Meniscal injury is more likely to cause a permanent alteration of the biomechanical and biological environment of the knee joint, mainly due to the morphological mismatch and substantial loss of meniscal tissues. Herein, to overcome this challenge, we developed an improved bioink with enhanced printability, while maintaining the biocompatibility of major cellular component of the meniscus, namely fibrochondrocytes. Firstly, cellulose nanofiber (CNF) was mixed with gelatin-alginate thermal-responsive bioinks to improve the printability. Afterward, individual-specific meniscal prototypes based on the 3D reconstruction of MRI data were bioprinted using our bioink. The rheological and printability properties of the bioinks were characterized to select proper bioink content and bioprinting parameters. And then, a series of biological characterizations of the bioprinted samples, such as cell viability, metabolic activity, and extracellular matrix accumulation, were carried out in vitro. The results indicated that superior rheological performance and printability of CNF-modified bioink were achieved, ensuring high-precision bioprinting of specific-designed meniscal prototype when compared with the non-CN-containing counterparts. Meanwhile, biological tests indicated that fibrochondrocytes encapsulated within the CNF-modified bioink maintained long-term cellular viability as well as acceptable extracellular matrix accumulation. This study demonstrates that the CNF-modified bioink is in favor of the printing fidelity of specific meniscus by improved rheological properties, minimizing the mismatch between artificial meniscal implants and native knee joint tissues, thereby permitting the evolution of clinical therapeutic methods of meniscal reconstruction.

1. Introduction

The menisci are two semilunar fibrocartilage structures located at the medial and lateral surface of the tibial plateau, which play a vital role in stabilization, nourishment, and force distribution of the knee joint [1]. Unfortunately, menisci are commonly involved in acute knee injury and osteoarthritis, resulting in substantial loss of meniscal tissue and permanent alteration of the biomechanical and biological environment of the knee joint [2]. Generally, meniscal allografts are applied in clinical with acceptable long-term follow-up results [3]. However, like all other allograft transplantation treatments, meniscal allografts are limited to tissue resources and the potential risk of mismatch, immunoreactivity, and disease transmission [4].

Several tissue engineering scaffolds have been developed with encouraging progress in avoiding meniscal allografting-relevant complications [5]. However, the mismatch of meniscal defect size and baseline articular cartilage still exists, leading to worsening articular cartilage status and greater displacement during long-term follow-up [6]. It has been reported that 1/3 mismatch of the meniscus could result in a 65% increase in peak local contact stresses, significantly deteriorating the degenerative changes of the articular cartilage [7]. Thus, the patient-specific design meeting individuals’ joint requirements has been a vital element to
improve the biological and biomechanical performances of tissue engineering meniscal scaffolds.

Bioprinting technology as a group of additive, bottom-up, nature-like technologies has made it possible to spatially pattern cells, bioactive factors, and biomaterials in the 3D microenvironment [8–11]. Bioprinting meniscal-shaped structures with stem cells or chondrocytes via microextrusion bioprinting technology as proof-of-concept tests have been reported [12, 13]. However, there are few meniscal bioprinting studies involving patient-specific design using major cellular components of the native meniscus with proper mechanically featured bioink until now.

Gelatin-alginate-based hydrogel has been widely applied as a bioink source in microextrusion approaches [14, 15]. As a hydrolytic product of collagen, gelatin can provide a favorite microenvironment enhancing cell attachment and proliferation, while alginate presents the feature of instant gelation via cross-linking by calcium ions (Ca²⁺), enabling acceptable mechanical performance for bioprinted products [16]. Gelatin-alginate-based bioink systems present various advantages including widely available and economical material source, relatively simple producing process, and verified cytocompatibility [17]. Moreover, the thermal reversible feature of the gelatin-alginate-based bioink system enables a relatively simplified bioprinting process and device design [18]. However, during the microextrusion bioprinting process, the gelatin-alginate based hydrogels are fragile and changeable before further cross-linking, causing unstable mechanical condition especially in fabricating complex structures like the meniscus [19, 20].

Originating from the most abundant bioresource, wood biomass, cellulose nanofiber (CNF) is representative in building components in nature and has attracted significant interest as a potential biomaterial candidate [21]. In addition, CNF provides acceptable mechanical properties, relatively good rheological properties, and most importantly, cytocompatibility [22, 23]. However, to our knowledge, there has been no report of adapting CNF to improve the printability of alginate-gelatin-based bioink systems via hydrophobic interactions and hydrogen bonding for individual-specific meniscus biofabrication.

In this study, CNF was mixed with gelatin-alginate thermal-responsive bioinks to improve the printability. The rheological and morphological properties of the bioinks were characterized. While cell-laden meniscal prototype basing on MRI data was bioprinted using our CNF-modified bioinks, subsequently, the bioprinted prototypes underwent histological and immunohistochemical assessments to verify biological potentials as presented in Scheme 1. The current work aims at carrying out the preliminary practice on bioprinting implant prototype of patients’ specific-designed, cell-laden products for potential meniscus transplantation.

2. Materials and Methods

2.1. Materials. The bioprinter (BioPrinter-1, BP-1) used in the current study was developed by our research group. Gelatin type B from bovine skin (G8061) and Na-alginate (A9640) were purchased from Solarbio Inc. (China). Wood-based cellulose nanofiber powder was purchased from Guilin Qihong Technology Co. (China). High-glucose Dulbecco’s modified Eagle’s medium (DMEM), fetal bovine serum (FBS, HyClone), penicillin/streptomycin, and other reagents for cell preparation were all purchased from GE Inc. (China). A live/dead staining kit (BB-4126) was purchased from BestBio Inc. (China). An alamar blue assay kit

**Scheme 1**: Schematic representation of the study process.
Table 1: Proportion of ingredients in the tested hydrogels.

| Group | Abbreviation | Gelatin | Alginate | CNF |
|-------|--------------|---------|----------|-----|
| 1     | HGA-CNF      | 20% (w/v) | 1.25% (w/v) | 0.25% (w/v) |
| 2     | HGA          | 20% (w/v) | 1.25% (w/v) | — |
| 3     | LGA-CNF      | 10% (w/v) | 1.25% (w/v) | 0.25% (w/v) |
| 4     | LGA          | 10% (w/v) | 1.25% (w/v) | — |

(A7631) was purchased from Solarbio Inc. An immunohistochemistry tool kit (SP-0023), rabbit monoclonal primary antibodies for collagen type I (BS-10423R 1:400, 1 mg mL⁻¹), collagen type II (BS-10589R, 1:100, 1 mL mL⁻¹), and collagen type X (BS-0554R, 1:200, 1.4 mg mL⁻¹) were purchased from Thermo Scientific Bioss Inc. (China). Collagen type II (A7631) was purchased from Solarbio Inc. An immunohistochemistry tool kit (SP-0023), rabbit monoclonal primary antibodies for collagen type I (BS-10423R 1:400, 1 mg mL⁻¹), collagen type II (BS-10589R, 1:100, 1 mL mL⁻¹), and collagen type X (BS-0554R, 1:200, 1.4 mg mL⁻¹) were purchased from Thermo Scientific (China). Collagen type II (A7631) was purchased from Solarbio Inc. An immunohistochemistry tool kit (SP-0023), rabbit monoclonal primary antibodies for collagen type I (BS-10423R 1:400, 1 mg mL⁻¹), collagen type II (BS-10589R, 1:100, 1 mL mL⁻¹), and collagen type X (BS-0554R, 1:200, 1.4 mg mL⁻¹) were purchased from Thermo Scientific Bioss Inc. (China).

2.2. Preparation of Bioinks. CNF powder was diluted in deionized water to a final concentration of 0.25 wt% through stirring at 1000 rpm for 30 min at room temperature to acquire a homogenous solution. The solution was then heated to 70 °C, and the Na-alginate powder was added under magnetic stirring for 30 min. Gelatin powder was added and dissolved when the solution was cooled to approximately 50 °C under magnetic stirring for another 30 min. NaOH was used to adjust the pH of the solution to approximately 7.5. The hydrogel solution was then placed in a constant temperature oscillator at 200 rpm and 37 °C overnight to eliminate bubbles and maintain a homogeneous mixture. The bioink was prepared on a clean bench, and all the components and equipment were autoclaved and/or sterilized using UV light. After the preparation was finished, the final hydrogel was heated to 100 °C for 1 h and then sealed to keep it sterile. The bioink was stored at 4 °C and heated to 37 °C for 30 min in a water bath before the bioprinting process. To verify the printability and biocompatibility of the bioinks, six groups of hydrogels were prepared with varying concentrations of gelatin and CNF. To simplify the narration, hydrogels containing higher and lower content of gelatin were recorded as high-gelatin-containing alginate group (HGA) and low-gelatin-containing alginate group (LGA), respectively. The CNF-modified hydrogels were named as HGA-CNF and LGA-CNF, respectively (Table 1).

2.3. Rheological Characterization. The rheological analysis was performed using an MCR 702 rheometer (AtonParr, Austria) using a 25 mm parallel plate. All 4 groups of cell-free bioink samples were tested before and after cross-linking. 0.1% (w/v) CaCl₂ solution was used as a cross-linker. The cross-linking time was 1 min. The samples were stored at 37 °C before the test and heated to the experimental temperature for 5 min before the rheological measurements. Non-cross-linked hydrogels were characterized through flow and temperature sweeps. The flow sweeps were conducted to evaluate viscosity at a shear rate ranging from 0.1 to 100 s⁻¹ at 25 °C. To evaluate the thermal responsivity of the hydrogels, the viscosity was measured at temperatures ranging from 1 °C to 40 °C, at a fixed shear rate of 1 s⁻¹. The cross-linked hydrogels were characterized through oscillatory frequency sweeps between 0.1 and 10 Hz in the linear viscoelastic region of the hydrogels at 37°C.

2.4. Cell Preparation. Primary rabbit fibrochondrocytes (rFCs) were used in bioprinting the cell-laden meniscal prototypes. The isolation and culture of rFCs have been described by previous studies [24]. Briefly, sequential treatment of finely diced meniscus (from 6-month-old rabbits) using 0.5% hyaluronidase, 0.2% trypsin, and 0.2% collagenase was applied to obtain the cells. Then, the cells were filtered through a fine nylon mesh to separate them from undigested tissue and debris. The filtrate was washed several times through gentle suspension in D-Hank’s solution and underwent centrifugation at low speed (1000 × g) to form a cell pellet. The cells were then cultured under conventional conditions at 37°C in a humidified atmosphere of 5% CO₂. Once a sufficient cell quantity was achieved, usually after the third or fourth passage, the cells were used for further bioprinting process.

2.5. Design of Microextrusion Bioprinter. In the current study, a microextrusion-based bioprinter (BP-1) was developed by our group. The minimum mechanical travel distance is ±0.05 mm for the XY-axes and ±0.01 mm for the Z-axis. The extrusion module was consisting of a precision piston for extruding materials, a syringe heater for thinning printing materials, and a standard medical syringe (10 mL) as a cartridge with 22G metal nozzles for depositing bioinks. The whole printing system is enclosed by a clean bench with a temperature-humidity control system. A UV light source was also integrated for sterilization. In a typical print procedure, the XY stage will translate the extruder at speeds ranging from 1 mm/s to 10 mm/s to pattern the predetermined 2D cross-section on the printing substrate, and then the Z stage elevates to repeat the process for the next layer. Special methods were applied to ensure the sterilization of the whole bioprinting system. Firstly, the whole bioprinter was installed in a modified super clean bench which was divided into two separate cabins by a plate. The mechanical structures of the bioprinter were in the upper cabin where the bioprinting process was carried out and the control system was installed in the cabin below. All cables in the upper cabin were waterproof. Secondly, before the bioprinting process, thorough disinfection by alcohol spray and UV irradiation was carried out in the upper cabin for 40 min. At last, the super clean bench was kept closed the positive pressure during the bioprinting process.

2.6. Printability Verification Process. The bioinks were heated in a water bath at 37°C for 30 min and extracted into a standard sterile syringe with 22G (0.42 mm inner diameter) metal needles. The samples were printed directly onto sterile Petri dishes on the printing platform, which was cooled down to 4°C by the printing platform during the process. Square blocks (15 mm × 15 mm × 2 mm) consisting of four perpendicular layers with a 40% filling rate were 3D printed. Considering the nonlinear rheological performance of living cells, the samples used for printability tests were cell-free to reduce disturbance. The printing temperature range was
20°C–25°C. The best printing temperature was decided on the base of the actual filamentation of each group under a certain temperature. The platform temperature for all groups was set to 4°C. The moving speed of the nozzle in XY-axes was 3 mm/s. All samples were cross-linked immediately using 0.1% w/v CaCl₂ for 1 min and then washed twice using Dulbecco’s phosphate-buffered saline (DPBS).

2.7. Patient’s Specific Design of Meniscal Prototype. To mimic the native shape of the meniscus, a 3D model of the medial meniscus was achieved based on MRI scans from one healthy volunteer of the authors. MRI images were acquired through the 3Tesla Discovery MR750 GEM Magnetic Resonance Imaging System (GE Healthcare, United Kingdom) with a 16-element phased-array flexible coil. A general supine position with a relaxed limb was used when performing the scan. Sagittal 3D fast-spin-echo (FSE) with high spatial, contrast resolution and no chem SAT was applied for image achievement. MRI scan parameters for the knee imaging are listed in Table 2.

The MRI data were saved as a sequence of Digital Imaging and Communications in Medicine (DICOM) files and processed via the Mimics software (Materialise 21, Belgium). The threshold was set to 50-200 GV to identify meniscal tissues from surrounding tissues. The model was then carefully trimmed artifically by an experienced orthopedic surgeon to reconstruct the morphology of the medial meniscus. After that, an automatic adjustment was applied to fill the cavities caused by signal noise and exported as an STL file. The STL model was then converted to a motion program (G-code) using an open-source software (Repetier-Host, Germany), which describes the operation of the dispensing nozzles, printing speed, nozzle heat, and other relative information. A different open-source software (Printrun, by Kliment Yanev) was applied to transfer the G-code file to the printing computer for bioprinting.

2.8. Bioprinting Process. After the printability verification process, HGA-CNФ was selected as the bioink used for further bioprinting tests. Centrifuged rFCs were resuspended with 5 mL liquefied HGA-CNФ directly. The cell number of the resuspension solution was 5 × 10⁶. Slight oscillation of the mixed bioink was performed for 10 min at 37°C to homogenize the solution. The cell-laden bioinks were extracted into syringes with 22G (0.42 mm inner diameter) metal needles. To eliminate the bubbles in the bioinks, all syringes were settled in the incubator at 37°C for 20 min. The bioprinting process was conducted with BP-1, and the structures were printed directly onto sterile Petri dishes on the printing platform, which was cooled down to 4°C during the process. Since it has been reported that cellular activities would be compromised when encapsulated into solidly structured hydrogel scaffolds [25, 26]. To improve the nutrition exchange inside the hydrogel-based scaffolds, all structures were designed based on the lattice structure. The printing temperature was 25°C. The platform temperature was set to 4°C. The 3D-reconstructed meniscus model with a 40% filling rate and rectilinear filling patterned was bioprinted. The printing speed was 3 mm/s. All samples were cross-linked immediately after bioprinting using 0.1% w/v CaCl₂ for 1 min. To simplify the process, cell-laden, square block samples were bioprinted for metabolic activity test. The bioprinted samples were cultured in a petri dish (φ = 55 mm) and incubated for 14 days. The medium was changed every 3 d.

2.9. Printability Evaluation and Characterization. The printing quality was evaluated by imaging the cross-linked cell-free samples using a ×4 air objective with an SX16 stereoscopic microscope (OLYMPUS, Japan), and bioinks from all groups were used in this test. Briefly, cross-linked cell-free structures with four layers were imaged to measure the strand width and pore size using the ImageJ software. The spreading ratio was calculated by dividing the strand width by the inter diameter (22G, ~420 µm) of the nozzle. The printed structures were stained using trypan blue after printing for better visualization. Scanning electron microscopy (SEM, Zeiss EVO 18, Cambridge, England) was applied to further characterize the porosity and interconnectivity of the bioprinted samples. The samples were lyophilized and sputter-coated with gold to observe the topography. Images (×25 and ×1500) were taken using SEM at an accelerating voltage of 3 keV to evaluate the morphology and microarchitectures. Three random images were taken from each sample from every region. To better visualize the rFCs encapsulated inside the structures, three samples were first graded ethanol dehydrated and then went through the same processes as other samples.

2.10. Cell Viability and Metabolic Activity. The cell viability of cells in the bioprinted structures was qualitatively analyzed at day 1, as well as 4, 7, and 14 d of culture using a live/dead staining kit in accordance with the manufacturer’s instructions. Briefly, the bioprinted structures were rinsed using DPBS twice and incubated in 5 mL of live/dead solution (Calcein-AM and ethidium homodimer) at 4°C for 20 min. The samples were examined using a fluorescence microscope (Zeiss, Axio Imager 2, Germany). Five randomly selected images obtained at ×4 magnification were used to evaluate the percentage of cell viability via ImageJ. The viability was calculated as the average ratio of live cells to total cells. An alamar blue assay kit was used to measure cell proliferation. The metabolic activity of the encapsulated cells was analyzed at 1, 4, 7, and 14 d after bioprinting. The square block bioprinted structures were incubated in 6-well plates for 7 h with 0.3 mL of alamar blue solution for each 3 mL of culture media. Fluorescence intensity was measured at an excitation

### Table 2: MRI parameters for knee scanning.

| FOV, mm | Matrices | TR, ms | TE, ms | Slice thickness, mm |
|---------|----------|--------|--------|---------------------|
| 180 × 180 | 320 × 320 | 1500 | 32 | 0.8 |

Bandwidth, kHz

| Echo train length | NEX | Scan time, mins |
|-------------------|-----|-----------------|
| 83.33             | 60  | 1               | 6.29 |

FOV: field of view; TR: repetition; TE: echo time; NEX: number of excitations.
2.11. Histology and Immunohistochemistry. The prototypes incubated for 14 d in the culture medium were fixed in paraformaldehyde/barium chloride overnight. The fixed samples were then cryosectioned, embedded in paraffin wax, sectioned at 5 μm, and affixed to microscope slides. Hematoxylineosin, picrosirius red, and alcian blue staining were applied to assess calcium, collagen, and glycosaminoglycan accumulation, respectively. Immunohistochemistry techniques were applied to evaluate and identify collagen types I, II, and X, as previously described [27]. Briefly, samples were treated using peroxidase and chondroitinase ABC and then incubated with fetal calf serum to reduce nonspecific binding. Collagen type I, collagen type II, and collagen type X rabbit monoclonal primary antibodies were applied for 1 h at room temperature. The secondary antibody (anti-rabbit IgG biotin conjugate) was added for 1 h, followed by incubation with an ABC reagent for 45 min. Finally, sections were developed using DAB peroxidase for 5 min.

2.12. Statistical Analysis. Results are expressed as mean ± standard deviation. Statistical analyses were performed using the SPSS v22.0 software (SPSS Inc., Chicago, IL, USA).

3. Results

The rheological behavior is considered as the major concern to select the bioink candidate for further bioprinting tests. All
four groups of prepared hydrogels with varying gelatin and CNF content were tested (Figure 1). The samples showed a significant decrease in viscosity as the temperature increased during a temperature sweep (Figure 1(a)). Hydrogels containing a higher concentration of gelatin (20% w/v) maintained higher viscosity among all the groups when the temperature raised to 25°C, which is the minimum acceptable printing temperature for the bioprinter. CNF-modified hydrogels exhibit prior viscosities and lower decreasing rates than their non-CNF-containing counterparts. At the designated bioprinting temperature window of 25°C to 30°C, HGA-CNF exhibited more than a threefold increase in viscosity compared with HGA (Figure 1(b)), indicating prominent superiority in printability potential. A shear-thinning behavior was observed among all the bioink candidates. As CNF strengthened the viscosity, hydrogels modified with CNF tended to maintain more solid-like behaviors than non-CNF-containing hydrogels at the same shear rate (Figure 1(c)). The frequency sweep also showed a similar tendency, and hydrogels modified with CNF showed higher storage modulus than pure gelatin with alginate when being cross-linked by Ca²⁺, indicating priority in the mechanical performance of CNF-modified samples (Figure 1(d)).

Printability is the vital access to maintain the fidelity of the meniscal prototype. Cell-free square block samples and rFCs cell-laden meniscal prototypes were printed using our microextrusion-based bioprinter (Figure 2). The quality assessments and characterization of the printed structures are presented in Figure 3. To acquire better printing filamentation, all the hydrogels were tested under a printing temperature range from 20°C to 25°C, which was decided based on the rheological results of the hydrogels (Figure 3(a)). HGA-CNF and HGA showed the best filamentation under 25°C while the highest acceptable printing temperature for LGA-CNF and LGA was 20°C. The other printing parameters remained the same for all hydrogels. Top-view images are presented in Figure 3(b). The results showed that only HGA-CNF and HGA formed 3D lattice structures with relatively acceptable fidelity and integrity. Samples bioprinted with LGA showed completely fused constructs during the test. Thus, the measurement of the filament was not performed for LGA. To analyze the bioprinted samples in more detail, stereomicroscope images of the top layers were taken. The strand width and pore size were measured (Figures 3(c) and 3(d)). The spreading ratios of HGA-CNF, HGA, and LGA-CNF were 0.88 ± 0.08, 1.07 ± 0.03, and 1.03 ± 0.07, respectively. The strands of the CNF-modified hydrogels (HGA-CNF and LGA-CNF) were stretched at different levels during the printing process, which led to a thinner strand width than their non-CNF-containing counterparts. The pore size measurement further confirmed this feature. The presence of CNF resulted in a thinner strand width and better lattice formation in the printed structures. Structural investigations of the printed structures of cell-laden HGA-CNF via scanning electron microscopy (SEM) are presented in Figure 3(e). Hierarchical porous structures of micron order were observed, presenting a similar microenvironment to the morphological features of the extracellular matrix (ECM). The sign of cellular component in the bioink was more distinct after treating the samples with graded ethanol dehydration. The bioink scaffold and rFCs encapsulated inside were colored as green and fuchsia, respectively, for better visualization. Homogeneous distribution of rFCs within the hydrogel-based scaffolds was observed, indicating the biocompatibility of our bioprinting process. In summary, good printability of HGA-CNF was demonstrated, and this was then selected as a bioink for further bioprinting tests.
volumetric information and then bioprinted into a hydrogel-based, cell-laden meniscal prototype (Figure 4). The selected bioink (HGA-CNF) presented good printability and shape fidelity during the printing process. The 3D-printed meniscus prototype was consisting of 19 layers, consuming approximately 991.85 mm$^3$ volume of bioink. The actual size of the printed meniscus prototype was approximately 48.3 × 40.5 × 8.28 mm before cross-linking, while the designed shape of the 3D model reconstructed from MRI data was 45.42 × 41.13 × 8.81 mm, indicating highly consistency.

The viability and metabolic activity of rFCs in the printed samples are presented in Figure 5. The results showed very high viability after the bioprinting process in a live/dead staining assay (Figure 5(a)). In detail, the average cell viability was very high.
of the rFCs on day 1 was 94 ± 0.03%. However, the bioprinted cells recovered from the printing shock within 4 d of culture (95 ± 0.05%) and retained high performance throughout the culture period. When it came to day 7, the viability of the rFCs reached a peak of nearly 99% and then dropped to 96 ± 0.02% for the following 7 d of culture, demonstrating the excellent biocompatibility of HGA-CNF (Figure 5(b)). The alamar blue assay of the bioprinted rFCs (Figure 5(c)) was in line with the viability results. The metabolic activity of the printed cells was initially hampered by the bioprinting process and then recovered to a significantly high level at 4 d after bioprinting, indicating high cell proliferation activity during the first few days. The metabolic activity decreased gradually between 4 and 7 d of culture and stabilized at day 14.

To ascertain the deposition of collagen and sGAG in vitro, bioprinted samples encapsulating rFCs were histologically stained after 14 d of incubation (Figure 6(a)). Picrosirius red and alcian blue staining was positive in the intercellular space of rFCs encapsulated in the bioprinted samples, indicating collagen and glycosaminoglycans accumulation. HE staining of the bioprinted structures revealed rFCs were homogeneously distributed in the bioink. Immunohistochemical staining showed barely signs of collagen type I and abundant accumulation of collagen types II and X (Figure 6(b)).

4. Discussion

To meet the growing demand for meniscus tissue engineering, both synthetically and naturally derived biomaterials have been reported [28]. However, although it has been reported that shape mismatch can lead to a negative influence on the biomechanical performance of implanted meniscus [29, 30], the morphological design of the meniscus and development of relative fabrication approaches were unvalued to some
extent. To address this situation, the precise morphological design was stressed in this study. A 3D-reconstructed model was achieved from the patient’s MRI images, aiming at accurately revealing the native shape parameters of the meniscus. During the MRI scanning, modified parameters were applied to acquire better accuracy. Generally, OAx fs PD, OSag fs PD, and OCor fs PD models were used for a meniscal scan while our scan was a combination of these three models with a thinner slice thickness. The time consuming for a general meniscal MRI scan was approximately 5.23 min while our modified model was 6.29 min. The modified parameters were prestored in the MRI device and could be readily switched as needed. Neither additional equipment nor extra economic cost was needed for the modified scan. Moreover, with the assistance of the 3D reconstruction software like Mimics, the meniscal model could also be achieved via mirroring the healthy side MRI data of the patient when the major substance of injured meniscus has been damaged. As for partial meniscal injure cases, a mirroring of the healthy side MRI data could be used as a reference. The injured area could be underlined via a matching between the healthy and injured meniscus.

Figure 5: Viability and metabolic activity of bioprinted samples. (a) Live/dead staining of rFCs immediately at 1, 4, 7, and 14 d of culturing in vitro. Living cells are stained green and dead cells red. Scale bar 100 μm. (b) Quantitative statistics of the bioprinted samples at different time points using ImageJ. (c) Metabolic activity values of bioprinted rFCs using an alamar blue assay at different time points.
Benefitting from a bottom-up style of manufacturing and precise cell patterning, bioprinting is naturally suitable for biofabricating tissue engineering scaffolds with complex structures. In this process, the balance between printability and biocompatibility has been a major challenge since the early development of bioinks. In this context, a microextrusion-based bioprinting system was built by the current research group to explore its potential of personalized-designed meniscal reconstruction. Gelatin-alginate-based bioink was adapted to achieve thermoresponsive and calcium ions cross-linking gelling features [31]. However, the thermal curing of gelatin is unstable and reversible during bioprinting, which commonly leads to inferior printability and fidelity when fabricating complex structures. Thus, maintaining a self-supporting hydrogel before ionic cross-linking is also an attractive feature that would improve printability.

The rheological characterization of the prepared hydrogels further confirmed the influence of CNF. Hydrogels containing CNF showed prior viscosity performance than their non-CNFe-containing counterparts at a wide range of temperatures. This phenomenon could be attributed to the physicochemical interaction between CNF, alginate, and gelatin. It has been reported that hydroxyl surface groups exist in CNF and alginate can interact through hydrogen formation with carbonyl and amine groups in gelatin [32], improving the gelling strengthen of the hydrogel. Moreover, the existence of CNF also significantly improved the shear-thin behaviors of the hydrogels via the breakage of fiber networks under shear force [33]. In addition, nanofibers of CNF underwent shear-induced alignment from the nozzle, which further enhanced the bioink flow [34], resulting in better printability. The role of CNF in the bioink is a “booster” for both the extrusion and curing process through enhancing the shear-thinning behavior. During the extrusion process, the extrusion flow through the nozzle was enhanced by CNF via shear-induced fibril alignment, which results in a smoother and more continuous flow out of the nozzle. When the gelatin had undergone gelation, the CNF material present resulted in relatively high stiffness and self-supporting features through the formation of a self-aligned fibrous network scaffold which served as a stiff filler with favorable shear-thinning properties [35]. To enhance the printability of gelatin-alginate bioink, 0.25% w/v CNF was added in the current study. The bioprinted meniscus prototype can maintain shape fidelity without being instantly cross-linked.

Origin from natural resources, the biocompatibility of gelation and alginate has been verified [36]. As for CNF, although it is not degradable inside the human body, it has been reported that owing to the small size and inertness of CNF, the newly formed matrix simply surrounds the fibers and integrates them into the tissue without any adverse side effects [37]. The density value of CNF in this study is

![Figure 6: Histological and immunohistochemical results of bioprinted structures. (a) Histological staining of bioprinted rFCs structures with HE, picrosirius red, and alcian blue. (b) Immunohistochemical for collagen I, II, and X accumulation in the bioprinted rFCs samples.](image)
towards the reconstruction of the patient like the native meniscus. This progress will pave the way
neously. The results showed the ECM components were sim-
drocytes, sGAG and collagen types II and X were con
fi
knowledge, this is the
bioactive meniscal substitution. Moreover, to the best of our
advance of bioprinting and biomaterial approaches.
clinical implantation demands would not be a barrier as the
prototype with proper biomechanical features meeting the
performances of the meniscal prototype, coprinting of ther-
performance. Considering the inferior mechanical properties
primary fibrochondrocytes have survived through the bioprinting process and
viability and biofunction during further in vitro incubation. The bioprinted meniscal prototype would bring
ability bioprinting meniscal relevant cellular components of
our system. Although applying stem cells in meniscal recon-
construction has been reported by some groups. However,
complex-staged-inducing methods were required for stem
cells in fibrocartilaginous differentiation, consuming a rela-
tively long period of cell culture which compromised the
clinical potential [39]. Thus, mature cells were preferred by
our group. Nevertheless, as the progress of coculturing stem
cells with other meniscal relevant cells such as fibrochondro-
cytes and chondrocytes in 3D environments are reported
recently [40, 41], further experiments on bioprinting mixed
cell groups to enhance the proliferation and differentiation
will be carried out by our group.

To further assess the ECM accumulation of fibrochon-
drocytes, sGAG and collagen types II and X were confirmed
by histological and immunohistochemical tests simulta-
nously. The results showed the ECM components were sim-
ilar to the native meniscus. This progress will pave the way
wards the reconstruction of the patient’s specific-designed, 
bioactive meniscal substitution. Moreover, to the best of our
knowledge, this is the first time that primary fibrochondro-
cytes have survived through the bioprinting process and
maintained viability and biofunction during further in vitro
incubation. The bioprinted meniscal prototype would bring
more flexible, adaptable, and easily accessible methods in
meniscal repairment and reconstruction.

However, the current study still has some limitations.
Firstly, further biomechanical tests of cell-laden meniscal
prototypes are still needed to evaluate the in vitro mechanical
performance. Considering the inferior mechanical properties
of hydrogel-based bioinks, to further improve the mechanical
performance of the meniscal prototype, coprinting of ther-
oplasty polymer and bioink would be our further direction
[11]. Benefiting from the highly scalable of our bioprinter,
the modification would be easily accessed. Moreover, in vivo
implantation on animal models should be carried out further
to verify the tissue formation abilities of our bioprinted proto-
types. We predict that biofabrication of cell-laden meniscus
prototype with proper biomechanical features meeting the
clinical implantation demands would not be a barrier as the
advance of bioprinting and biomaterial approaches.

5. Conclusions

The CNF-modified gelatin-alginate bioink has achieved
encouraging results in mechanical performance and print-
ability. Combining with MRI 3D reconstruction, the prelimi-
inary attempt of applying modified bioink in bioprinting
individual-specific meniscus prototype is achieved with favor-
able biofunction. The application of bioprinting approaches
might minimize the mismatch between artificial meniscal
implants and native knee joint tissues, thereby permitting
the evolution of clinical therapeutic methods of meniscal
reconstruction.

Data Availability

The data used to support the findings of this study are avail-
able from the corresponding author upon request.

Conflicts of Interest

The funders had no role in the design of the study; in the col-
collection, analyses, or interpretation of data; in the writing of
the manuscript; or in the decision to publish the results.

Authors’ Contributions

Conceptualization was done by Wenbin Luo and Zhengyi
Song; methodology by Wenbin Luo; formal analysis by Wen-
bio Liao and Zhenguo Wang; investigation by Zuhao Li and
Chenyu Wang; resources by He Liu; data curation by Jinch-
eng Wang and Qingping Liu; writing original draft prepara-
tion by Wenbin Luo; visualization by He Liu and Qingping
Liu; supervision by Jincheng Wang and Qingping Liu; project
administration by Jincheng Wang and Qingping Liu. All
authors have read and agreed to the published version of
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Supplementary Materials

Supplementary 1. Video S1: bioprinting process of the
meniscal prototypes.

Supplementary 2. Video S2: bioprinting process of the square
block samples.

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