgeries.^[6] Failure of these treatment strategies usually leads to more invasive total knee replacement. While total knee replacement is successful in older patients, it is not suitable for

younger patients for whom the implant is likely to fail within their lifetime, thus requiring a second invasive surgery. For example, risk of revision surgery for total knee replacement is 5% for men in their 70's, but rises to 35% for men in their early 50's.^[10] Thus there is a clear need for minimally invasive treatment options that treat cartilage lesions with a low failure rate, enable rapid recovery, and are widely available.

Given the need for a less invasive alternative to total knee replacement for treatment of OA, there are ongoing efforts to replace damaged cartilage with a device made of traditional orthopedic materials, such as a cobalt-chrome alloy or ultrahigh-molecular-weight polyethylene.^[11–13] However, these materials have a much higher coefficient of friction (COF) than cartilage that leads to an unacceptable level of wear on the opposing cartilage surface. In addition, these materials are much stiffer than cartilage and will therefore cause an abnormal, non-physiological stress distribution in the joint, potentially contributing to the damage of surrounding cartilage. The failure rate for such implants is $\approx 20\%$ after four years.^[14]

Hydrogels, polymer networks swollen with water, are the most promising synthetic material for replacement of cartilage because hydrogels can be made to have identical mechanical and tribological properties as articular cartilage.^[15–17] We have previously reported an approach to create the first hydrogel that is equivalent to cartilage in its tensile and compressive





Figure 1. A) Compressive versus tensile strength and B) compressive versus tensile modulus for annealed BC-PVA and annealed BC-PVA-PAMPS (this work, denoted with stars) compared with other strong hydrogels (denoted with circles, see Table S1 in the Supporting Information for data and references).

strength. This hydrogel was made by infiltrating a bacterial cellulose (BC) nanofiber network with polyvinyl alcohol (PVA) and poly(2-acrylamido-2-methyl-1-propanesulfonic acid sodium salt) (PAMPS).^[15] This hydrogel exhibited a tensile

strength of 22.6 MPa and a compression strength of 20 MPa. In comparison, the range of tensile and compression strengths reported for human cartilage are 8.1–40 and 14–59 MPa, respectively.^[18–20] Thus, there is still room to improve the

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Figure 2. A,B) Representative tensile and compressive stress-strain curves for PVA hydrogels annealed at different temperatures. C) Tensile strength and moduli of PVA annealed at different temperatures (mean \pm SD). D) Compressive stress at 0.8 strain and moduli of PVA annealed at different temperatures (mean \pm SD). E) Crystallinity and solid content weight fraction of PVA annealed at different temperatures (mean \pm SD). Numerical data for panels (C-E) is listed in Table S2 (Supporting Information).

strength of hydrogels to be at the higher end of the range of strengths reported for cartilage, or to even exceed cartilage in strength, while having a similar modulus, coefficient of friction, and resistance to wear. Achieving higher strengths would reduce the risk of failure for a weight-bearing hydrogel-based implant.

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In preparing the BC-PVA-PAMPS hydrogel, we used the freeze-thaw method to gel the PVA-water mixture after infiltration into the BC. This gelation step is necessary to increase the strength of the PVA hydrogel, and to prevent dissolution of the PVA in the following PAMPS infiltration step. The freeze-thaw method is commonly used to create PVA hydrogels with tensile strengths up to 1 MPa.^[21-24] The increase in strength upon freezing and thawing the PVA is attributed to crystallization of the PVA chains and phase segregation.^[21,22] The tensile strength of PVA hydrogel can be further increased to 18-20 MPa by drying and annealing the PVA, followed by rehydration.^[25-27] We will hereafter often refer to the process of drying, annealing, and rehydration as simply "annealing" to be concise, i.e., an "annealed" hydrogel is one that has gone through the process of drying, annealing, and rehydration. The reason for the higher strength of annealed hydrogels is that the annealing process greatly increases the crystallinity and decreases the water content of PVA relative to the freeze-thaw process. A disadvantage of the annealing process is that it can result in more bubbles and cracks in the PVA, especially as the sample thickness or water content increases.

Given the higher tensile strength of annealed PVA relative to freeze-thawed PVA, we decided to test whether changing from a freeze-thaw to annealing process can improve the mechanical strength of a BC-PVA-PAMPS hydrogel while retaining adequate control over the hydrogel shape and defect content. Given the tensile strength of a BC-PVA-PAMPS hydrogel (22.6 MPa), is already similar to the tensile strength of a PVA hydrogel made by annealing (20 MPa), it was not obvious that switching to the annealing process for a BC-reinforced hydrogel would vield further improvements in the mechanical strength. In addition, the presence of BC or PAMPS could potentially interfere with the crystallization of PVA that occurs during the annealing process, thereby hindering the improvement in mechanical strength that occurs as a result of crystallization. It was also not clear whether we could obtain high-quality, bubble-free, crack-free samples after annealing PVA reinforced with BC. Minimizing



Figure 3. A,B) Representative tensile and compressive stress-strain curves for BC-PVA hydrogels annealed at different temperatures. C) Tensile strength and moduli of BC-PVA annealed at different temperatures (mean \pm SD). D) Compressive strength and moduli of BC-PVA annealed at different temperatures. Crystallinity and solid content weight fraction (mean \pm SD) E) of BC-PVA annealed at different temperatures (mean \pm SD). Crystallinity and solid content weight fraction (mean \pm SD) E) of BC-PVA annealed at different temperatures (mean \pm SD). Crystallinity here stands for the weight fraction of crystalline PVA in the entire hydrogel sample, including the PVA and water. Numerical data for panels (C–E) is listed in Table S3 (Supporting Information).



defects is necessary to maximize the mechanical strength of the hydrogel. Finally, it was unclear whether the lower water content of the annealed hydrogel increases the COF and wear of an opposing cartilage surface.

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This work shows that reinforcement of annealed PVA with BC leads to a 3.2-fold improvement in the tensile strength and a 1.7-fold increase in the compressive strength (see Figure 1; Table S1, Supporting Information). The highly crystallized BC-PVA hydrogel that results from annealing is the first hydrogel with a tensile and compressive strength that exceeds that of cartilage. Reinforcement of the PVA with BC essentially eliminated the deformation and bubbles that would otherwise occur during annealing. When tested against cartilage for one million cycles, annealed BC-PVA wore an opposing cartilage surface to the same extent as cartilage and was three times more resistant to wear than cartilage. Annealed BC-PVA was 4.3 times more wear resistant than annealed PVA. The COF of BC-PVA against cartilage was equivalent to that of cartilage against cartilage, whereas the COF of annealed PVA increased over the course of the test to be 6.75 times greater than cartilage. In contrast to results with freeze-thawed BC-PVA, addition of PAMPS to the annealed BC-PVA decreased the tensile strength of the hydrogel due to a loss of crystallized PVA and an increase in the water content. The improved tensile strength of annealed BC-PVA enabled it to attach to a metal base with a shear strength 68% greater than the shear strength of cartilage on bone. The high strength, high wear resistance, and low COF of annealed BC-PVA make it an excellent material for replacing damaged cartilage.

2. Results

2.1. Effect of Annealing on Morphology

We first examine the effect of the hydrogel composition on the shape of the sample after drying, annealing and rehydration (or simply "annealing" to be concise). Figure S1A (Supporting Information) shows a sample of BC became wrinkled and folded at the edges after annealing. BC samples that were annealed in a 10 wt.% solution of PVA were also deformed (Figure S1B, Supporting Information). The PVA layer that forms on top of the BC after annealing contains a large number of bubbles and easily delaminates from the BC film. Figure S1C (Supporting Information) shows a sample of 40 wt.% PVA also formed a large number of bubbles and deformed during the annealing process. However, reinforcement of 40 wt.% PVA with BC allowed the hydrogel to retain its shape without deformation after annealing. We attribute this lack of deformation to the higher solid content and tensile modulus of the BC-reinforced PVA. The nanoscale network of the BC layer appears to suppress the formation of the large bubbles that are visible in the 40 wt.% PVA sample. Comparing Figure S1B,D (Supporting Information) indicates that the approach of infiltrating a high concentration of PVA into BC in a hydrothermal bomb, followed by removal of excess PVA from the BC surface, results in a more uniform hydrogel than if a BC sample is placed in a more dilute PVA

solution that is concentrated via drying. These results demonstrate that, unlike BC alone, PVA alone, or the combination of BC with a 10 wt.% of PVA, the BC infiltrated with 40 wt.% PVA could retain its shape and remain relatively free of bubbles and other defects after annealing.

2.2. Effect of Annealing on the Mechanical Properties of PVA Hydrogel

To determine the effects of annealing on the mechanical properties of various hydrogel compositions, we first analyzed the effects of annealing on a PVA hydrogel as a reference point. The PVA was fully hydrolyzed with a molecular weight of 145 000 g mol⁻¹. A 40 wt.% PVA solution was dried at 90 °C for 24 h, annealed at 90, 120 or 140 °C for 1 h, and then placed in a 0.15 M phosphate-buffered saline (PBS) solution for 24 h for rehydration. PVA samples that underwent a freeze-thaw cycle were tested for comparison. **Figure 2**A,B shows that annealing



Figure 4. A) Tensile strength, tensile moduli and solid content weight fraction of BC-PVA-PAMPS hydrogels that were made with solutions containing different concentrations of the AMPS monomer (mean \pm SD). BC-PVA samples were annealed before infiltration of AMPS. B) Compressive strength and moduli of BC-PVA-PAMPS hydrogels (mean \pm SD). Numerical data for the figure is listed in Table S4 (Supporting Information).

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the hydrogel dramatically increased the tensile and compressive strength relative to samples that had undergone a freeze-thaw final length – initial length

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cycle. Note strain is calculated as initial length

Figure 2D and Table S1 (Supporting Information) shows that, relative to the freeze-thaw process, annealing increased the tensile strength by 60 times (0.26-15.6 MPa) and the compressive strength by 9 times (14.8-140.8 MPa). Increasing the annealing temperature from 90 to 140 °C led to an increase in the tensile strength and modulus, similar to previous work.^[25,26] The increase in strength and modulus has been ascribed to the increase in the crystallinity and solid content of the hydrogel after annealing. Figure 2E confirms that the crystallinity and solid content of the annealed PVA hydrogels are much greater than that of a freeze-thawed PVA hydrogel. For example, a PVA hydrogel made via the freeze-thaw process has an overall solid content of 8.8 wt.% and a PVA crystallinity of 20.6 wt.%, whereas a PVA hydrogel made via annealing at 90 °C has an overall solid content of 42.2 wt.% and a PVA crystallinity of 58.1 wt.%.

Numerous previous studies have examined the crystallization mechanism of PVA, and interested readers can refer to that work for a more in-depth analysis of PVA crystallization.^[28] The crystallites formed during annealing strengthen the otherwise amorphous PVA by acting as tough cross-links that redistribute applied stresses and hinder crack propagation.^[25,29] The crystallites also increase the solid content and strength of the hydrogel by reducing the amount of water taken up by the PVA when it is soaked in PBS (0.15 M) after annealing. When PVA is crystallized, it is bonded to itself and thus cannot bond to or be swollen with water.^[25,28]

2.3. Effect of Annealing on a BC-PVA Hydrogel

Next, we applied the same annealing process to BC-PVA hydrogels. As with the PVA hydrogels, the BC-PVA hydrogels were dried at 90 °C for 24 h, annealed at 90, 120 or 140 °C for 1 h and then placed in a PBS (0.15 M) solution for 24 h for rehydration. Figure 3A,C shows the tensile strength of the annealed BC-PVA hydrogels reached 50.4 MPa, an increase of 4.6 times relative to the BC-PVA that went through a freeze-thaw cycle, and an increase of 3.2 times relative to annealed PVA that was not reinforced with BC. Figure 3B,D shows the compressive strength increased from 55.3 to 95.4 MPa after annealing. Note the compression strength was measured for a BC-PVA hydrogel attached to a metal pin (see Supporting Information). Similar to the PVA hydrogel, this dramatic increase in strength can be attributed to the increase in crystallinity and solid content after annealing. Figure 3E shows the crystallinity of the BC-PVA hydrogel increased from 7.2 wt.% after a freeze-thaw cycle to 39.5 wt.% after annealing. The solid weight fraction of the BC-PVA hydrogel increased from 11.2 wt.% after a freeze-thaw cycle to 53.5 wt.% after annealing.



Figure 5. A) A schematic for how the wear of hydrogels versus cartilage was measured. B) Micro-CT cross section images and C) the wear depth of cartilage and hydrogel samples after 10^6 cycles under 1 MPa of pressure, a spin rate of 100 mm s⁻¹, and with FBS as the lubricant (mean \pm SD). Numerical data for panel (C) listed in Table S5 (Supporting Information). D) The coefficient of friction between cartilage and the hydrogels during the tests.



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Fourier transform infrared spectra (see Figure S2, Supporting Information) show that there was a decrease in the frequency of the hydroxyl stretching peak (from 3330 to 3299 cm⁻¹) after annealing, indicating an increase in hydrogen bond formation.^[30,31] This peak shift and increase in hydrogen bond formation corroborates the increase in crystallinity measured with differential scanning calorimetry (DSC). These results show that PVA can crystallize within the nanofibrous BC network, and that these crystallites increase the solid content and strength of the hydrogel.

2.4. Effect of PAMPS on an Annealed BC-PVA Hydrogel

We previously reported that the incorporation of PAMPS into a BC-PVA hydrogel made with a freeze-thaw cycle resulted in an increase in the tensile and compressive strength of the hydrogel.^[15] Thus, we next sought to determine the effect of incorporating PAMPS into an annealed BC-PVA hydrogel. As shown in Figure 4A, the addition of PAMPS into the annealed BC-PVA hydrogel led to a decrease in the solid content relative to BC-PVA alone, from 0.53 to 0.37. DSC thermograms (see Figure S3C, Supporting Information) shows that after the addition of 10 wt.% PAMPS, the peak from melting crystalline PVA disappeared, indicating the addition of PAMPS destroys the PVA crystallites that form during the annealing process. The decrease in solid content and loss of crystallinity upon addition of PAMPS led to a decrease in the tensile strength (48.9-20.8 MPa), tensile modulus (444.8-150.5 MPa) and compressive strength (98.1–56.0 MPa in Figure 4B) of the hydrogel. The increase in water content of the hydrogel and loss of strength was likely due to the fact that PAMPS is a negatively charged polymer, and this negative charge results in an osmotic pressure that swells the hydrogel with water.^[15]

2.5. Wear and COF of Hydrogels against Cartilage

The wear resistance of a potential replacement for cartilage should exceed that of cartilage to ensure durability and minimize the generation of wear debris that could potentially cause an adverse biological reaction. We have previously shown that the wear resistance of a BC-PVA-PAMPS hydrogel is equivalent to that of cartilage and is superior to that of PVA or PVA-PAMPS when tested against a stainless-steel pin.^[15] These hydrogels were made by applying a freeze-thaw cycle to crystallize the PVA. Here we compare the wear resistance of PVAbased hydrogels (PVA, BC-PVA, and BC-PVA-PAMPS) that have been dried and annealed at 90 °C to that of porcine cartilage when tested against a porcine cartilage plug in fetal bovine serum (FBS). Figure S4 (Supporting Information) shows our pin-on-disc configuration for testing the wear of hydrogels in FBS. The porcine cartilage plug was rotated against the hydrogel surface 10⁶ times under 1 MPa of pressure and at a speed of 319 rotations per minute (maximum linear velocity was 100 mm s⁻¹). Figure 5A is a schematic illustration of how the wear test was performed.

Figure 5B shows cross-sectional X-ray microtomography (μ -CT) images of the hydrogels that were acquired in the







Figure 6. A) A schematic for how the wear of cartilage versus hydrogels was measured. B) Micro-CT cross section images and C) the wear depth of cartilage and hydrogel samples after 10^6 cycles under 1 MPa of pressure, a spin rate of 100 mm s⁻¹, and with FBS as the lubricant (mean \pm SD). Numerical data for panel (C) is listed in Table S6 (Supporting Information).



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center of the wear mark to measure the maximum wear depth. Figure 5C compares the wear depth of the hydrogels and cartilage. The wear depth of the BC-PVA hydrogel with 0% AMPS was 70.1 µm. The addition of 20% AMPS decreased the mean wear depth to $65.9 \,\mu\text{M}$, but the difference between the 0% and 20% AMPS samples was not statistically significant. This comparison illustrates that the negative charge and higher water content caused by incorporating PAMPS into an annealed BC-PVA hydrogel does not significantly improve the wear resistance. Both of these values were three times lower than the wear depth on the cartilage sample, which was 227.8 µм. The wear depth for annealed and rehydrated PVA was 301.0 µm, four times greater than either BC-PVA sample. These results indicate the presence of BC in the hydrogel can dramatically improve the wear resistance of an annealed PVA hydrogel to be superior to that of cartilage

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We also recorded the COF during the wear test. Cartilage maintained a constant COF of 0.020 during the test. The COF of BC-PVA decreased during the test from 0.040 to 0.021. The BC-PVA hydrogel with 20% AMPS had a similar COF as that without AMPS. In contrast, the COF of PVA increased dramatically during the test, from 0.033 to 0.135. Previous work has similarly demonstrated the COF of PVA against cartilage increases over time while the COF of cartilage against cartilage is constant.^[32,33] The increase in the COF for a PVA-Cartilage interface has been ascribed to transfer of damaged PVA to

the cartilage surface, which in turn decreases the ability of the cartilage surface to maintain a lubricating water layer.^[34] The incorporation of BC into PVA clearly inhibits damage of the hydrogel, allowing it to maintain a low coefficient of friction similar to that of cartilage during the wear test. The presence of PAMPS in the hydrogel is not necessary for maintaining a low COF and high resistance to wear.

It is critical that materials used for cartilage replacement on one side of the joint, i.e., on the femoral condyle, do not cause wear of cartilage on the opposing surface, i.e., the tibial plateau. Traditional orthopedic materials like cobalt-chrome and ultrahigh molecular-weight polyethylene are known to damage an opposing cartilage surface to a greater extent than hydrogels due to the higher COF and hardness of traditional orthopedic materials.^[34–36] To assess the wear caused by BC-PVA and BC-PVA-PAMPS hydrogels on cartilage, we created hydrogel plugs for wear testing (as described in Supporting Information). Hydrogel plugs were pressed against cartilage samples (see **Figure 6**A; Figure S4B, Supporting Information) with 1 MPa of pressure and rotated 10⁶ times at a speed of 319 rotations per minute (the maximum linear velocity at the circumference of the pin was 100 mm s⁻¹).

Figure 6B shows cross-sectional μ -CT images of the cartilage samples that were acquired in the center of the wear mark to measure the maximum wear depth. Figure 6C compares the wear depth on cartilage caused by the hydrogels or cartilage.



Figure 7. A) Results for shear testing of pig cartilage and hydrogels secured to metal pins with adhesive and a shape memory alloy clamp (mean \pm SD). Numerical data for the panel (A) is listed in Table S7 (Supporting Information). B–D) Images of samples after testing to failure. The osteochondral plug was extracted from a pig knee. The BC-PVA-PAMPS hydrogel was fabricated with the freeze-thaw process. The BC-PVA hydrogel was annealed at 90 °C and rehydrated.

The wear caused by the BC-PVA on cartilage ($247 \pm 16 \mu M$) was not significantly different from the wear caused by cartilage on cartilage ($228 \pm 12 \mu M$). The addition of PAMPS into the BC-PVA reduced the wear on the opposing cartilage surface to $81 \pm 27 \mu M$, significantly below the wear of cartilage on cartilage.

2.6. Shear Strength

We hypothesized that increasing the tensile strength of the hydrogel should also increase the shear strength. In order to be used for a cartilage replacement material, a synthetic hydrogel must be secured into a defect site with the same shear strength as the junction between cartilage and bone. One way to accomplish this goal is to have hydrogels that directly attach to bone or cartilage with sufficient strength, but this has not been achieved. Alternatively, the hydrogel can be attached to a metallic base, such as titanium, which has the ability to integrate with bone. As illustrated in Figure S5 (Supporting Information), we did this by first attaching freeze-dried BC to a metal rod with a combination of an adhesive and a shape memory alloy clamp, followed by infiltration of the hydrogel into the BC.

The setup used for shear testing is illustrated in Figure S6 (Supporting Information). **Figure 7** shows the results for the shear testing of a plug of porcine cartilage on bone extracted from a pig knee, a BC-PVA-PAMPS hydrogel made with the previous freeze-thaw process, and a BC-PVA hydrogel annealed at 90 °C and then rehydrated. Both of the hydrogels

are attached to stainless-steel rods with a combination of RelyX Ultimate cement and a shape memory alloy ring. The BC-PVA shear strength of 1.98 MPA (Figure 7A) is significantly greater than that of porcine cartilage (*p*-value from one-way ANOVA is <0.05). The average value of the shear strength for BC-PVA is also 40% greater than that of BC-PVA-PAMPS, but the error in the measurements is such that the difference in these values is not statistically significant. Comparison of the samples after failure shows that while pig cartilage was sheared completely off of the underlying bone (Figure 7B), both freeze-thawed BC-PVA-PAMPS and annealed BC-PVA were fractured on one side of the cylindrical sample but remained attached (Figure 7C,D). These results show that the shear strength of attachment for the annealed BC-PVA is greater than that of pig cartilage.

2.7. Application to an Implant for Partial Knee Resurfacing

Thus far we have described the compression strength and shear strength of BC-reinforced hydrogels attached to a metal pin with a diameter of 5.2 mm. While this size is convenient for testing, such a diameter is too small to serve as an implant for partial knee resurfacing. In addition, the samples lacked the curvature necessary to mimic the natural curvature of the femoral condyle. Thus, we sought to demonstrate the ability of the hydrogel to attach to a metal base with a size and shape representative of an implant for partial knee resurfacing.

Figure 8A show images of an implant 20 mm in diameter with a radius of curvature of 20 mm. An implant diameter of



Figure 8. A) Process for attaching the BC-PVA-PAMPS hydrogel to a titanium implant for B,C) treatment of osteochondral defects.



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20 mm is a typical size used for an osteochondral allograft, and a 20 mm radius of curvature is within the range of typical curvatures for the femoral condyle.^[37] In this case, five pieces of BC were cut into octagonal shapes with 8 legs to enable the BC to fold over the edge of the implant. A 0.25 mm-thick coating of commercially pure titanium was applied to the stem of the implant and underneath the base with a plasma spray process in order to improve integration with bone.^[38] Figure 8B,C show

an example of how such an implant would be used to replace a cartilage defect (Figure 8B). The surgeon would drill out a hole over the defect site that is complementary to the shape of the hydrogel-capped implant. The hydrogel-capped implant would then be pressed into the hole to replace the damaged cartilage.

3. Conclusion

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This work shows that reinforcement of an annealed PVA hydrogel with BC enables the production of the first hydrogel with a compression and tensile strength greater than cartilage. Annealing increased the tensile strength of BC-PVA by five times and the compressive strength by 1.8 times relative to a freeze-thaw process due to the greater crystallization and lower water content that was achieved by annealing. Reinforcement of PVA with BC lowered the wear of the hydrogel by four times relative to PVA alone, and three times relative to cartilage. The annealed BC-PVA hydrogel caused a minimal amount of opposing surface wear, similar to what was caused by cartilage on itself. Attachment of the BC to a metal plug via an adhesive and clamp, followed by infiltration and annealing of the PVA, enabled attachment of the BC-PVA hydrogel to a metal backing with a shear strength greater than the attachment of cartilage to bone. These advances in hydrogel strength and attachment enable the creation of an implant with a hydrogel surface and titanium backing that can enable durable resurfacing of damaged cartilage in an articulating joint.

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

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Conflict of Interest

B.J.W. and K.G. have an equity interest in Sparta Biomedical.

Data Availability Statement

The data that support the findings of this study are available in the supplementary material of this article.

Keywords

bacterial cellulose, cartilages, hydrogels, implants, polyvinyl alcohol

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