Materials Today Bio 27 (2024) 101135

Contents lists available at ScienceDirect

Materials Today Bio

journal homepage: www.journals.elsevier.com/materials-today-bio

Light-based 3D bioprinting technology applied to repair and regeneration of different tissues: A rational proposal for biomedical applications

Wenzhuo Fang [a,1], Zhenwei Yu [a,1], Guo Gao [b], Ming Yang [a], Xuan Du [b], Ying Wang [a,**], Qiang Fu [a,*]

[a] Department of Urology, Shanghai Sixth People's Hospital Affiliated to Shanghai Jiao Tong University School of Medicine, Shanghai Eastern Institute of Urologic Reconstruction, Shanghai Jiao Tong University, Shanghai, 200233, China

[b] Key Laboratory for Thin Film and Micro Fabrication of the Ministry of Education, School of Sensing Science and Engineering, School of Electronic Information and Electrical Engineering, Shanghai Jiao Tong University, Shanghai, 200240, China

ARTICLE INFO

Keywords:
Light-based 3D bioprinting 1
Bio-ink 2
Tissue engineering 3
Photo-crosslinked polymers 4
Hydrogel 5

ABSTRACT

3D bioprinting technology, a subset of 3D printing technology, is currently witnessing widespread utilization in tissue repair and regeneration endeavors. In particular, light-based 3D bioprinting technology has garnered significant interest and favor. Central to its successful implementation lies the judicious selection of photosensitive polymers. Moreover, by fine-tuning parameters such as light irradiation time, choice of photoinitiators and crosslinkers, and their concentrations, the properties of the scaffolds can be tailored to suit the specific requirements of the targeted tissue repair sites. In this comprehensive review, we provide an overview of commonly utilized bio-inks suitable for light-based 3D bioprinting, delving into the distinctive characteristics of each material. Furthermore, we delineate strategies for bio-ink selection tailored to diverse repair locations, alongside methods for optimizing printing parameters. Ultimately, we present a coherent synthesis aimed at enhancing the practical application of light-based 3D bioprinting technology in tissue engineering, while also addressing current challenges and future prospects.

1. Introduction

With the advancement of technology and medical treatment, many medical problems have been overcome. However, due to the restricted regenerative capacity of the human body, especially for damage to vital organs and aging, it is still not possible to effectively repair and induce regeneration of damaged organs and tissues in the medical community [1]. In addition, the limited number and quality of organ donors make it extremely difficult for the organ transplantation approach to address the immediate needs [2]. Tissue engineering technology, as one of the emerging technologies in recent years, has been discovered and demonstrated to facilitate the repair and regeneration of damaged tissues and organs, which has aroused great curiosity among researchers and clinicians. This technology is now considered to be one of the most effective approaches to address the difficulties of tissue and organ repair and the limitations of organ transplantation in clinical practice [3,4]. Tissue engineering technology is designed to integrate biology and engineering techniques to simulate the structure and biological function of the human body as much as possible, to repair or even replace diseased or aging organs and tissues [5]. Cells, scaffolds and growth factors, as the three elements of tissue engineering in the past, often played an integral role in tissue repair. However, with the advancement of science and technology, the current three elements of tissue engineering are mostly cells, bioinks and deposition strategies. In addition, materials cannot provide cellular elements, which requires researchers to additionally incorporate different types and amounts of cells depending on the different properties of the constructs [6–8].

3D printing (3DP), also known as additive manufacturing, is a technology that constructs objects based on digital models using materials such as powdered metals or plastics by adding material prints layer by layer [9]. 3D bioprinting technology (3DBP), as a branch of 3D printing technology, has the same basic printing principles as 3D printing but is more stringent in the selection of bio-inks and the printing technology requirements [10,11]. They often require the precise

* Corresponding author.
** Corresponding author.
E-mail addresses: sdzbbswangying@alumni.sjtu.edu.cn (Y. Wang), jamesqfu@aliyun.com (Q. Fu).
[1] These authors contributed equally to this work.

https://doi.org/10.1016/j.mtbio.2024.101135
Received 2 February 2024; Received in revised form 10 June 2024; Accepted 21 June 2024
Available online 26 June 2024

assembly of materials with high biocompatibility and cells or bioactive factors, etc. This demands that they intelligently distribute the distribution of growth factors to meet the needs of tissue regeneration at the same time as ensuring high bioactivity of the cells [12,13].

Among the many 3D printing platforms currently in use, light-based 3D bioprinting technology has also gradually gained extensive development and applications. Among them, digital light processing (DLP), stereolithography (SLA), and two-photon polymerization (2 PP) technologies have been widely studied and popularized [14,15]. Compared with other types of 3D bioprinting technologies, such as fused deposition modeling (FDM) and selective laser melting (SLM), light-based 3D bioprinting technologies have great advantages in terms of maintaining cell viability and tunability of printing performance [16,17]. For example, SLA technology is one of the most widely used light-based 3D bioprinting technologies due to its good biocompatibility and high precision and spatial resolution [18]. In addition, the printing process of photopolymerization-based printing methods is gentler, and the molding can be cured quickly at room temperature [19]. Researchers can also control the printing process more conveniently by varying the time and intensity of light exposure, and customizing the printed scaffolds to match the requirements of the implant site [20]. Among the numerous components of light-based 3D bioprinting, photosensitive polymers are frequently the most noticeable and directly affect the characteristics of the final printed product. For researchers who want to customize the appropriate light-based 3D bioprinting materials for different repair sites, choosing the appropriate photosensitive polymers is the first and one of the most significant steps [21].

In this review, we begin by elucidating various properties of photosensitive polymers, offering insights into common photo-crosslinked bio-inks and their diverse biomedical applications across different repair sites. Subsequently, we delve into the innovative strategies employed by researchers to fine-tune scaffold properties, such as wavelength and exposure time adjustments of light sources, optimization of photoinitiators, and control over material pore size during the 3D bioprinting process. Finally, we provide a comprehensive summary of the extensive applications of photo-crosslinked materials in tissue engineering, deliberate on the challenges associated with bioprinting and photo-crosslinked bio-inks, and offer perspectives on their future development. (Fig. 1).

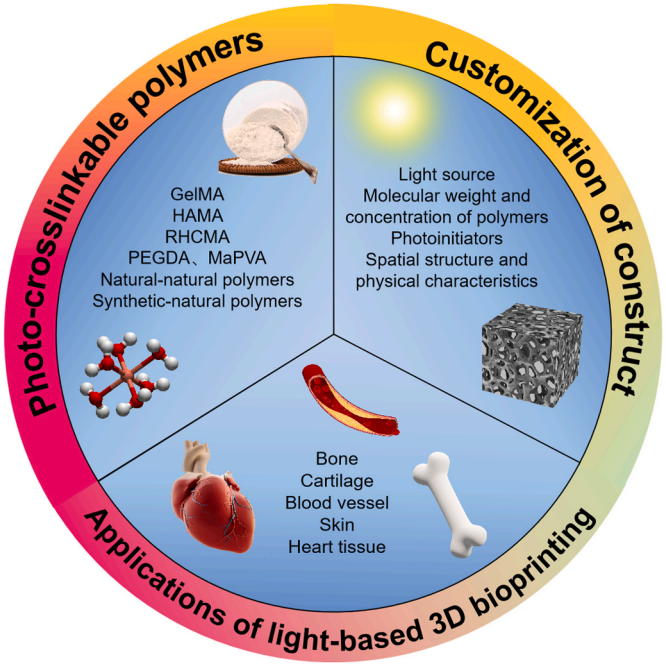

Fig. 1. Overview of polymer selection, customization strategies and their applications for photocrosslinking-based 3D printing technology.

2. Photo-crosslinkable polymers

Photo-crosslinkable polymers are a class of polymers that undergo crosslinking reactions in the presence of UV or visible light via photoinitiators or photosensitive monomers [22]. As shown in Table 1, they have a range of characteristics that have led to their widespread use in printing three-dimensional tissue engineering constructs. However, photocrosslinking groups are basically absent in naturally occurring polymers, so in order to make compounds with photocrosslinking properties, researchers often modify photosensitive groups on polymers from different sources. Therefore, we categorize polymers into two main groups, natural and synthetic polymers, according to the source of the polymer prior to modification, and present strategies for forming composite polymers by modifying different kinds of polymers with photosensitizing groups. In addition, we provide a brief overview of examples of photo-crosslinkable polymers, which are currently mainly used in tissue engineering, with accompanying descriptions of the properties of each material.

2.1. Natural polymers

2.1.1. Gelatin

Gelatin, as one of the most common natural polymers, is widely used in tissue engineering due to its low antigenicity and good biocompatibility, as well as its efficient absorption in the body without toxic degradation [29]. However, gelatin itself does not have photosensitizing properties, so in light-based 3D bioprinting, gelatin can be functionally modified to make it photosensitive. A common practice is to introduce photosensitizing groups into the molecular structure of gelatin to make it responsive to specific wavelengths of light. Gelatin methacryloyl (GelMA) is a polymer material that introduces methacryloyl groups into gelatin molecules. The synthesis procedure is as follows: Firstly, gelatin

Table 1
Several common characteristics of photo-crosslinked polymers.

Characteristics	Specific features	The type of polymers	References
Controllability	The cross-linking reaction of photo-crosslinked polymers is usually controllable, and precise control of the degree of cross-linking can be realized by adjusting the intensity, time and wavelength of light.	Nature polymers, Synthetic polymers	[23,24]
Reversibility	Some photo-crosslinked polymers are reversibly photosensitive, i.e. they can undergo repeated light-induced cross-linking and de-cross-linking under different conditions. This reversibility allows the material to be used multiple times.	Nature polymers	[25]
Local crosslinking	Due to the directionality of light, photo-crosslinked polymers allow for spatially selective cross-linking reactions, enabling localized modification of specific regions.	Nature polymers	[26]
High resolution	Photo-crosslinked polymers perform well in a number of nanofabrication and 3D printing fields because the wavelength of light and the focusing system of the beam can significantly change the resolution of the print.	Nature polymers	[27,28]

powder was added to an appropriate amount of a mild buffer solution (e.g. phosphate buffer) and stirred to make it uniformly dispersed. Subsequently, methacrylic anhydride is slowly added dropwise to the gelatin solution with continuous stirring to allow for sufficient reaction. The reaction mixture is allowed to react for some time at an appropriate temperature, usually room temperature or slightly above room temperature. Upon completion of the reaction, the reaction mixture is neutralized with an alkaline buffer solution (e.g., sodium hydroxide solution) to neutralize the unreacted methacrylic anhydride. Finally, the reaction mixture is washed and purified repeatedly to remove unreacted compounds and by-products to obtain purified GelMA hydrogels [30]. Albrecht et al. [31] successfully printed soft adipose tissue by adjusting the composition and handling of GelMA during printing according to different cell types in order to homogeneously mix human primary mature adipocytes (MA) and adipose-derived stem cells (ASCs), respectively. Zhang et al.[32]developed a novel corneal decellularized extracellular matrix/gelatin methacryloyl (CECM-GelMA) bio-ink as an artificial cornea customized to personalize the cornea using digital light processing (DLP) 3D bioprinting technology. The hydrogel was also loaded with human corneal fibroblasts (hCFs), which showed excellent performance in promoting epithelial regeneration and restoring transparency However, to print scaffolds for repairing hard tissues, although it is possible to increase the concentration of GelMA, a high concentration of GelMA can lead to an increase in the viscosity of the material, and even to clogging, making it impossible to print. To solve this problem, Shen et al. [33] printed scaffolds for cartilage repair by mixing GelMA and glycidyl methacrylate silk fibroin (SG) as a bio-ink (Fig. 2). GelMA hydrogels are limited in cartilage printing due to their high sensitivity to matrix metalloproteinases (MMP) degradation sites, resulting in fragility and rapid degradation [28]. In comparison, GelMA/SG has better mechanical strength and cell survival, as well as greatly improved shape retention and bio-printability (Fig. 3). Overall, GelMA is widely used in tissue bioprinting, especially soft tissues, due to its photosensitivity as well as low immunogenicity and high cell adhesion (it's arginine-glycine-aspartate (RGD) sequence). However, its weaknesses make it less popular for use in hard tissues such as bone and cartilage. Therefore, it is recommended that GelMA be printed with other polymers such as acrylamide, polycaprolactone, and silk fibroin nanofibers (SF) to improve its stiffness and printability and to broaden its application range [34].

2.1.2. Hyaluronic acid

Hyaluronic acid (HA), a polysaccharide molecule naturally occurring in human tissues, has been widely used in tissue engineering as one of the main components of ECM due to its excellent biocompatibility and ability to promote cell migration and repair [35]. However, pure hyaluronic acid hydrogels lack printability and suitable mechanical properties, thus limiting their application for 3D bioprinting. Therefore, Hyaluronic acid methacryloyl (HAMA), derived from methacryloylation of hyaluronic acid was formulated. The main synthesis procedure is to completely dissolve a certain amount of HA in ultrapure water, followed by a slow dropwise addition of methacrylic anhydride (MA) solution, and to maintain the pH value at about 8.5 by adding sodium hydroxide. After 8 h at room temperature, the reaction was dialyzed with distilled water in a dialysis bag (12,000–14,000 Da) for 3–5 days. The resulting solution was freeze-dried to obtain spongy HAMA [36]. It can be promote gelation with lithium phenyl-2,4,6-trimethylbenzoylphosphinate (LAP) under UV irradiation, and the produced hydrogels are not only stable but can also be used to construct biological organisms [37]. Si et al. [38] produced a double-crosslinked hyaluronic acid-based wound dressing by UV cross-linking with HAMA and cross-linking with 3, 3′-dithiobis(propionylhydrazine) (DTP) modified (HA-SH) by click reaction. The results showed that the dressing had a high dissolution rate and a high controlled degradation rate as the percentage of HAMA increased. In addition, both live-dead staining and CCK8 showed good biocompatibility. In addition to wound dressing, HAMA hydrogel can be used as bio-ink for organoid printing. Wang et al. [39] developed a 3D bioprinting islet organoid model using pancreatic extracellular matrixs (pECM) and HAMA as bio-inks. The results showed that the organoid promoted the attachment and growth of new blood vessels, had a low immune-inflammatory response after transplantation, as well as maintained blood glucose in the normal range in mice. The researchers concluded that HAMA in the bio-ink played a crucial role in improving cell survival and reducing immune responses. There are also many applications for HAMA in the repair of other tissues. Although HAMA already has good mechanical stiffness and long-term stability, like

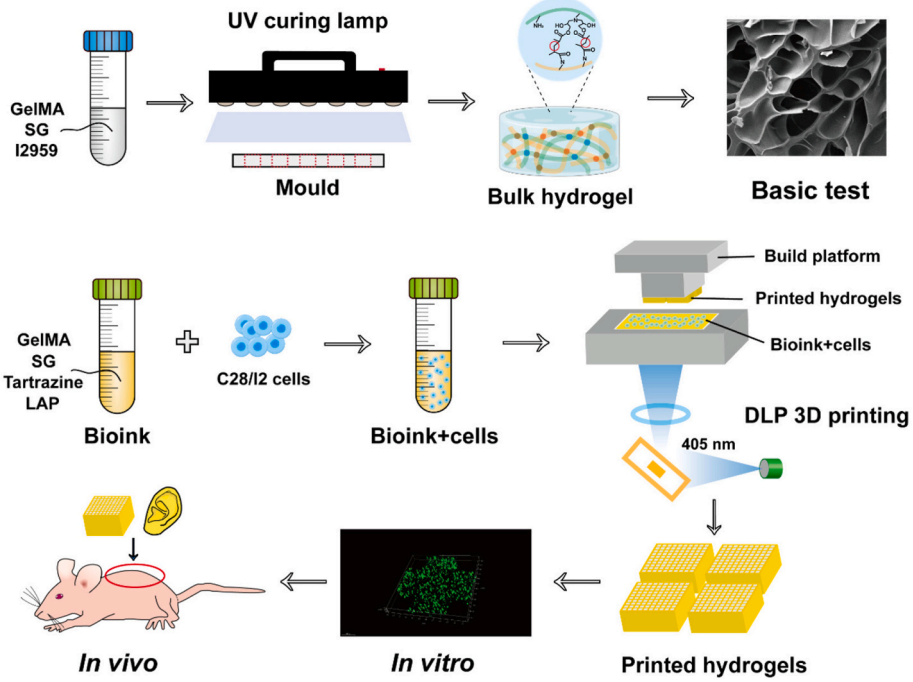

Fig. 2. Schematic diagram of the preparation of GelMA/SG bio-ink and its application to DLP bioprinting. Reproduced with permission [33].

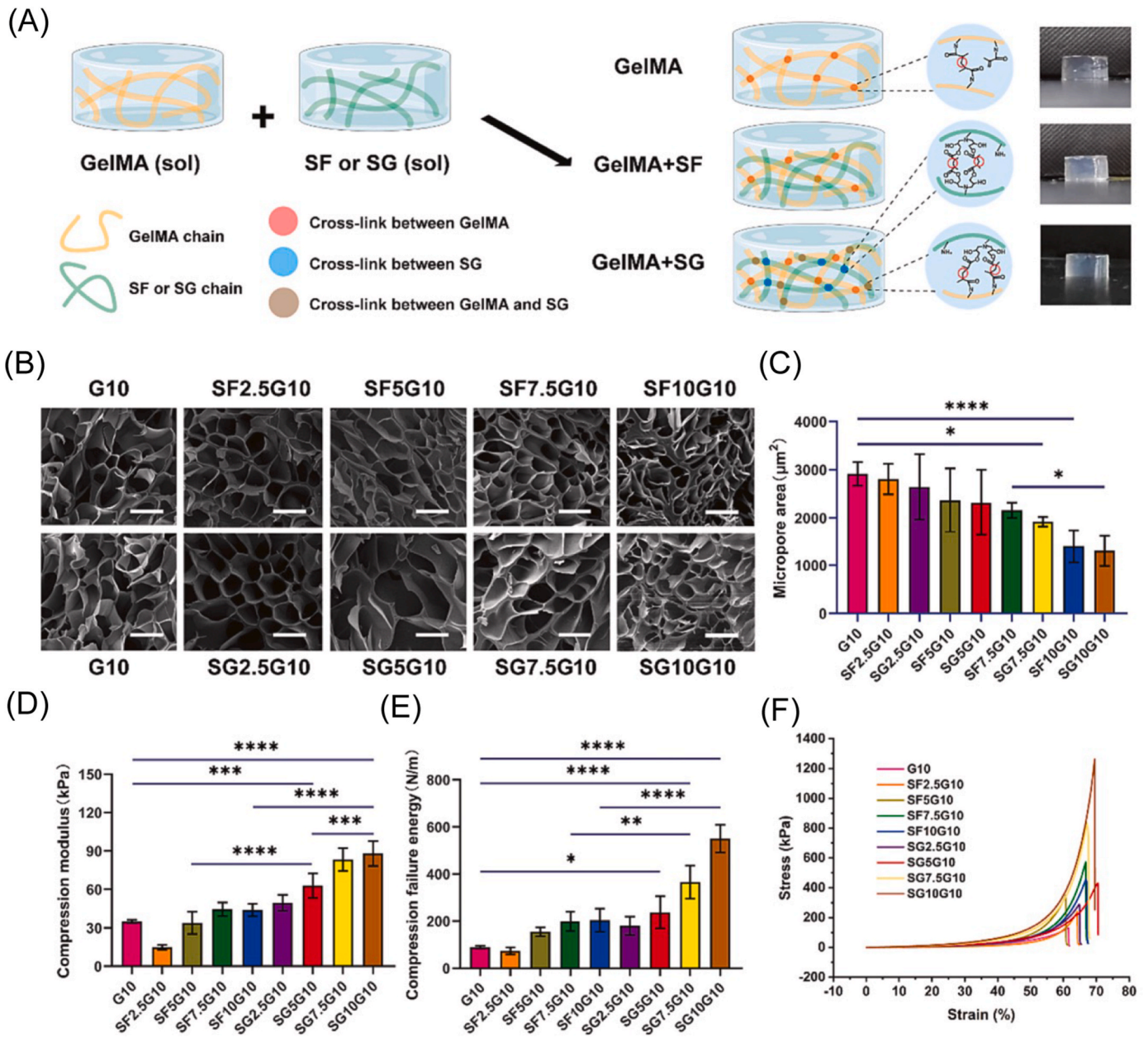

Fig. 3. GelMA/SG exhibits enhanced mechanical strength, cell survival, shape retention, and bioprintability. This section details the preparation and analysis of interpenetrating network (IPN) hydrogels. (a) Preparation schematic and internal network of GelMA, GelMA/SF, and GelMA/SG hydrogels. (b) SEM images reveal hydrogel microstructure (scale bar = 100 μm). (c) Pore size statistics. (d) Compressive modulus, (e) failure energy, and (f) stress–strain curves are presented. Data show mean ± SD (*p < 0.05, **p < 0.005, ***p < 0.001, ****p < 0.0001). Reproduced with permission [33].

GelMA, it may be difficult to apply it alone for bone tissue repair [40]. Therefore, researchers often choose to mix HAMA with other biomaterials for 3D bioprinting. Liu et al. [41] customized a biomimetic multiphase composite bio-scaffold based on polycaprolactone (PCL) and HAMA. In addition, they incorporated bone marrow-derived mesenchymal stem cells (BMSCs) and diclofenac sodium (DC). The former mainly plays a role in secreting biologically active substances to promote repair, while the latter mainly plays an anti-inflammatory role. The PCL chosen by the researchers is a commonly used biomaterial in bone repair, with suitable mechanical properties, which can greatly improve the printing ability and mechanical properties of HAMA hydrogel [42]. The combination of the two can better mimic the structure of bone tissue, stimulate the regeneration of osteochondral bone, and realize the recovery of joint function.

2.1.3. *Recombinant human collagen*

Collagen, a natural biomaterial commonly used in tissue engineering, is a key structural protein of the ECM that supports the morphology and integrity of native tissues. However, most of the commercially available collagen is derived from animal sources and usually needs to be preserved in an acidic environment, which is not conducive to 3D bioprinting as well as maintaining cell viability [43]. Additionally, collagen from animal sources also often has a certain degree of immunogenicity and potential for viral transmission [44]. When using these collagens for tissue engineering studies or for the preparation of animal serum, rigorous handling and processing can reduce or even remove immunogenicity and retain the richness of nutrients. In recent years, researchers have chosen recombinant human collagen (RHC) as an alternative to animal-sourced collagen. Although it is synthesized through genetic engineering techniques, it is classified as a natural

polymer because its structure and properties are very similar to those of collagen in nature. Recombinant human collagen is more purified, is fermented from microorganisms or genetically modified crops, and has a high degree of reproducibility, as well as low immunogenicity [45]. To enhance the printability of collagen, researchers chose to modify RHC with methacrylic anhydride to obtain RHC methacryloyl (RHCMA). Yang et al. [46] mixed RHCMA with acidified chitosan (CS) as a bio-ink and printed to form a CS-RHCMA composite scaffold. By varying the proportion of CS, the mechanical properties of the scaffolds could be greatly adjusted and the printability of the bio-ink could be improved. In addition, RHC has good solubility due to its well-defined molecular structure, which helps to maintain cell viability in 3D bioprinting. In vitro experiments confirmed that human umbilical vein endothelial cells (HUVECs) can still survive in the scaffold for some time after being printed. Although the polymer network formed by RHCMA is relatively inhomogeneous and has insufficient mechanical properties, on the whole, RHCMA provides new ideas for the application of collagen in tissue engineering.

In addition to the above-mentioned Natural photo-crosslinkable polymers, many other natural materials can be modified for a wider range of biomedical applications, such as Methacrylated chitosan, Methacrylated alginate (MAALG), and Methacrylated silk fibroin et al. [20,47,48]. Most of them have superior cell responsiveness, cell adhesion, and non-toxic degradation in vivo, but their shortcomings are obvious when applied to 3D bioprinting, such as high degradation rate and poor mechanical properties. Therefore, many researchers use synthetic photosensitive materials as suitable alternatives.

2.2. Syntheticpolymers

2.2.1. Polyethylene glycol

Polyethylene glycol (PEG) is a synthetic hydrophilic polymer that has been widely used in biomedical applications because of its excellent biocompatibility and low immunogenicity [49]. Xin et al. [50] synthesized PEG hydrogels via thiol-ene click chemistry, which feature rapid reaction kinetics, high cell compatibility, photo-initiated polymerization, tunable crosslink density, and easy incorporation of bioactive molecules. Compared to conventional nanoporous hydrogels, these hydrogels form interconnected microporous structures, providing a more permissive three-dimensional environment for cells, thereby enhancing cell spreading, migration, and functional performance [51]. In particular, PEG can be chemically modified to achieve a wide range of functions, with its terminal functional groups and its highly controllable molecular weight adding to the diversity of materials [52]. For example, Polyethylene glycol diacrylate (PEGDA) was synthesized by introducing photosensitive groups on the PEG chain. Inevitably, however, the synthesized polymers resulted in less bio-functional materials due to the lack of bioactive substances. To solve this problem, Zhu et al. [53] fabricated novel PEGDA/dCECM scaffolds using a 3D printer based on stereolithography with decellularized cartilage extracellular matrix (dCECM) and PEGDA as the bio-ink. The addition of dCECM, which has a large number of growth factors and abundant extracellular matrixs, greatly improves the biological function of the material and makes up for the shortcomings of PEGDA scaffolds. The good mechanical properties and high biocompatibility of PEGDA make it a new carrier for osteochondral defect repair as a skeleton.

2.2.2. Poly (vinyl alcohol)

Polyvinyl alcohol (PVA) is a hydrophilic synthetic polymer that has been widely used in bioscaffolds and gel formation because of its good biocompatibility and non-toxicity for many biological systems [54]. In addition, its large number of side hydroxyl groups provides attachment sites for many biomolecules as well as the possibility of modification. The rational utilization of this feature provides infinite possibilities for the modification of the chemical properties of PVA and its derivatives, such as for the introduction of methacryloyl groups into PVA molecules to obtain Methacrylated Poly (vinyl alcohol) (MaPVA). However, MaPVA hydrogels are poorly effective for cell growth and adhesion because of their surface resistance to protein adsorption, which limits their application in biomedical and tissue engineering fields [55]. Therefore, researchers often mix it with other bio-macromolecular materials to improve its cell adhesion as well as cell growth promotion. Chen et al. [56] developed maleilated hyaluronate (MHA)/MaPVA composite nanofibers and formed crosslinked nanofibrous network scaffolds by photocrosslinking polymerization. HA, as one of the main components of the extracellular matrix, promotes cell adhesion, migration, and proliferation, and the incorporation of MHA into this water-stable electrospun MHA/MaPVA nanofiber membrane gives it high water absorption, and at the same time, it can promote cell adhesion and growth, which greatly increases the potential of MaPVA hydrogel for biomedical applications.

In summary, synthetic photo-crosslinkable polymers are less biologically active than natural photo-crosslinkable polymers and are often unable to support cell adhesion and growth [57]. However, their chemical and mechanical properties are mostly reproducible, stable, and tunable due to controllable chemical and biological functional groups. In addition, customization of chemical properties can be achieved by introducing other functional groups, providing unlimited possibilities for their application in multi-tissue repair [58]. Therefore, researchers should make full use of the customizable chemical properties of synthetic polymers while focusing on improving their biological activities to achieve the function of promoting cell proliferation and growth.

2.3. Composite photo-crosslinkable polymers

2.3.1. Natural-natural polymers

In addition to the modification of photosensitizing groups on a single natural-source polymer, there are many researchers who have used multiple modified photosensitized natural polymers together as bio-inks for 3D bioprinting to provide customizable mechanical characterization and additional bioactive sites to maintain cell fates and functionalization [59,60]. Feng et al. [24] crosslinked hyaluronic acid modified with methacrylate and phenylboronic acid groups (HAMA-PBA) and GelMA, and subsequently assembled with a dynamic cross-linking agent (dopamine-modified hyaluronic acid, HA-DA) to prepare to present dynamic cross-linked microgel assembly (DC-MA) bio-inks (Fig. 4). While previous conventional microgel bio-inks mostly increased their viscosity by increasing the mechanical modulus of individual microgels [61], the bio-inks designed by the researchers increased their viscosity by establishing stronger interactions while still maintaining the relatively low mechanical modulus of the microgels, resulting in high printability/-shape fidelity and high cell viability. The resulting hydrogels can be prepared to print 3D structures with high shape fidelity and maintain cell viability without subsequent special treatment, and can also improve the microporosity, tissue adhesion, and self-healing of the printed structures, which opens up more possibilities for their application in tissue engineering (Fig. 5).

2.3.2. Synthetic-natural polymers

To obtain prosthetic materials with flexible design, adjustable physical and chemical properties, excellent biocompatibility, mechanical properties, and printability, the use of synthetic polymers together with natural polymers as inks for 3D bioprinting is also a good option. Zhou et al. [62] innovatively printed a small-diameter blood vessel with two layers of different cell types (vascular endothelial cells (VEC) and vascular smooth muscle cells (VSMC)) by utilizing an advanced 3D coaxial extrusion platform. They used GelMA, PEGDA, alginate, and alginate lyase as novel bio-inks. Among them, PEGDA was used to improve the printability and mechanical strength of the printed scaffolds [63]; alginate lyase was mainly used to degrade alginate to create enough space for the further growth of vascular cells and to promote cell growth and proliferation. Overall, the novel bionic blood vessels they fabricated

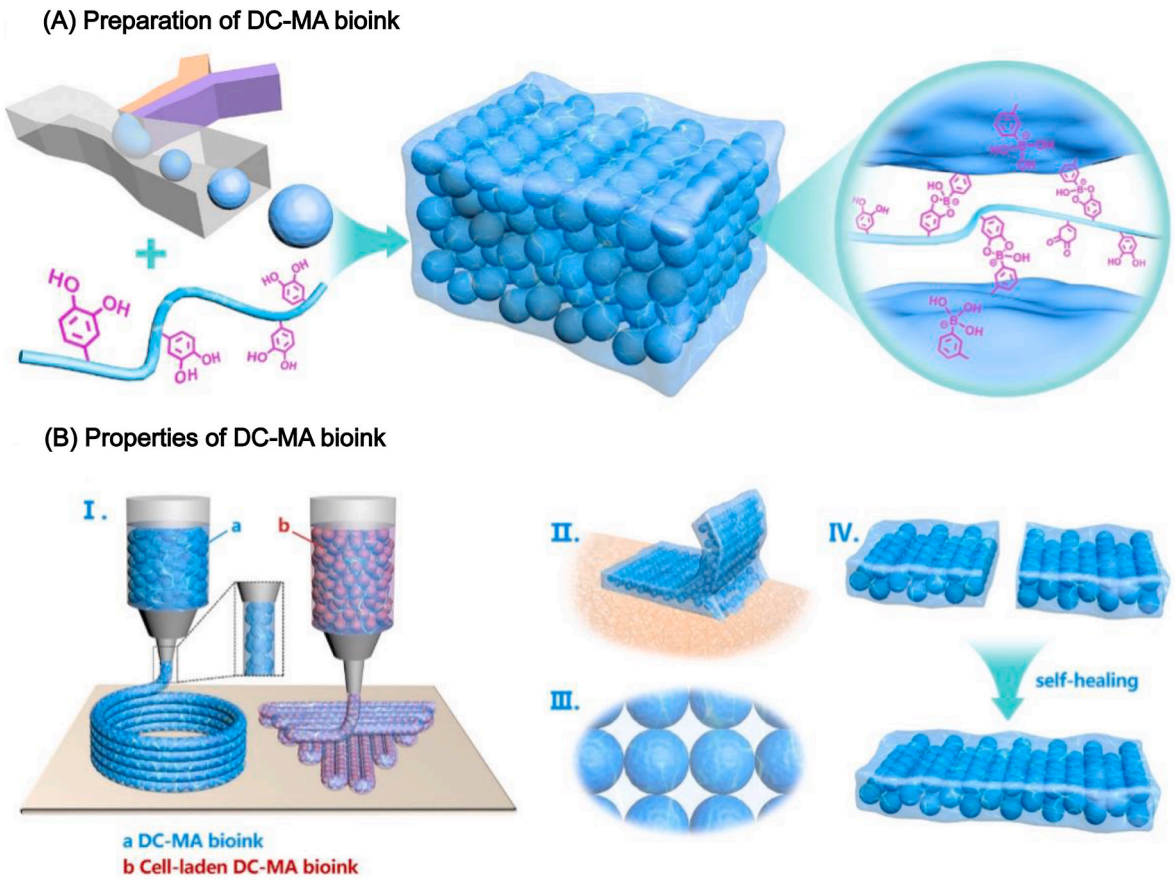

Fig. 4. Fabrication and characterization of DC-MA bio-inks. Reproduced with permission [24].

showed good perfusability and mechanical properties, with good cell growth on the scaffolds to form a distinct two-cell layer structure, as well as good angiogenic expression. This synthetic-natural polymer material provides a new idea for clinical small-diameter vascular replacement applications.

In addition to photo-crosslinked polymers alone, researchers have also blended photo-sensitive polymers with thermoresponsive polymers for dual-network crosslinking to print high-resolution scaffolds with good cell viability [64]. By chemically modifying alginate to introduce photosensitive groups, it gains the ability for photo-crosslinking while retaining its ionic gelation properties. Samorezov et al. [65] incorporated methacrylate groups and RGD peptides into alginate, enhancing it with photo-crosslinking ability and additional cell adhesion sites. The covalent crosslinking of methacrylate-modified alginate avoids the weakening of ionic crosslinking due to ion loss, thereby maintaining the hydrogel's mechanical and structural stability. Photo-crosslinking alginate enables the stereolithography of bioinks. Ooi et al. [66] utilized thiol-ene click chemistry to react norbornene-alginate with thiol-containing polymer crosslinkers. This approach provides enhanced spatio-temporal control over alginate's rheological and mechanical properties during bioprinting. Norbornene-functionalized alginate demonstrated good printability at a lower concentration (2 wt%) and resulted in a more stable 3D structure compared to solely relying on ionic crosslinking. Kesti et al. [67] blended thermoresponsive polymer poly(N-isopropylacrylamide) grafted hyaluronan (HA-pNIPAAM) with HAMA blend as bio-ink and synthesized layered cartilage constructs using layer-by-layer bioprinting. As a natural component of articular cartilage, hyaluronic acid (HA) has been widely used in cartilage tissue engineering [68]. Being one of the main components of the bio-ink, the thermoresponsive HA-pNIPAAM gels rapidly after deposition onto a heated substrate and maintains the structural fidelity immediately after printing [69], and the HAMA ensures long-term mechanical stability after photocrosslinking. This temperature-sensitive-photosensitive cartilage construct exhibits excellent rheological properties, swelling behavior, printability, and good cell survival after loading with cells.

Composite photo-crosslinkable polymers dramatically improve the mechanical properties and cytocompatibility of a single biopolymer, improving the mechanical properties of bio-inks as well as the chemical customizability of the material while retaining the lower antigenicity and good biocompatibility of natural polymers [70]. Although some of the synthesized chemical components are not well utilized and degraded by cells, in general, synthetic-natural hydrogels integrate the advantages of both polymers and are worthy of extensive future research as well as clinical translation.

3. Customization of parameters in the photocrosslinking process

To customize photo-crosslinkable 3D bioprinted scaffolds for repair according to the characteristics of different repair sites, in addition to choosing appropriate photo-crosslinkable polymers, adjusting the wavelength and exposure time of the light source and photoinitiators as well as material pore size during the 3D bioprinting process, etc., the properties of the final product can be tailored to meet the characteristics of the repair site [71,72]. Generally, more reactive photoinitiators and shorter polymer chains produce physically stronger photopolymers, while shorter exposure times and weaker light intensities result in lower crosslink densities [23,73]. Researchers can use these methodologies to create 3D bioprinted materials with specific physicochemical properties depending on the mechanical properties and cross-linking densities required for different repair areas (e.g., harder bone and cartilage tissues, softer adipose and vascular tissues, or wound dressings, etc.).

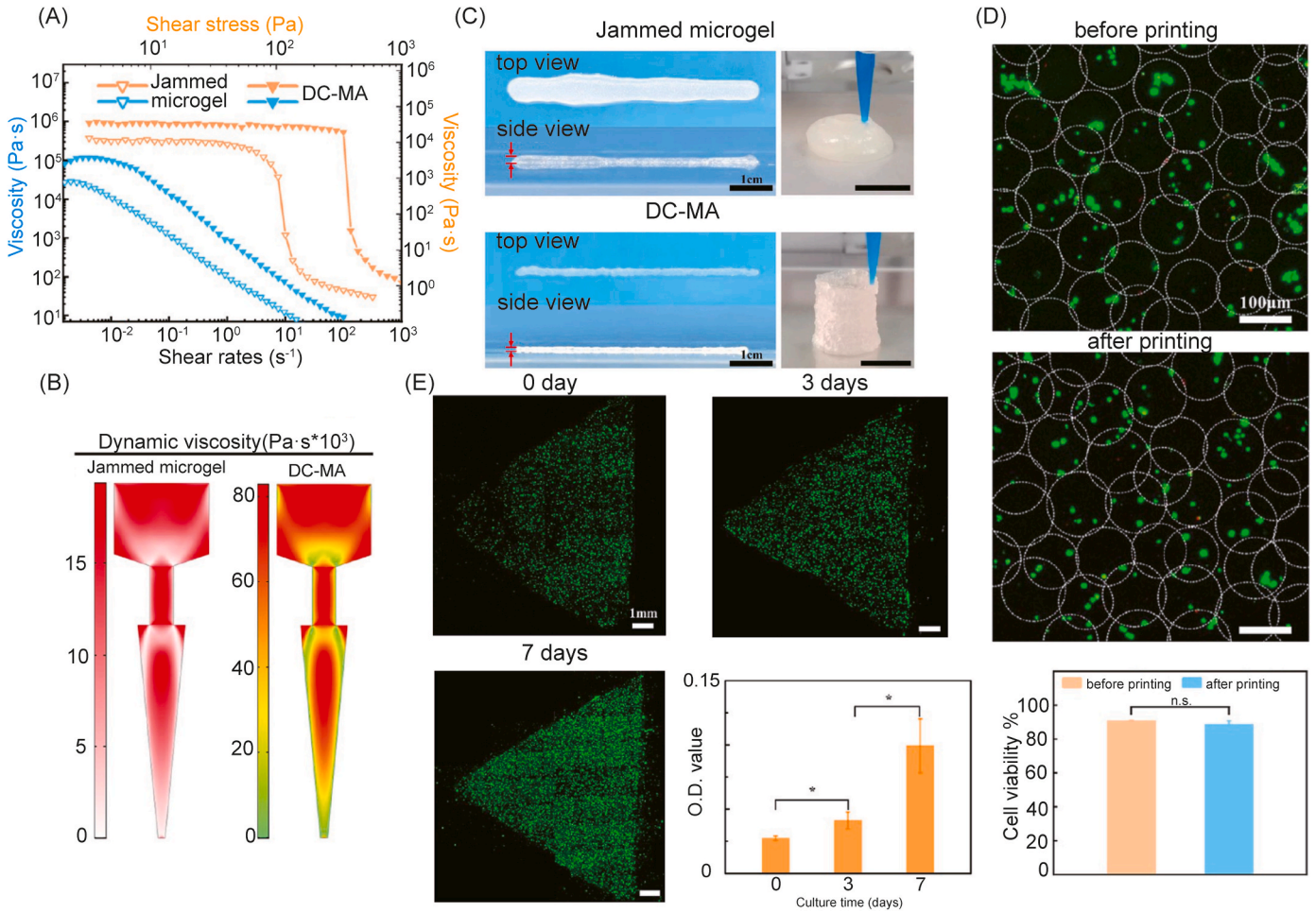

Fig. 5. The DC-MA bio-ink demonstrates enhanced viscosity, shape fidelity, and cell viability. (A) Rheological data reveal viscosity's dependence on shear rate and stress, indicating the ink's adaptability under different shear conditions. (B) Simulations show dynamic viscosity distribution in a tapered 600 μm nozzle, illustrating the ink's behavior during printing. (C) Comparisons of shape fidelity and extrusion stability highlight the DC-MA ink's superiority over jammed microgel inks. (D) Cell viability assessments of L929 cells, encapsulated in the ink before and after printing, confirm its biocompatibility. (E) Proliferation studies of these cells within the DC-MA ink post-printing underscore its supportive environment for cell growth. Statistical significance is marked by *P < 0.05, with n = 3. Reproduced with permission [24].

3.1. Control of light source

3.1.1. Wavelength of the light source

The light source is one of the most significant elements of the photocrosslinking process that can be controlled. The wavelength of the light source has a direct effect on the survival rate of the loaded cells in the hydrogel and the degree of crosslinking and other mechanical properties of the material [73,74]. Generally speaking, the wavelengths of ultraviolet light (UV-A (320–400 nm)) and visible light (400–700 nm) are used. UV-B (290–320 nm) is not recommended because cellular nucleic acids typically absorb ultraviolet light in this wavelength range in small amounts, leading to the synthesis of cyclobutane-pyrimidine dimers (CPD), which causes cellular damage. The shorter wavelengths and longer exposure times are likely to induce mutations [75–77], and also because the smaller depth of penetration of UV light can lead to uneven cross-linking in the hydrogel region, thus limiting the application of hydrogels. The reason for not using UV-C (200–290 nm) is that this wavelength of UV light is mainly used for sterilization, and prolonged use of this wavelength irradiation of the cells will cause the effects of cellular decline or even death [72,78]. Consequently, despite the minimal damage inflicted by short-term ultraviolet light exposure [79], researchers favor using visible light (400–700 nm) as the light source for its reduced cellular impact. The reason is that visible light is relatively biologically safe and has a deeper optical penetration depth than UV light, which allows the hydrogel to crosslink more uniformly and adequately, resulting in higher cell viability and cell survival after crosslinking [80]. Goto et al. [81] investigated and compared the mechanical characteristics and cell survival of photo-crosslinked GelMA at visible wavelengths (VW) and at UV wavelengths, and evaluated its feasibility as a scaffold for bone regeneration. They used Irgacure2959 and riboflavin as photoinitiators in making GelMA hydrogels, respectively. It was found that hydrogels with the same concentration of GelMA-Riboflavin (RF) irradiated with VW for 60 s and GelMA-Irgacure2959 (IR) irradiated with ultraviolet light for 20 s demonstrated showed similar mechanical strength, and the hardness was suitable for osteoblast differentiation. As shown in Fig. 6, The survival rate of KUSA-A1 cells encapsulated in GelMA-RF hydrogel polymerized with visible light was greater than that of cells encapsulated in GelMA-IR hydrogel polymerized with UV light. It should be noted that low concentrations of Irgacure do not significantly affect cell viability; cell death may result from the cross-linking process of the photosensitive polymers during long-term ultraviolet irradiation [79].

Furthermore, it was surprising that after they underwent osteoblast differentiation induction, osteoblasts encapsulated in GelMA-RF showed a significant increase in osteogenesis-related gene expression at the later stage of differentiation. This suggests that visible-light crosslinked GelMA hydrogels have better biosafety as well as biological properties than UV-crosslinked GelMA hydrogels. Therefore, we recommend the

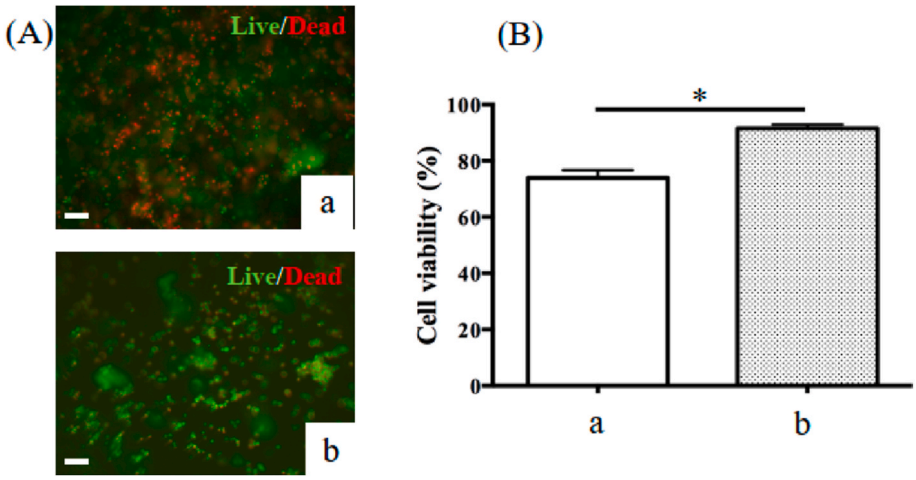

Fig. 6. Fluorescence pictures of KUSA-A1 cells stained green (live cells) and red (dead cells). (A) Viability assay of 20 % GelMA hydrogel-coated cells after exposure to (a) UV or (b) VW light and incubation in vitro for 1 day. (B) KUSA-A1 cells showed decreased cell viability when exposed to (a) UV light in contrast to (b) VW light; *p < 0.05. Scale bar: 100 μm. Reproduced with permission [81].

use of photoinitiators corresponding to visible light (400–700 nm), which ensure the biosafety while enhancing the other roles of the hydrogels, for a wider range of applications in the field of tissue engineering.

3.1.2. *Exposure time of the light source*

In addition to the wavelength of the light source, the exposure time also greatly affects the mechanical and other properties of the resulting scaffolds. In general, a longer exposure time ensures good mechanical properties of 3D printed scaffolds [82]. Kim et al. [83] synthesized chitosan-lactam (Ch-LA) hydrogel scaffolds for bone tissue engineering. They investigated the effects of exposure time at different UV light on the mechanical properties, crosslink density, and protein release kinetics of the scaffolds. It was found that the pore size of the scaffolds was smaller, the cross-linking density was higher, and the hardness of the hydrogels was higher as the exposure time increased. In addition, through the in vitro degradation experiments of the scaffolds, they also found that the longer the UV exposure time, the lower the wet weight remaining ratios and the slower the degradation rate. By examining the release kinetics of BSA from the scaffolds they found that increasing the UV exposure time reduced the initial burst release of BSA and prolonged the release duration (Fig. 7). Therefore, varying the exposure time of the light source can greatly adjust the mechanical, degradation, and biological properties of the scaffold to tailor the scaffold's functionality to fit and meet the requirements of the implantation site.

However, as mentioned above, prolonged exposure to light sources (especially UV light) also inevitably leads to a decrease in cell viability [72]. Stratesteffen et al. [84] found that cell viability decreased significantly with increasing UV exposure time. If the exposure duration to ultraviolet light exceeds 12 min, the one-week survival rate of hMSCs significantly decreases. They also concluded that the decrease in cell viability was not only related to the exposure time but also to the concentration of the photoinitiator-Irgacure. In recent years, with the depth of the study, researchers gradually found that in addition to adjusting the exposure time of the light source can determine the initial characteristics of photopolymers, light irradiation by post-curing can adjust the characteristics of shape memory materials as well [85]. Le Fer et al. [23] synthesized shape memory printed scaffolds using poly(propylene fumarate) (PPF) star polymers. They found that the glass transition temperatures and mechanical properties of the generated scaffolds increased with post-curing time, and their degradation rates showed the same trend. This finding allows shape memory printed scaffolds to adapt to large bone defects in complex situations by recovering the desired complex shape through dynamic response. More than that, the researchers found that longer light times could mean longer cross-linking times per layer, which could reduce overall printing speeds. This could have implications for printing efficiency and build preparation time. Consequently, choosing the appropriate exposure time can not only greatly improve the initial mechanical and other specific properties of the material, but also tailor the properties of post-curing, which is important for better control of the scaffold microshape.

Other light source parameters, such as optical density, light intensity, and irradiation distance, also play a crucial role in regulating the mechanical, rheological, and degradation properties, biosafety, and other special properties of the resulting 3D printed scaffolds. For example, with the increase of irradiation distance, the light penetration depth gradually decreases, and the hydrogel crosslinking may be inhomogeneous [86]. In conclusion, during light-based 3D bioprinting, subtle adjustment of different light source conditions, including wavelength, irradiation time, irradiation distance, and light intensity, etc., can better control the microgeometry and other physicochemical properties of the materials.

3.2. *Control of bio-ink systems*

3.2.1. *Molecular weight and concentration of polymers*

In addition to the above-mentioned changes in light source conditions that affect the properties of 3D bioprinted material scaffolds, the researchers also found that they can adjust the molecular weight and concentration of the polymers in the bio-inks, which in turn affects the physical properties of the resulting scaffold material [87,88]. In general, the higher the molecular weight of the polymer in the bio-ink, the stronger the physical strength of the material. This is due to the fact that polymer networks with higher molecular weights are less bendable. Nijst et al. [89] constructed elastomeric networks using poly(glycerol-*co*-sebacate) acrylate (PGSA). The molecular weight of PGSA was varied while changing its degree of acrylation (DA). The results showed that the Young's modulus, ultimate strength, and elongation at the break of the material were greatly enhanced with the increase in molecular weight. In addition, the change in the concentration of the polymer also leads to a change in the mechanical properties printing properties of the scaffolds. GelMA bio-inks incorporating calcium phosphates micro/-nanoparticles (CNP) were fabricated by Bhattacharyya et al. [90]. The growth and morphology were also controlled by varying the concentration of GelMA. It was found that as the concentration of GelMA increased, the nanoparticle shape gradually transformed into a spherical shape with a significant size reduction, and the shape fidelity and structural stability of the printed scaffolds were also improved.

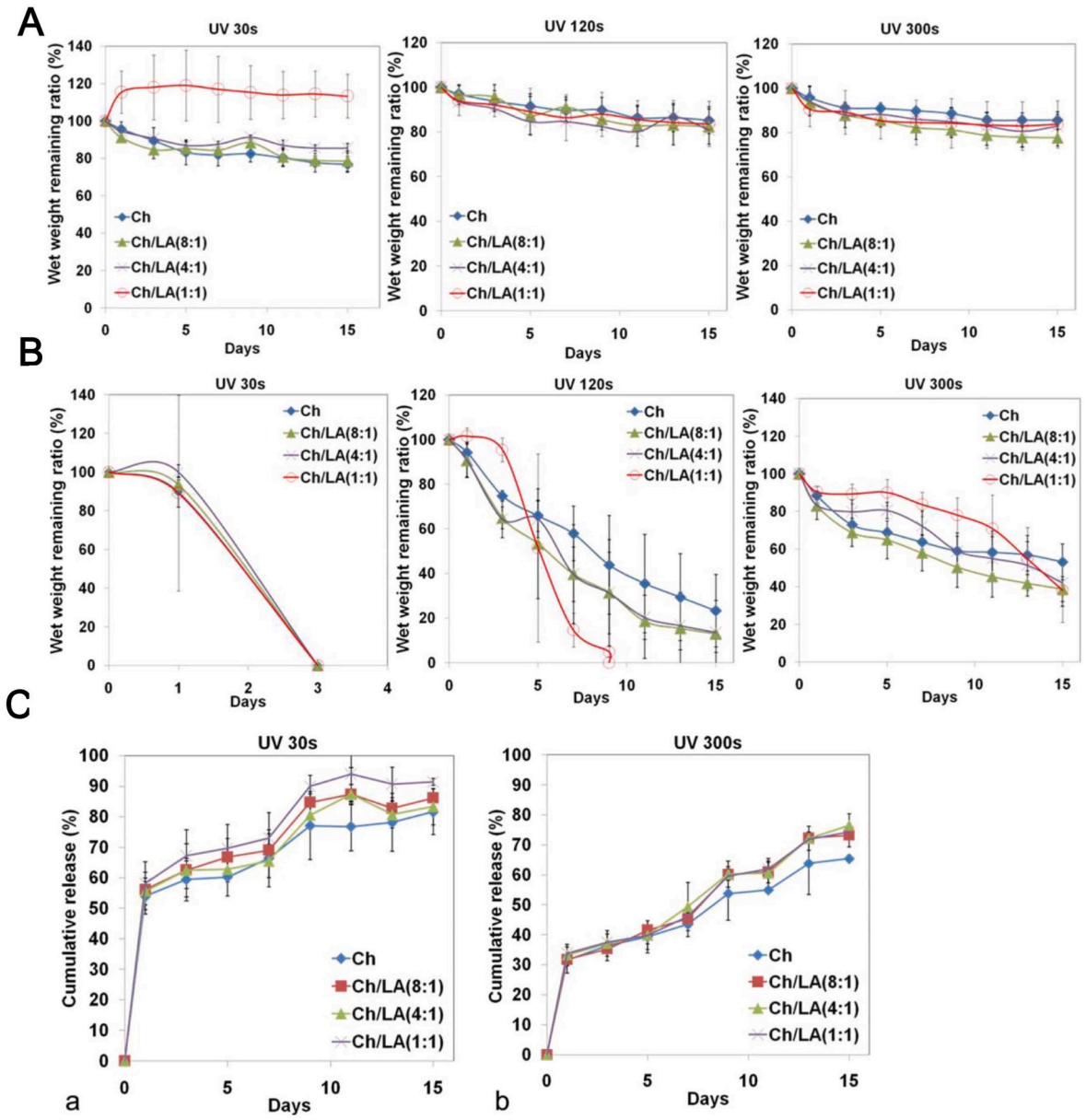

Fig. 7. In vitro breakdown of hydrogels in (A) PBS (pH 7.4) and (B) PBS (pH 7.4) with lysozyme (100 mg/mL) at 37 °C over 15 days. The degradation pattern was assessed by measuring the ratio of remaining wet weight of the hydrogels at each interval. Each value represents the mean ± standard deviation (n = 4). C. In vitro cumulative release profiles of BSA from Ch-LA hydrogels over 15 days. BSA was loaded into the prepolymer solution and exposed to 6.9 mW/cm^2 UV light for (a) 30 s and (b) 300 s. Release was measured by BCA assay at 560 nm. Each value represents the mean ± SD (n = 3). Reproduced with permission [83].

However, higher molecular weights and concentrations of polymers are not always better. Excessive increases in molecular weight or concentration can lead to increased viscosity of the bio-ink, which reduces its fluidity, decreases printing efficiency, and even leads to clogging of the nozzle. In addition, increased viscosity can lead to increased shear stress, inhibiting cell survival and cell behavior [17]. Appropriately reducing the molecular weight or concentration allows the bio-ink to pass quickly through the nozzle and deposit at the designated print area. In addition, lowering the concentration of the polymer also improves the pore size of the material, thus providing enough space to support cell growth. Bandyopadhyay et al. [91] mixed different concentrations of silk methacrylate (SilMA) and a fixed concentration of PEGDA as a bio-ink and mixed it with chondrocytes. The results showed that by adjusting the concentration of SilMA, the researchers could customize the material that could be adapted to cartilage regeneration with a suitable porous structure inside, excellent rheological properties, printability, and good degradation characteristics. Therefore, it is necessary to select the appropriate polymer concentration and molecular weight to develop new bio-inks with low viscosity but high performance.

3.2.2. *Photoinitiators*

Photopolymerization of light-based biomaterials is a process in which monomers/oligomers are exposed to a light source (UV or visible light) and cured by light irradiation to form a fixed material [92]. There are typically two types of photopolymerization: polymerization without the addition of a photoinitiator and polymerization caused by the stimulation of the photoinitiator by a light source. Photoinitiator-free polymerization can be initiated directly by UV or visible light. Farkas et al. [93] achieved photoinitiator-free polymerization of PEGDA material scaffolds using a novel mask projected excimer laser stereolithography (MPExSL) technique. They used UV light at wavelengths of 248 nm or 308 nm for irradiation. Sakai et al. [80] fabricated 3D

constructs loaded with human adipose stem cells (hADSCs) by using HA and gelatin derivatives as bio-inks. The scaffolds were formed by polymerization of HA with phenolic hydroxyl groups and gelatin derivatives (labeled with phenolic residues) under Ru (II)/sodium ammonium persulfate (SPS)(Fig. 8). They used a visible light-initiated crosslinking system and avoided the addition of photoinitiators. By evaluating the hADSCs in the material, it turned out that the scaffolds promoted their proliferation while maintaining their differentiation potential.

In addition, researchers have found that avoiding the use of photoinitiators can lead to scaffolds that exhibit better biocompatibility [94]. However, the use of photoinitiators for light-based 3D bioprinting applications is still the dominant cross-linking mode at present due to limitations such as material or technology.

Photoinitiators play a crucial role in the process of photo-crosslinking, because photoinitiator molecules, after directly or indirectly absorbing light energy, often jump from the ground state to the excited single-linear state, through the inter-systems scampering to the excited triple-linear state; in the excited single-linear state or the triple-linear state undergoes a chemical action, free radicals will be generated, cations, etc., which will trigger the polymerization of monomers to cross-link the cured compounds [95]. Moreover, one photoinitiator cannot be applied to all polymers, and researchers must select specific photoinitiators according to the bio-ink polymers in order to achieve the desired photocrosslinking reaction. Depending on the wavelength of the light source, the more commonly used photoinitiators can be broadly categorized as UV-sensitive (e.g., Irgacure 2959 and LAP) and visible light-sensitive (e.g., Ivocerin and camphorquinone). Among the UV-sensitive photoinitiators, Irgacure 2959 has a certain degree of water solubility. In addition, it has a high biosafety profile and is widely used for crosslinking of biomaterials such as GelMA and PEGDA [96]. Compared with Irgacure 2959, LAP has higher water solubility and photon absorption. Therefore, the material scaffolds formed by using LAP as a photoinitiator usually have better mechanical strength and cell growth properties, slower degradation rate, and smaller pore size [79].

Recently, the use of visible light-sensitive photoinitiators to initiate the photopolymerization of bio-inks has gradually gained more and more attention from researchers because it can circumvent UV irritation, improve biosafety as well as optical penetration, and maintain high cell viability in the constructs [97,98]. Double-crosslinked tyramine-modified methylcellulose(MC-Tyr) conjugates were prepared by a two-step method by Shin et al. [99]. The dual-crosslinked methylcellulose-tyramine (MC-Tyr) hydrogel demonstrates significant performance enhancements. Its maximum compressive strength is 16.62 kPa, compared to only 4.01 kPa for the physically crosslinked sample, with a notable increase in storage modulus. Regarding swelling properties, the dual-crosslinked sample shows a swelling ratio below 200 % within 72 h, whereas the physically crosslinked sample reaches 600 %. Printability and structural stability tests indicate that the dual-crosslinked hydrogel remains stable for 30 min post-printing and retains its initial structure even after 60 days in PBS. Cell viability tests reveal that the dual-crosslinked hydrogel maintains over 90 % cell survival 72 h post-printing, demonstrating excellent biocompatibility and cell proliferation capabilities, making it suitable for tissue engineering and 3D bioprinting applications. The so-called double cross-linking is the formation of reversible physical cross-linking (temperature cross-linking) and irreversible chemical cross-linking (photocrosslinking) by changing the temperature and or visible light. In this case, photocrosslinking is promoted by irradiation of visible light to promote crosslinking of the MC-Tyr solution using a highly biocompatible photoinitiator (RF and riboflavin 5′-monophophate, RFp) [100]. The 3D bioprinted constructs exhibited excellent mechanical properties and printability, superior cell survival and good biocompatibility.

Therefore, in conclusion, by choosing the appropriate photoinitiator system (photoinitiator or adding photoactive groups to the polymer), researchers can not only customize the construct mechanical strength and physical and chemical properties, etc., but also prevent free radical-induced cytotoxicity and improve biosafety.

3.3. Adjustment of the spatial structure and physical characteristics of the scaffolds

The application of photo-crosslinked 3D printed scaffolds in tissue

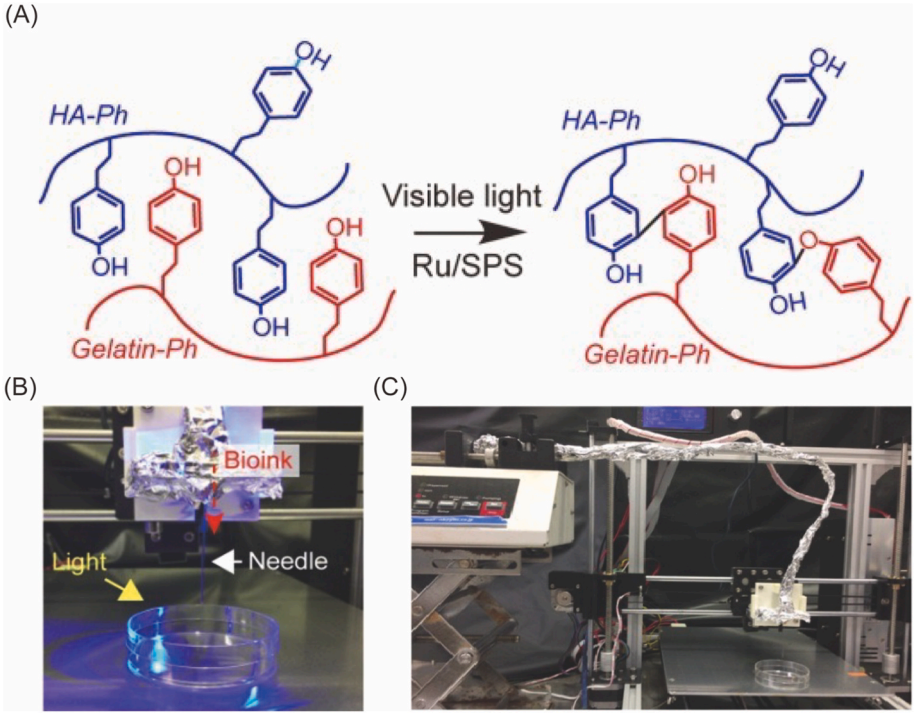

Fig. 8. (A) Schematic of the cross-linking system initiated by visible light using the Ru/SPS system; (B) Micro-extrusion process under visible light irradiation; (C) Overall view of the printing system. Reproduced with permission [80].

engineering is influenced by their structural and physical properties. For example, the morphology of the scaffold surface influences cell attachment and proliferation. The roughness and microstructure of the surface can regulate cell adhesion and differentiation, thus affecting tissue regeneration. Mechanical properties and mechanical strength of scaffolds, such as modulus of elasticity and tensile strength, are critical for supporting growing tissues and load bearing [101,102]. In addition, in recent years, researchers have come to realize that appropriate structural accuracy and resolution are critical for simulating complex tissue structures, such as bone and soft tissue, and for providing more accurate scaffold shapes. Therefore, they often choose Multi-Cross-Linking bio--ink for 3D bioprinting. Hao et al. [103] adopted a step-by-step Multi--Cross-Linking strategy using Pluronic F127 and HAMA as bio-inks (Fig. 9).

The bio-inks they designed underwent pre-crosslinking at low temperatures (4–20 °C), followed by self-assembly at body temperature (37 °C), and final photo-crosslinking. The results showed that the unique step-by-step cross-linking mechanism they designed ensured that the bio-inks printed complex structures with very high shape fidelity at different printing stages, all exhibiting a uniform and interconnected porous structure, which is similar to the natural extracellular matrix and maintains the bioactivity of the implanted cells.

The study of the pores of scaffolds has received much attention from researchers in recent years [104]. Researchers have found that the pore structure and porosity of printed scaffolds will directly affect cell spreading, nutrient and waste discharge. Suitable porosity can provide space for cell growth and contribute to the formation of newborn tissues [105,106]. Therefore, researchers often customize the pore space of materials through various ways. To create a three-dimensional porous gradient structure, Wang et al. [107] mixed two GelMA bio-inks with the same concentration but containing different substances. The two GelMA bio-inks contained human mesenchymal stem cells(hMSCs)and different concentrations of porogen, respectively. By varying the concentration of porogen (0.5 wt%, 1.5 wt%, and 3 wt%), three gradient zones with different pore diameters could be observed from an optical microscope, with average pore diameters ranging from 12 μm, 29 μm, and 65 μm, respectively. Compared with the bio-ink mixed with 0.5 wt% porogen, the size of cell clusters increased 2.5-fold and 4-fold in the hydrogel regions containing 1.5 wt% and 3 wt% porogen, respectively. These results all indicate that the presence of porogen promoted the formation of pores in the bioprinted hydrogel constructs, and the higher the concentration, the larger the pores formed. The large pores facilitated the diffusion of nutrients and oxygen between cells and the removal of waste products, thus providing an environment that promotes cell spreading and proliferation.

In conclusion, by tuning the photo-crosslinked polymers and using different additives, final printed constructs with complex structures with gradient porosity, density and high shape fidelity can be obtained, providing ideas for the customization of photo-crosslinked hydrogels.

4. Applications of light-based 3D bioprinting

4.1. Bone tissue engineering

Bone tissue regeneration has been one of several clinical challenges due to the requirement of high specialized structural and mechanical properties [108]. In recent years, bone tissue repair and regeneration using tissue engineering techniques has received much attention from researchers [109]. Among them, 3D bioprinting is more commonly used because it can directly print scaffolds with tissue-specific properties, such as large pores, high mechanical strength, and high pore-interaction capacity, according to the specific structure and properties of the patient's graft site [110,111]. Reconstruction of large bone defects is challenging due to the lack of adequate blood supply and nutrients such as growth factors. 3D bioprinting allows for better bone regeneration by incorporating specific types of cells and growth factors during the printing process to mimic the hierarchical structure and physiological functions of natural bone [112]. Rajput et al. [20] proposed a light-cured methacrylated silk fibroin (SF-MA), and the physical characterization showed that the hydrogels (10–25 wt%) exhibited viscoelastic behavior similar to bone tissue and good degradability as well as complex structures with high precision. In addition, they have encapsulated preosteoblasts within the hydrogel, where the DLP bioprinted hydrogel with 15 % SF-MA was able to support the strongest cell proliferation with good cell morphology and cytoskeletal organization. Cell-mediated calcium deposition gradually increased over 14 days, substantiating the gel's ability to drive osteogenesis. As shown in Fig. 10, for the purpose of repairing bone tissue, Byambaa et al. [113] used GelMA hydrogel containing silicate nanoplatelets as a bio-ink for 3D bioprinting and incorporated vascular endothelial growth factors with graded concentrations to promote vascular proliferation while inducing bone regeneration. In addition, in order to form perfusable blood vessels inside the above printed constructs, they also printed cylinders in the center of the constructs using 5 % gelatin hydrogel with low methacryloyl substitution as bio-ink. The results showed that the construct exhibited good structural stability and biocompatibility. In vitro

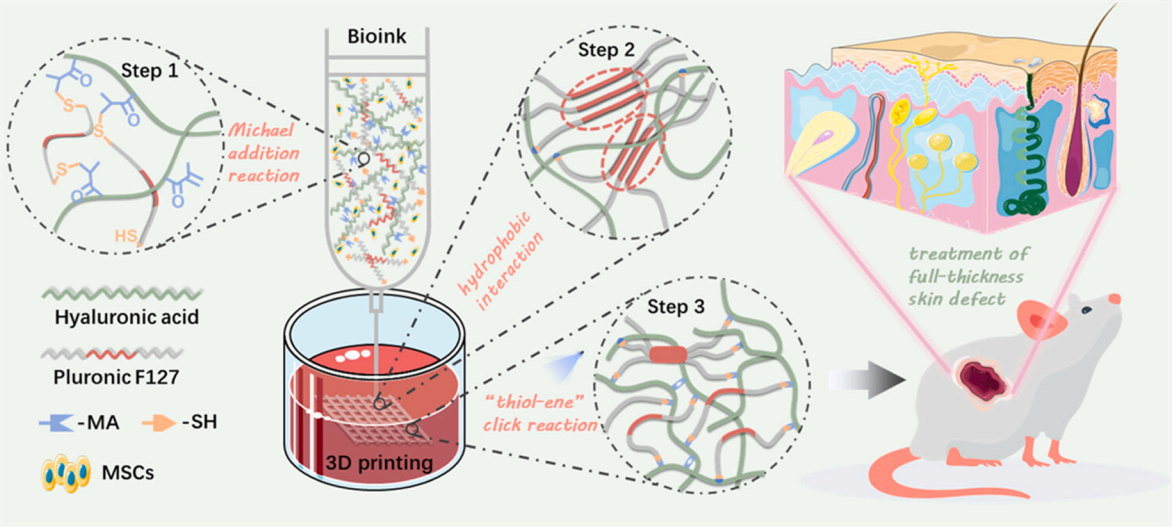

Fig. 9. Schematic representation of the development of a stepwise multi-crosslinked bio-ink for 3D bioprinting, and its therapeutic illustration. Reproduced with permission [103].

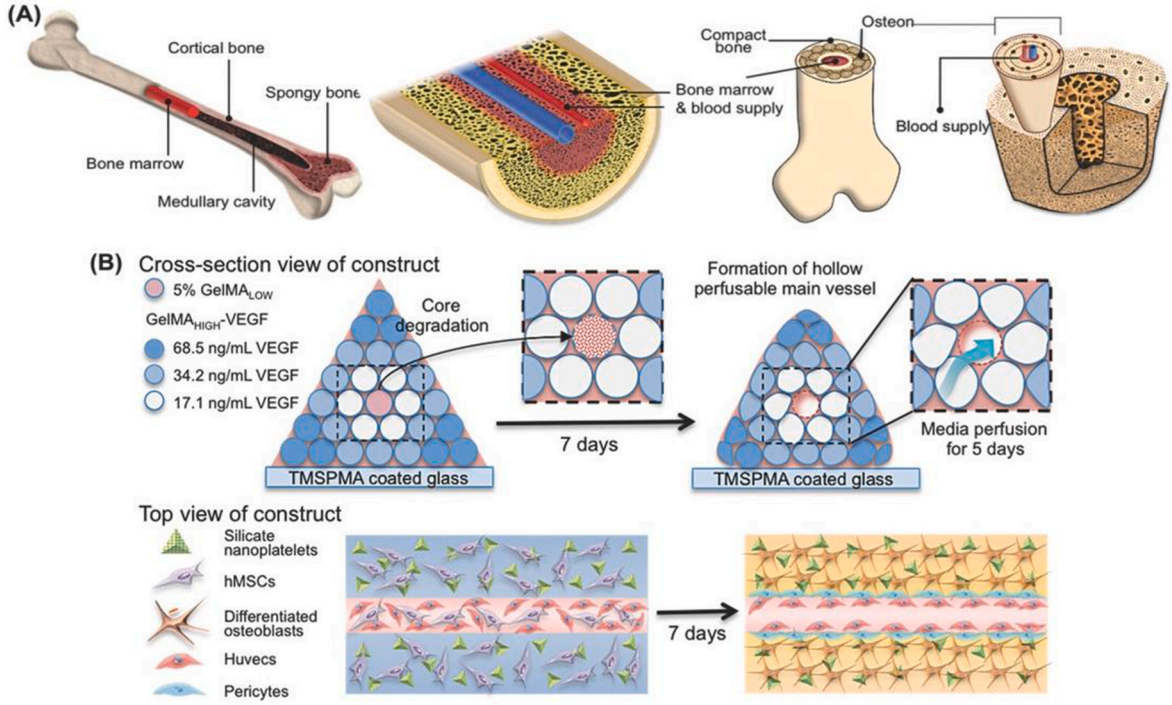

Fig. 10. Produced 3D structures mimicking bone. (A) Schematic representation of human bone tissue structure. (B) Illustration of produced bioprinting strategies for repairing bone defects. Reproduced with permission [113].

experiments also demonstrated that the scaffold could promote endothelial cell value-addition and induce osteogenic and myogenic differentiation of stem cells.

The downside, however, is that they did not validate the functionality of the construct by transplanting it into an animal, if the construct also exhibits excellent regenerative properties in vivo, the bio-ink would provide a valuable idea for engineered bone construction for the treatment of large bone defects. Unavoidably, however, although good progress has been made in 3D bioprinting for repairing bone tissue defects, because of the complexity of the bone tissue structure and the limitations of the existing printing technologies and materials, it is difficult for constructs to accurately simulate the multiscale layered structure of natural bone tissue as well as to realize the gradient printing of the mechanical properties [114], thus there are still many problems yet to be solved in the repair of bone tissues.

4.2. Cartilage tissue engineering

The body's natural cartilage tissue is a type of smooth and elastic tissue that is found in a wide range of organs. However, due to the lack of blood vessels, nerves, and lymphatic vessels, it is less capable of self-repair and even more difficult to regenerate than bone tissue [115,116]. Therefore, the need to repair cartilage tissue using tissue engineering techniques has become more urgent. Photosensitive polymers for cartilage tissue engineering can be printed while varying the photocrosslinking conditions according to different implantation sites, thus customizing the mechanical characteristics of the constructs [67,117,118]. In addition, bioactive components such as cytokines and cells can be added to the bio-ink and distributed to specific sites through precise localization to increase the ability to repair [119,120].

Glycosaminoglycans (GAGs) are one of the main components of the extracellular matrix and play an irreplaceable role in cell proliferation and regeneration. And the two most abundant GAGs in cartilage are HA and chondroitin sulfate (CS) [121,122]. Therefore Levett et al. [68] added these two GAGs individually or together into Gel-MA hydrogels (HA-MA and CS-MA) and mixed human chondrocytes into them. The results of both in vivo and in vitro experiments showed that the incorporation of a small amount of HA into the bio-ink significantly enhanced chondrogenesis, promoted matrix distribution, and improved the mechanical properties of the constructs. And the addition of CS could enhance certain redifferentiation properties of chondrocytes. They demonstrated that the incorporation of components from natural cartilage into bio-inks can significantly improve the function of the constructs, which provides an important idea for future research on cartilage regeneration or even other tissue regeneration. In recent years, other researchers have also noted the advantages of extracellular matrices for repairing cartilage regeneration [123]. Visscher et al. [124] processed chondro-derived dECM (cdECM) into photo-crosslinked hydrogel (cdECMMA) by methacrylation and mixed with chondrocytes to make bio-inks. The results of testing the mechanical properties of the constructs showed that cdECMMA exhibited excellent mechanical properties, structural integrity, and tissue stability. In addition, the chondrocytes in the construct maintained strong viability and proliferation, and secreted cartilage ECM components, including collagen and GAG. This cdECMMA-based bioprinted construct provides a great help in personalizing the therapeutic approach to ear cartilage reconstruction.

4.3. Vascular tissue engineering

A fully functional artificial organ reconstruction cannot be achieved without the construction of a vascularized network [125]. Although vascular tissues have some regenerative ability, the ability is relatively limited. And by printing blood vessels through tissue engineering technology, larger and more complex tissue structures can be constructed to facilitate the delivery of blood, nutrients and oxygen in the body and improve the survival and function of engineered tissues in the organism [126,127]. In addition, 3D bioprinting better mimics the structure and function of the natural vascular system, which has potentially important implications for improving the survival of transplanted organs and reducing the risk of rejection [128]. However, the development of perfusable blood vessels for efficient transplantation remains one of the

biggest current challenges. Currently, vascular tissues that can be constructed by 3D bioprinting fall into two main categories: tubular channels (1–6 mm in diameter) and microvascular networks. Jia et al. [129] used GelMA, sodium alginate and four-arm poly (ethylene glycol)-tetra-acrylate (PEGTA) as bio-inks for bioprinting with a multilayer coaxial extrusion system. The bio-ink was initially crosslinked with calcium ions via sodium alginate, followed by covalent photocrosslinking of GelMA and PEGTA, and finally PEGTA was used to adjust the biomechanical strength of the bio-ink and encapsulate endothelial and stem cells. The tubular perfusable vascular scaffold exhibited good biocompatibility and high histocompatibility. In addition to tubular perfusable vessels, microvascular systems can also be constructed with light-based 3D bioprinting systems. Moor et al. [130] utilized triculture HUVEC/HFF/ADSC spheroids in conjunction with GelMA hydrogel and Irgacure 2959 to construct a microvascular system. Upon implantation of this microvascular construct into the chick chorioallantoic membrane, the constructs showed features compatible with host blood vessels and exhibited strong angiogenic potential. Zhu et al. [131] used glycidal methacrylate-hyaluronic acid (GM-HA) and GelMA as a bio-ink for the construction of a microvascular system by utilizing DLP-based rapid bioprinting and microscale continuous optical bioprinting (μCOB) techniques to construct vascular structures. Both in vivo and in vitro experiments showed that endothelial cells could survive and proliferate in the constructs, spontaneously forming lumen-like structures and functional endothelial networks. Functional blood vessels containing anastomosing regeneration of erythrocytes were observed between the constructs and the host circulation. However, despite some initial successes in printing large tubular blood vessels and small microvascular networks with light-based 3D bioprinting, there is still a long way to go to achieve long-term perfusion of blood vessels ex vivo as well as mimicking vascular characteristics of certain complex tissues (e.g., alveoli and glomerulus) and successful clinical translation [132].

4.4. Skin tissue engineering

The skin is one of the largest organs of the human body and has a variety of important physiological and biological functions that are of great significance to the human body [133]. It can act as the body's first line of defense against external physical, chemical and biological stimuli. And after trauma, burns and chronic inflammatory ulcers, improper healing of the skin tissue may lead to infection and scar tissue formation, which not only fails to prevent the invasion of microorganisms and harmful substances, but also impacts aesthetics [134,135]. With 3D bioprinting, researchers can create tissues that resemble a patient's own skin for burn and trauma repair and to reduce the patient's immune rejection and promote wound healing. In addition, 3D bioprinting can be performed by adding the patient's own cells or growth factors, for example, to the bio-ink to achieve high bioactivity of the constructs [136]. Michael et al. [137] produced a skin substitute using laser-assisted bioprinting (LaBP) technology and placed fibroblasts and keratinocytes in the upper layers. After placing this skin substitute on a full-thickness skin wound in nude mice, the results showed that the grafts completely fused with the surrounding tissue, formed a thin stratified tissue in the epidermal region, and began to differentiate and form the stratum corneum. In addition, fibroblasts secreted large amounts of collagen, and all of these results demonstrated the formation of a new skin-like tissue.

Since the skin is the first barrier to contact with the outside world, there are many bacteria that colonize it [138]. Therefore, early wound management is essential to prevent wound infection and promote wound healing [139,140]. 3D bioprinted scaffolds with some antimicrobial properties can prevent and treat wound infection and enhance the success of construct grafting. Li et al. [141] synthesized methylene blue (MB)-loaded UiO-66(Ce) nanoparticles (NPs). And the synthesized NPs were compounded with photo-crosslinked silk fibroin (SF)/gelatin hydrogel to finally form MB@UiO-66(Ce)/photo-crosslinked hydrogel. The in vitro evaluation of the constructs revealed that the printability and mechanical properties of the bio-inks were greatly improved by the incorporation of NPs. In addition, the constructs exhibited good biocompatibility and could promote the migration and proliferation of fibroblasts. What's more, the hydrogels exhibited good antimicrobial effects, and the antimicrobial effects were more obvious with the increase of concentration. In vivo experiments also confirmed the constructs' function of promoting wound repair.

However, most of the current research focuses on adding growth factors, cells or other additives, etc. to the constructs to achieve their function of repair, and less research has been conducted on mimicking the multiple structures of natural skin tissues (e.g., blood vessels, nerves, and skin appendages, etc.) [142,143], which may be attributed to the thin structure of the skin, making it difficult to achieve multi-layer, multi-tissue printing in the constructs. Therefore, for the clinical translation of 3D light-based bioprinting, this will be one of the next challenges that researchers will work together to solve.

4.5. Cardiac tissue engineering

The significance of the cardiac tissue to the human body cannot be overstated; it is the central organ that sustains life and plays a decisive role in maintaining blood circulation and the supply of oxygen [144]. Heart disease and dysfunction can lead to a wide range of health problems, so keeping the heart healthy is key to maintaining overall physical health [145]. Heart tissue is poorly regenerated, with cells no longer increasing after birth in most mammals, and heart injuries are usually filled with scar tissue rather than new heart muscle cells. However, heart transplantation has a very limited application due to scarcity of donors and immune rejection, making it difficult to save most patients. Therefore, the use of tissue engineering techniques to repair the heart muscle using the patient's own cells or biocompatible materials not only reduces immune rejection, but also effectively solves the donor problem [146,147]. In addition, through the application of tissue engineering technology, researchers can also study the mechanisms as well as the structure of cardiac diseases in greater depth, which can help to understand cardiac development and disease processes. Ma et al. [148] developed an in vitro human cardiac tissue model using light-based 3D bioprinting technology by mixing cardiomyocytes from healthy wild-type (WT) and Long QT Syndrome Type 3 (LQT3)-labeled pluripotent stem cell-derived cardiomyocytes (iPS-CMs) into a UV-cured filamentous matrix and precisely regulating the structural arrangement of the CMs and adjusting the mechanical environment of the cells. Using this heart model, they found that these iPS-CMs beat arrhythmically compared to normal cardiomyocytes. They also studied the different responses of the diseased cardiomyocytes to different drugs, and preliminarily explored the phenomenon and causes of their contractile dysfunction. In addition to the complex structure, multicellular interactions greatly influence the function of the heart. To study the interactions between cardiomyocytes and fibroblasts in vivo, Kumar et al. [149] fabricated fibrin-gelatin bio-inks in which iPS-CMs and cardiac fibroblasts were mixed. The results showed that the printed constructs displayed a porous mesh structure with long-term in vitro stability. Cardiomyocytes showed excellent viability and proliferation in the constructs, and the expression of troponin I cardiac markers was increased. In addition, immunochemical results showed heterocellular coupling between the two cell types through Connexin43 adhesion junctions, which is extremely important for revealing cellular interactions, disease regulatory mechanisms, and the maintenance of normal physiological function of the human heart wall. In addition, the model can be used for cardiac drug cytotoxicity screening or revealing the causative factors of cardiac diseases in vitro. In addition to mimicking the complex structure of the heart as well as the cells, photo-crosslinked polymeric biomaterials for cardiac tissue printing should also have electrical conductivity to facilitate electrical coupling between neighboring cells. Zhu et al. [150] used GelMA and gold

nanorods (GNRs) as bio-inks for cardiac tissue scaffold printing. The results showed that the addition of GNR increased the adhesion of cardiac cells and improved cell-to-cell coupling, enabling synchronized contraction of this construct compared to the construct without GNR. In conclusion, although many researchers have revealed cardiac tissue ultrastructure as well as in vivo cellular morphology and responses (e.g., cell-cell or cell-matrix interactions) [151], how to integrate various cellular and complex cardiac structures to improve the applicability of 3D bioprinting remains a major challenge.

4.6. Skeletal muscle tissue engineering

Skeletal muscle constitutes approximately 45 % of human body weight, involving over 600 muscles responsible for support, stability, locomotion, and metabolic regulation [152]. Muscles are prone to trauma and myopathies, and in the United States, around 4.5 million reconstructive surgeries are performed annually, incurring healthcare costs in the billions of dollars [153]. Skeletal muscle tissue engineering (SMTE) aims to replace or restore functionality in muscles damaged by disease, injury, or surgery. To enhance the functionality of engineered muscles, researchers focus on mimicking the structure and microenvironment of in vivo skeletal muscle. A common strategy involves creating anisotropic scaffolds to align muscle cells and promote myogenesis. Despite various materials and methods used in SMTE to generate muscle tissue in vitro, the functionality of these constructs remains inferior to that of natural muscle.

Three-dimensional bioprinting allows precise deposition of matrix and cells to construct complex structures [154]. Gao and Cui demonstrated that bioprinting can deposit mouse myoblasts (C2C12) with high precision and viability, enhancing physiological responses [4]. Bashir et al. created bio-bots using stereolithography, driven by skeletal muscle cells to perform simple tasks [155]. To overcome the spontaneous contraction of skeletal muscle tissue on flexible substrates, researchers developed biohybrid robots powered by antagonistic muscle pairs, showing large-range movements and sustained activity [156]. Additionally, advanced constructs mimicking skeletal muscle tissue have been fabricated using additive manufacturing and other scaffold technologies. For example, Yeo and Kim used wet electrospinning to create aligned and random PCL microfiber bundles, observing higher organization and differentiation of cells on aligned fiber scaffolds [157]. Lee et al. bioprinted human muscle progenitor cells and implanted them in a rat model, restoring significant muscle strength and achieving good vascular and neural integration [158]. Kaplan et al. cultured human myoblasts in silk and hydrogel, observing functional neuromuscular junctions [159].

The potential of 3D bioprinting in skeletal muscle applications is substantial, including addressing organ transplant shortages, improving drug screening accuracy and safety, and advancing personalized medicine. However, several challenges must be addressed for clinical implementation. These include slow progress in soft material printing, the need for improved printing resolution and speed, high-throughput printing, and the bioprinting of thick tissues. Furthermore, identifying suitable cell sources and developing optimal bioinks are crucial for the advancement of this technology.

5. Conclusion and outlooks

In recent years, light-based 3D bioprinting has shown great potential in the field of tissue engineering and regeneration due to its excellent characteristics [130,160,161]. The main advantages are: 1. High resolution and precision: Light-based 3D bioprinting technology has been noticed for its high resolution and precision. This characteristic is mainly due to its use of photolithography. Photolithography is a method of patterning compounds using light-sensitive compounds in response to light. In bioprinting, this process enables high-precision localization and patterning of biomaterials. During photolithography, a photomask can be used to project light through the light-transmitting portion of the template, exposing the light-sensitive biomaterial to light. This permits the formation of high-resolution structures on a microscopic scale. In addition, light-based 3D bioprinting typically employs layer-by-layer processing by stacking biomaterials layer by layer, resulting in complex three-dimensional structures. The high resolution and precision of each layer is compromised by photolithography and therefore maintained throughout the structure [162]. 2. Multi-material printing: A single material may be difficult to meet the various requirements of a specific repair site, so in recent years researchers have worked to develop technologies that can print multiple biomaterials (cells, scaffolding materials, etc.) at the same time in order to better mimic the complexity of human tissues [163].3. High cell viability: Light-based 3D bioprinting allows for the curing of materials through a rapid photopolymerization reaction. This process which is often rapid, helps to reduce the time the cells are exposed to unsuitable environments, thus improving cell survival. In addition, the technology allows for precise control of the distribution of cells in a three-dimensional structure. Through precise lithography, the cells can be accurately positioned in each layer of the print, allowing them to be evenly distributed throughout the structure [102,164]. Prior to bioprinting, the cells can also be pre-treated, such as co-mixing with scaffolding materials and addition of growth factors, to improve cell adaptation and survival.

To improve the success rate of transplantation of 3D bioprinted constructs, it is important to understand the tissue specificity of the different tissues that are transplanted. For example, different tissues have different mechanical properties and structural characteristics. For example, skeletal tissues require supports with high rigidity and strength [165], while soft tissues such as blood vessels require softer, stretchable supports [130,166], and cardiac tissues even require scaffolding materials that can conduct electrical signals [167]. Therefore, when selecting bio-inks for use in the repair of different tissues, it is necessary to consider whether their mechanical and other physical properties will meet the needs of a particular tissue. In addition, some tissues need to provide support for cell attachment and growth, such as skeletal and vascular tissues [168]. The surface properties and chemical composition of the constructs may need to be adjusted to promote attachment and growth of specific cell types. Furthermore, the degradation properties of the scaffold need to be adjusted to the physiological properties of the target tissue. Some tissues may require prolonged support, while others may require shorter support and allow natural regenerative processes to occur [169,170]. Thus, it is imperative to tailor the characteristics of the scaffold to the tissue specificity of the repair site.

In general, the customization of the constructs can be achieved by selecting special polymers, adjusting the wavelength of the light source and the duration of the light exposure, as well as changing the polymer concentration and selecting suitable photoinitiators. The constructs formed by different photosensitive polymers after photo-crosslinking mostly have different properties. In addition, various nanomaterials such as graphene oxide (GO) and carbon nanotubes (CNTs) or magnetic nanoparticles (MNPs) can be added to the polymer bio-inks [171,172]. Some researchers have found that these nano-additives can simultaneously improve the printability, size-structure stability and mechanical properties of biomaterials [173,174]. The light source also has a great impact on the final properties of the constructs. Different wavelengths often represent different types of light sources (UV-A (320–400 nm), UV-B (290–320 nm), UV-C (200–290 nm) and visible light (400–700 nm)). Most photoinitiators operate in the ultraviolet range, where prolonged exposure to large doses of UV light may affect cell viability or even trigger cell mutations, but short-term UV exposure is essentially harmless to cells [79]. Compared to UV light, visible light is less efficient for photoinitiation although it is more penetrating and effective in providing good cytocompatibility. Light exposure time also significantly affects the physicochemical characteristics of the constructs [84,85,175]. Whether it is UV or visible light, varying the light exposure time changes the degree of cross-linking of the constructs. This may improve

the mechanical strength and stability of the constructs, but it may also lead to excessive cross-linking, affecting the elasticity and plasticity of the constructs. In addition, longer light times may mean longer cross-linking times per layer, which may reduce overall printing speed, affect cell survival, etc. Changing the structure and pore size of the constructs has also been shown to affect the properties and performance of the constructs in a number of ways. Increasing the porosity of a construct can result in a lighter and looser construct, potentially reducing its overall strength and stiffness. Suitable pore size and distribution may provide better cell invasiveness and promote cell growth and differentiation [176]. Therefore the structure and pore design of the constructs need to be considered to mimic the microenvironment of the target tissue. Proper structural and pore design helps the constructs to better integrate with the surrounding tissues and improve biomatchability.

Despite the rapid development of photocrosslinking-based 3D printing technologies in recent years, there are still some challenges to further improve these technologies for clinical translational applications. The main reasons are as follows: 1. Selection of biomaterials: Although a wide range of polymers and additives can be used when designing bio-inks in vitro. However, for true clinical translation, a lower cost, simple to synthesize, but feature-rich bio-ink design is what is needed. Printed biomaterials need to be sufficiently mechanical, biocompatible and degradable at the same time. Finding the ideal biomaterials that can both meet these requirements and be printed is a challenge [177,178]. 2. Cell handling: How to maintain cell survival and function during the printing process and how to accurately position the cells in the desired location is a problem that needs to be solved [179]. 3. Speed and efficiency: Some printing techniques can take a long time to complete, limiting their efficiency in real-world applications. Increasing the speed of printing while maintaining the high precision of the structure is an issue that needs to be addressed. 4. Standardization: There is a lack of standardized methods for light-based 3D bioprinting, which makes it difficult to compare results between different laboratories. The development of uniform standards is crucial for the further development of this technology.

The application of photo-crosslinked 3D bioprinting technology in the field of tissue engineering has been evolving. It can rapidly produce high-precision, highly biocompatible, multi-material and multi-cell type printed constructs with complex geometries for repairing various tissues and organs [180]. In addition, with extensive multidisciplinary exchanges and collaborations in recent years, its unique advantages make it a powerful tool that is expected to advance the field of tissue engineering. Overall, the continuous innovation and development of photo-crosslinked 3D bioprinting technology has brought many new opportunities and challenges for tissue engineering. In the future, as the technology continues to advance, it is believed that it will play an even more important role in the field of medicine and bioscience.

Funding

The National Natural Science Foundation of China (No. 82100714、82170694 and 82370684), the Interdisciplinary Program of Shanghai Jiao Tong University (No. YG2021QN102 and YG2022ZD020), the shanghai “Science and Technology Innovation Action Plan” Medical Innovation Research Project (No. 22Y11905000), the Program of Shanghai Sixth People’s Hospital (No. ynts202004) and the Natural Science Foundation of Shanghai (No. 20ZR1442100).

Ethics approval and consent to participate

Not applicable.

Consent for publication

Not applicable.

Availability of data

Not applicable.

Further disclosure

Not applicable.

CRediT authorship contribution statement

Wenzhuo Fang: Writing – review & editing, Writing – original draft. **Zhenwei Yu:** Writing – review & editing, Visualization, Formal analysis. **Guo Gao:** Formal analysis, Data curation. **Ming Yang:** Methodology, Investigation. **Xuan Du:** Writing – original draft. **Ying Wang:** Conceptualization. **Qiang Fu:** Conceptualization.

Declaration of competing interest

The authors declare they have no competing interests.

Data availability

No data was used for the research described in the article.

Acknowledgments

Not applicable.

References

[1] C. Mandrycky, Z. Wang, K. Kim, D.H. Kim, 3D bioprinting for engineering complex tissues, Biotechnol. Adv. 34 (4) (Jul-Aug 2016) 422–434.
[2] P. Bajaj, R.M. Schweller, A. Khademhosseini, J.L. West, R. Bashir, 3D biofabrication strategies for tissue engineering and regenerative medicine, Annu. Rev. Biomed. Eng. 16 (Jul 11 2014) 247–276.
[3] Q. Feng, D. Li, Q. Li, X. Cao, H. Dong, Microgel assembly: fabrication, characteristics and application in tissue engineering and regenerative medicine, Bioact. Mater. 9 (Mar 2022) 105–119.
[4] G. Gao, X. Cui, Three-dimensional bioprinting in tissue engineering and regenerative medicine, Biotechnol. Lett. 38 (2) (Feb 2016) 203–211.
[5] S. Derakhshanfar, R. Mbeleck, K. Xu, X. Zhang, W. Zhong, M. Xing, 3D bioprinting for biomedical devices and tissue engineering: a review of recent trends and advances, Bioact. Mater. 3 (2) (Jun 2018) 144–156.
[6] J. Zhang, W. Xu, C. Li, et al., Tissue engineering Microtissue: construction, optimization, and application, Tissue Eng Part B Rev 28 (2) (Apr 2022) 393–404.
[7] G.H.D. Almeida, R.P. Iglesia, M.S. Araujo, et al., Uterine tissue engineering: where We Stand and the challenges Ahead, Tissue Eng Part B Rev 28 (4) (Aug 2022) 861–890.
[8] W. Fang, M. Yang, Y. Jin, et al., Injectable decellularized extracellular matrix-based bio-ink with excellent biocompatibility for Scarless Urethra repair 9 (11) (2023) 913.
[9] T. Pose-Boirazian, J. Martínez-Costas, G. Eibes, 3D printing: an emerging technology for Biocatalyst Immobilization, Macromol. Biosci. 22 (9) (Sep 2022) e2200110.
[10] Z. Wang, L. Wang, T. Li, et al., 3D bioprinting in cardiac tissue engineering, Theranostics 11 (16) (2021) 7948–7969.
[11] W. Fang, M. Yang, M. Liu, et al., Review on additives in hydrogels for 3D bioprinting of regenerative medicine: from mechanism to methodology, Pharmaceutics 15 (6) (Jun 9 2023).
[12] S.V. Murphy, P. De Coppi, A. Atala, Opportunities and challenges of translational 3D bioprinting, Nat. Biomed. Eng. 4 (4) (Apr 2020) 370–380.
[13] M.A. Heinrich, W. Liu, A. Jimenez, et al., 3D bioprinting: from Benches to translational applications, Small 15 (23) (Jun 2019) e1805510.
[14] H. Hong, Y.B. Seo, D.Y. Kim, et al., Digital light processing 3D printed silk fibroin hydrogel for cartilage tissue engineering, Biomaterials 232 (Feb 2020) 119679.
[15] F.P. Melchels, J. Feijen, D.W. Grijpma, A review on stereolithography and its applications in biomedical engineering, Biomaterials 31 (24) (Aug 2010) 6121–6130.
[16] L. Andjela, V.M. Abdurahmanovich, S.N. Vladimirovna, G.I. Mikhailovna, D. D. Yurievich, M.Y. Alekseevna, A review on Vat Photopolymerization 3D-printing processes for dental application, Dent. Mater. 38 (11) (Nov 2022) e284–e296.
[17] H. Quan, T. Zhang, H. Xu, S. Luo, J. Nie, X. Zhu, Photo-curing 3D printing technique and its challenges, Bioact. Mater. 5 (1) (Mar 2020) 110–115.
[18] W. Li, M. Wang, H. Ma, F.A. Chapa-Villarreal, A.O. Lobo, Y.S. Zhang, Stereolithography apparatus and digital light processing-based 3D bioprinting for tissue fabrication, iScience 26 (2) (Feb 17 2023) 106039.

[19] R.F. Pereira, P.J. Bártolo, 3D photo-fabrication for tissue engineering and drug delivery, Engineering 1 (1) (2015/03/01/2015) 90–112.
[20] M. Rajput, P. Mondal, P. Yadav, K. Chatterjee, Light-based 3D bioprinting of bone tissue scaffolds with tunable mechanical properties and architecture from photocurable silk fibroin, Int. J. Biol. Macromol. 202 (Mar 31 2022) 644–656.
[21] Y. Li, X. Ren, L. Zhu, C. Li, Biomass 3D printing: principles, materials, post-processing and applications, Polymers 15 (12) (Jun 15 2023).
[22] N. Monteiro, G. Thrivikraman, A. Athirasala, et al., Photopolymerization of cell-laden gelatin methacryloyl hydrogels using a dental curing light for regenerative dentistry, Dent. Mater. 34 (3) (Mar 2018) 389–399.
[23] Fer G. Le, M.L. Becker, 4D printing of Resorbable complex shape-memory poly (propylene fumarate) star scaffolds, ACS Appl. Mater. Interfaces 12 (20) (May 20 2020) 22444–22452.
[24] Q. Feng, D. Li, Q. Li, et al., Assembling microgels via dynamic cross-linking reaction improves printability, microporosity, tissue-adhesion, and self-healing of microgel bioink for extrusion bioprinting, ACS Appl. Mater. Interfaces 14 (13) (Apr 6 2022) 15653–15666.
[25] J. Van Hoorick, L. Tytgat, A. Dobos, et al., Photo-)crosslinkable gelatin derivatives for biofabrication applications, Acta Biomater. 97 (Oct 1 2019) 46–73.
[26] Y. Han, J. Yang, W. Zhao, et al., Biomimetic injectable hydrogel microspheres with enhanced lubrication and controllable drug release for the treatment of osteoarthritis, Bioact. Mater. 6 (10) (Oct 2021) 3596–3607.
[27] P. Kunwar, B.L. Andrada, A. Poudel, et al., Printing double-network Tough hydrogels using temperature-controlled projection stereolithography (TOPS), ACS Appl. Mater. Interfaces 15 (25) (Jun 28 2023) 30780–30792.
[28] W. Liu, M.A. Heinrich, Y. Zhou, et al., Extrusion bioprinting of shear-Thinning gelatin methacryloyl bioinks, Adv Healthc Mater 6 (12) (Jun 2017).
[29] W. Fang, M. Yang, L. Wang, et al., Hydrogels for 3D bioprinting in tissue engineering and regenerative medicine: current progress and challenges, Int J Bioprint 9 (5) (2023) 759.
[30] Z. Zhou, T. Deng, M. Tao, et al., Snail-inspired AFG/GelMA hydrogel accelerates diabetic wound healing via inflammatory cytokines suppression and macrophage polarization, Biomaterials 299 (Aug 2023) 122141.
[31] F.B. Albrecht, F.F. Schmidt, A.C. Volz, P.J. Kluger, Bioprinting of 3D adipose tissue models using a GelMA-bioink with human mature adipocytes or human adipose-derived stem cells, Gels 8 (10) (Sep 25 2022).
[32] M. Zhang, F. Yang, D. Han, et al., 3D bioprinting of corneal decellularized extracellular matrix: GelMA composite hydrogel for corneal stroma engineering, Int J Bioprint 9 (5) (2023) 774.
[33] J. Shen, W. Song, J. Liu, et al., 3D bioprinting by reinforced bioink based on photocurable interpenetrating networks for cartilage tissue engineering, Int. J. Biol. Macromol. 254 (Pt 1) (Oct 25 2023) 127671.
[34] L. Han, J. Xu, X. Lu, et al., Biohybrid methacrylated gelatin/polyacrylamide hydrogels for cartilage repair, J. Mater. Chem. B 5 (4) (Jan 28 2017) 731–741.
[35] A. Marinho, C. Nunes, S. Reis, Hyaluronic acid: a key Ingredient in the Therapy of inflammation, Biomolecules 11 (10) (Oct 15 2021).
[36] C. Li, C. Li, Z. Ma, et al., Regulated macrophage immune microenvironment in 3D printed scaffolds for bone tumor postoperative treatment, Bioact. Mater. 19 (Jan 2023) 474–485.
[37] S. A, M. Zeng, M. Johnson, et al., Green synthetic approach for photo-cross-linkable methacryloyl hyaluronic acid with a tailored substitution degree, Biomacromolecules 21 (6) (Jun 8 2020) 2229–2235.
[38] H. Si, T. Xing, Y. Ding, H. Zhang, R. Yin, W. Zhang, 3D bioprinting of the sustained drug release wound dressing with double-crosslinked hyaluronic-acid-based hydrogels, Polymers 11 (10) (Sep 27 2019).
[39] D. Wang, Y. Guo, J. Zhu, et al., Hyaluronic acid methacrylate/pancreatic extracellular matrix as a potential 3D printing bioink for constructing islet organoids, Acta Biomater. 165 (Jul 15 2023) 86–101.
[40] J.A. Burdick, C. Chung, X. Jia, M.A. Randolph, R. Langer, Controlled degradation and mechanical behavior of photopolymerized hyaluronic acid networks, Biomacromolecules 6 (1) (Jan-Feb 2005) 386–391.
[41] Y. Liu, L. Peng, L. Li, et al., 3D-bioprinted BMSC-laden biomimetic multiphasic scaffolds for efficient repair of osteochondral defects in an osteoarthritic rat model, Biomaterials 279 (Dec 2021) 121216.
[42] R. Dwivedi, S. Kumar, R. Pandey, et al., Polycaprolactone as biomaterial for bone scaffolds: review of literature, J Oral Biol Craniofac Res 10 (1) (Jan-Mar 2020) 381–388.
[43] L. Chen, Z. Li, Y. Zheng, et al., 3D-printed dermis-specific extracellular matrix mitigates scar contraction via inducing early angiogenesis and macrophage M2 polarization, Bioact. Mater. 10 (Apr 2022) 236–246.
[44] D. Olsen, C. Yang, M. Bodo, et al., Recombinant collagen and gelatin for drug delivery, Adv. Drug Deliv. Rev. 55 (12) (Nov 28 2003) 1547–1567.
[45] R. Gibney, J. Patterson, E. Ferraris, High-resolution bioprinting of recombinant human collagen type III, Polymers 13 (17) (Sep 1 2021).
[46] Y. Yang, Z. Wang, Y. Xu, et al., Preparation of chitosan/recombinant human collagen-based photo-responsive bioinks for 3D bioprinting, Gels 8 (5) (May 19 2022).
[47] J.H. Teoh, S.M. Tay, J. Fuh, C.H. Wang, Fabricating scalable, personalized wound dressings with customizable drug loadings via 3D printing, J Control Release 341 (Jan 2022) 80–94.
[48] K.Y. Lee, D.J. Mooney, Alginate: properties and biomedical applications, Prog. Polym. Sci. 37 (1) (Jan 2012) 106–126.
[49] F.W. Holtsberg, C.M. Ensor, M.R. Steiner, J.S. Bomalaski, M.A. Clark, Poly (ethylene glycol) (PEG) conjugated arginine deiminase: effects of PEG formulations on its pharmacological properties, J Control Release 80 (1–3) (Apr 23 2002) 259–271.
[50] S. Xin, O.M. Wyman, D.L. Alge, Assembly of PEG microgels into porous cell-Instructive 3D scaffolds via thiol-ene click chemistry, Adv Healthc Mater 7 (11) (Jun 2018) e1800160.
[51] S. Xin, D. Chimene, J.E. Garza, A.K. Gaharwar, D.L. Alge, Clickable PEG hydrogel microspheres as building blocks for 3D bioprinting, Biomater. Sci. 7 (3) (Feb 26 2019) 1179–1187.
[52] J. Liang, Z. Guo, A. Timmerman, D. Grijpma, A. Poot, Enhanced mechanical and cell adhesive properties of photo-crosslinked PEG hydrogels by incorporation of gelatin in the networks, Biomed Mater 14 (2) (Jan 4 2019) 024102.
[53] S. Zhu, P. Chen, Y. Chen, M. Li, C. Chen, H. Lu, 3D-Printed extracellular matrix/polyethylene glycol diacrylate hydrogel incorporating the anti-inflammatory Phytomolecule Honokiol for regeneration of osteochondral defects, Am. J. Sports Med. 48 (11) (Sep 2020) 2808–2818.
[54] Y. Chen, J. Song, S. Wang, W. Liu, PVA-based hydrogels: Promising Candidates for articular cartilage repair, Macromol. Biosci. 21 (10) (Oct 2021) e2100147.
[55] C.C. DeMerlis, D.R. Schoneker, Review of the oral toxicity of polyvinyl alcohol (PVA), Food Chem. Toxicol. 41 (3) (Mar 2003) 319–326.
[56] X. Chen, B. Lu, D. Zhou, M. Shao, W. Xu, Y. Zhou, Photocrosslinking maleilated hyaluronate/methacrylated poly (vinyl alcohol) nanofibrous mats for hydrogel wound dressings, Int. J. Biol. Macromol. 155 (Jul 15 2020) 903–910.
[57] K.S. Lim, R. Levato, P.F. Costa, et al., Bio-resin for high resolution lithography-based biofabrication of complex cell-laden constructs, Biofabrication 10 (3) (May 11 2018) 034101.
[58] Y. Zhou, Q. Dong, H. Yang, et al., Photocrosslinked maleilated chitosan/methacrylated poly (vinyl alcohol) bicomponent nanofibrous scaffolds for use as potential wound dressings, Carbohydr. Polym. 168 (Jul 15 2017) 220–226.
[59] G. Camci-Unal, D. Cuttica, N. Annabi, D. Demarchi, A. Khademhosseini, Synthesis and characterization of hybrid hyaluronic acid-gelatin hydrogels, Biomacromolecules 14 (4) (Apr 8 2013) 1085–1092.
[60] A. García-Lizarribar, X. Fernández-Garibay, F. Velasco-Mallorquí, A.G. Castaño, J. Samitier, J. Ramon-Azcon, Composite biomaterials as long-Lasting scaffolds for 3D bioprinting of highly aligned muscle tissue, Macromol. Biosci. 18 (10) (Oct 2018) e1800167.
[61] S. Xin, K.A. Deo, J. Dai, et al., Generalizing hydrogel microparticles into a new class of bioinks for extrusion bioprinting, Sci. Adv. 7 (42) (Oct 15 2021) eabk3087.
[62] X. Zhou, M. Nowicki, H. Sun, et al., 3D bioprinting-tunable small-diameter blood vessels with biomimetic Biphasic cell layers, ACS Appl. Mater. Interfaces 12 (41) (Oct 14 2020) 45904–45915.
[63] X. Zhou, H. Cui, M. Nowicki, et al., Three-dimensional-bioprinted dopamine-based matrix for promoting neural regeneration, ACS Appl. Mater. Interfaces 10 (10) (Mar 14 2018) 8993–9001.
[64] S.A. Costa, J.R. Simon, M. Amiram, et al., Photo-crosslinkable Unnatural Amino acids enable facile synthesis of thermoresponsive nano- to microgels of Intrinsically Disordered Polypeptides, Adv Mater. 30 (5) (Feb 2018).
[65] J.E. Samorezov, C.M. Morlock, E. Alsberg, Dual ionic and photo-crosslinked alginate hydrogels for Micropatterned spatial control of material properties and cell behavior, Bioconjug Chem 26 (7) (Jul 15 2015) 1339–1347.
[66] H.W. Ooi, C. Mota, A.T. Ten Cate, A. Calore, L. Moroni, M.B. Baker, Thiol-ene alginate hydrogels as versatile bioinks for bioprinting, Biomacromolecules 19 (8) (Aug 13 2018) 3390–3400.
[67] M. Kesti, M. Müller, J. Becher, et al., A versatile bioink for three-dimensional printing of cellular scaffolds based on thermally and photo-triggered tandem gelation, Acta Biomater. 11 (Jan 2015) 162–172.
[68] P.A. Levett, F.P. Melchels, K. Schrobback, D.W. Hutmacher, J. Malda, T.J. Klein, A biomimetic extracellular matrix for cartilage tissue engineering centered on photocurable gelatin, hyaluronic acid and chondroitin sulfate, Acta Biomater. 10 (1) (Jan 2014) 214–223.
[69] M. Peroglio, D. Eglin, L.M. Benneker, M. Alini, S. Grad, Thermoreversible hyaluronan-based hydrogel supports in vitro and ex vivo disc-like differentiation of human mesenchymal stem cells, Spine J. 13 (11) (Nov 2013) 1627–1639.
[70] Y. Liu, C.W. Wong, S.W. Chang, S.H. Hsu, An injectable, self-healing phenol-functionalized chitosan hydrogel with fast gelling property and visible light-crosslinking capability for 3D printing, Acta Biomater. 122 (Mar 1 2021) 211–219.
[71] A. Zennifer, S. Manivannan, S. Sethuraman, S.G. Kumbar, D. Sundaramurthi, 3D bioprinting and photocrosslinking: emerging strategies & future perspectives, Biomater. Adv. 134 (Mar 2022) 112576.
[72] D. Kulms, E. Zeise, B. Pöppelmann, T. Schwarz, DNA damage, death receptor activation and reactive oxygen species contribute to ultraviolet radiation-induced apoptosis in an essential and independent way, Oncogene 21 (38) (Aug 29 2002) 5844–5851.
[73] K.S. Lim, B.J. Klotz, G.C.J. Lindberg, et al., Visible light cross-linking of gelatin hydrogels offers an enhanced cell microenvironment with improved light penetration depth, Macromol. Biosci. 19 (6) (Jun 2019) e1900098.
[74] A. Bagheri, J. Jin, Photopolymerization in 3D printing, ACS Appl. Polym. Mater. 1 (4) (2019/04/12 2019) 593–611.
[75] N. Hanamura, H. Ohashi, Y. Morimoto, T. Igarashi, Y. Tabata, Viability evaluation of layered cell sheets after ultraviolet light irradiation of 222 nm, Regen Ther 14 (Jun 2020) 344–351.
[76] J.F. Sánchez-Pérez, D. Vicente-Agullo, M. Barberá, E. Castro-Rodríguez, M. Cánovas, Relationship between ultraviolet index (UVI) and first-, second- and third-degree sunburn using the Probit methodology, Sci. Rep. 9 (1) (Jan 24 2019) 733.

[77] R. Masuma, S. Kashima, M. Kurasaki, T. Okuno, Effects of UV wavelength on cell damages caused by UV irradiation in PC12 cells, J. Photochem. Photobiol., B 125 (Aug 5 2013) 202–208.

[78] D. Mackenzie, Ultraviolet light Fights new Virus, Engineering (Beijing) 6 (8) (Aug 2020) 851–853.

[79] H. Xu, J. Casillas, S. Krishnamoorthy, C. Xu, Effects of Irgacure 2959 and lithium phenyl-2,4,6-trimethylbenzoylphosphinate on cell viability, physical properties, and microstructure in 3D bioprinting of vascular-like constructs, Biomed Mater 15 (5) (Aug 7 2020) 055021.

[80] S. Sakai, H. Ohi, T. Hotta, H. Kamei, M. Taya, Differentiation potential of human adipose stem cells bioprinted with hyaluronic acid/gelatin-based bioink through microextrusion and visible light-initiated crosslinking, Biopolymers 109 (2) (Feb 2018).

[81] R. Goto, E. Nishida, S. Kobayashi, et al., Gelatin methacryloyl-riboflavin (GelMA-RF) hydrogels for bone regeneration, Int. J. Mol. Sci. 22 (4) (Feb 6 2021).

[82] R.Z. Lin, Y.C. Chen, R. Moreno-Luna, A. Khademhosseini, J.M. Melero-Martin, Transdermal regulation of vascular network bioengineering using a photopolymerizable methacrylated gelatin hydrogel, Biomaterials 34 (28) (Sep 2013) 6785–6796.

[83] S. Kim, Y. Kang, Á.E. Mercado-Pagán, W.J. Maloney, Y. Yang, In vitro evaluation of photo-crosslinkable chitosan-lactide hydrogels for bone tissue engineering, J. Biomed. Mater. Res. B Appl. Biomater. 102 (7) (Oct 2014) 1393–1406.

[84] H. Stratesteffen, M. Köpf, F. Kreimendahl, A. Blaeser, S. Jockenhoevel, H. Fischer, GelMA-collagen blends enable drop-on-demand 3D printablility and promote angiogenesis, Biofabrication 9 (4) (Sep 1 2017) 045002.

[85] Q. Zhang, H.P. Bei, M. Zhao, Z. Dong, X. Zhao, Shedding light on 3D printing: printing photo-crosslinkable constructs for tissue engineering, Biomaterials 286 (Jul 2022) 121566.

[86] Z. Luo, H. Zhang, R. Chen, et al., Digital light processing 3D printing for microfluidic chips with enhanced resolution via dosing- and zoning-controlled vat photopolymerization, Microsyst Nanoeng 9 (2023) 103.

[87] Q.Z. Chen, A. Bismarck, U. Hansen, et al., Characterisation of a soft elastomer poly(glycerol sebacate) designed to match the mechanical properties of myocardial tissue, Biomaterials 29 (1) (Jan 2008) 47–57.

[88] X. Ma, C. Yu, P. Wang, et al., Rapid 3D bioprinting of decellularized extracellular matrix with regionally varied mechanical properties and biomimetic microarchitecture, Biomaterials 185 (Dec 2018) 310–321.

[89] C.L. Nijst, J.P. Bruggeman, J.M. Karp, et al., Synthesis and characterization of photocurable elastomers from poly(glycerol-co-sebacate), Biomacromolecules 8 (10) (Oct 2007) 3067–3073.

[90] A. Bhattacharyya, G. Janarthanan, T. Kim, et al., Modulation of bioactive calcium phosphate micro/nanoparticle size and shape during in situ synthesis of photo-crosslinkable gelatin methacryloyl based nanocomposite hydrogels for 3D bioprinting and tissue engineering, Biomater. Res. 26 (1) (Oct 8 2022) 54.

[91] A. Bandyopadhyay, B.B. Mandal, N. Bhardwaj, 3D bioprinting of photo-crosslinkable silk methacrylate (SilMA)-polyethylene glycol diacrylate (PEGDA) bioink for cartilage tissue engineering, J. Biomed. Mater. Res. 110 (4) (Apr 2022) 884–898.

[92] A. Chiappone, E. Fantino, I. Roppolo, et al., 3D printed PEG-based hybrid Nanocomposites obtained by Sol-gel technique, ACS Appl. Mater. Interfaces 8 (8) (Mar 2 2016) 5627–5633.

[93] B. Farkas, S. Dante, F. Brandi, Photoinitiator-free 3D scaffolds fabricated by excimer laser photocuring, Nanotechnology 28 (3) (Jan 20 2017) 034001.

[94] J. Jang, T.G. Kim, B.S. Kim, S.W. Kim, S.M. Kwon, D.W. Cho, Tailoring mechanical properties of decellularized extracellular matrix bioink by vitamin B2-induced photo-crosslinking, Acta Biomater. 33 (Mar 2016) 88–95.

[95] A. Stiles, T.A. Tison, L. Pruitt, U. Vaidya, Photoinitiator selection and concentration in photopolymer formulations towards large-Format additive manufacturing, Polymers 14 (13) (Jul 1 2022).

[96] J.R. Choi, K.W. Yong, J.Y. Choi, A.C. Cowie, Recent advances in photo-crosslinkable hydrogels for biomedical applications, Biotechniques 66 (1) (Jan 2019) 40–53.

[97] D.E. Godar, C. Gurunathan, I. Ilev, 3D bioprinting with UVA1 radiation and photoinitiator Irgacure 2959: can the ASTM standard L929 cells Predict human stem cell cytotoxicity? Photochem. Photobiol. 95 (2) (Mar 2019) 581–586.

[98] Z. Wang, H. Kumar, Z. Tian, et al., Visible light photoinitiation of cell-adhesive gelatin methacryloyl hydrogels for stereolithography 3D bioprinting, ACS Appl. Mater. Interfaces 10 (32) (Aug 15 2018) 26859–26869.

[99] J.Y. Shin, Y.H. Yeo, J.E. Jeong, S.A. Park, W.H. Park, Dual-crosslinked methylcellulose hydrogels for 3D bioprinting applications, Carbohydr. Polym. 238 (Jun 15 2020) 116192.

[100] N. Diamantides, L. Wang, T. Pruiksma, et al., Correlating rheological properties and printability of collagen bioinks: the effects of riboflavin photocrosslinking and pH, Biofabrication 9 (3) (Jul 5 2017) 034102.

[101] D. Petta, A.R. Armiento, D. Grijpma, M. Alini, D. Eglin, M. D'Este, 3D bioprinting of a hyaluronan bioink through enzymatic-and visible light-crosslinking, Biofabrication 10 (4) (Sep 25 2018) 044104.

[102] J. Yin, M. Yan, Y. Wang, J. Fu, H. Suo, 3D bioprinting of low-concentration cell-laden gelatin methacrylate (GelMA) bioinks with a two-step cross-linking strategy, ACS Appl. Mater. Interfaces 10 (8) (Feb 28 2018) 6849–6857.

[103] L. Hao, X. Tao, M. Feng, et al., Stepwise multi-cross-linking bioink for 3D Embedded bioprinting to promote full-thickness wound healing, ACS Appl. Mater. Interfaces 15 (20) (May 24 2023) 24034–24046.

[104] Q.L. Loh, C. Choong, Three-dimensional scaffolds for tissue engineering applications: role of porosity and pore size, Tissue Eng Part B Rev 19 (6) (Dec 2013) 485–502.

[105] Y. Ma, M. Lin, G. Huang, et al., 3D Spatiotemporal mechanical microenvironment: a hydrogel-based platform for Guiding stem cell fate, Adv Mater 30 (49) (Dec 2018) e1705911.

[106] Y. Zhou, C.M. Duque, C.D. Santangelo, R.C.J.A.F.M. Hayward, Biasing Buckling direction in shape-Programmable hydrogel sheets with through-thickness gradients 29 (48) (2019) 1905273.

[107] M. Wang, W. Li, L.S. Mille, et al., Digital light processing based bioprinting with composable gradients, Adv Mater 34 (1) (Jan 2022) e2107038.

[108] R.G. Pearson, R. Bhandari, R.A. Quirk, K.M. Shakesheff, Recent advances in tissue engineering, J. Long Term Eff. Med. Implants 27 (2–4) (2017) 199–231.

[109] H. Cui, W. Zhu, M. Nowicki, X. Zhou, A. Khademhosseini, L.G. Zhang, Hierarchical fabrication of engineered vascularized bone Biphasic constructs via dual 3D bioprinting: Integrating regional bioactive factors into Architectural design, Adv Healthc Mater 5 (17) (Sep 2016) 2174–2181.

[110] Z. Wan, P. Zhang, Y. Liu, L. Lv, Y. Zhou, Four-dimensional bioprinting: current developments and applications in bone tissue engineering, Acta Biomater. 101 (Jan 1 2020) 26–42.

[111] Y. Yan, H. Chen, H. Zhang, et al., Vascularized 3D printed scaffolds for promoting bone regeneration, Biomaterials 190–191 (Jan 2019) 97–110.

[112] O. Guillaume, M.A. Geven, C.M. Sprecher, et al., Surface-enrichment with hydroxyapatite nanoparticles in stereolithography-fabricated composite polymer scaffolds promotes bone repair, Acta Biomater. 54 (May 2017) 386–398.

[113] B. Byambaa, N. Annabi, K. Yue, et al., Bioprinted osteogenic and Vasculogenic patterns for engineering 3D bone tissue, Adv Healthc Mater 6 (16) (Aug 2017).

[114] Y. Yang, Q. Zhang, T. Xu, et al., Photocrosslinkable nanocomposite ink for printing strong, biodegradable and bioactive bone graft, Biomaterials 263 (Dec 2020) 120378.

[115] K.W. Boere, J. Visser, H. Seyednejad, et al., Covalent attachment of a three-dimensionally printed thermoplast to a gelatin hydrogel for mechanically enhanced cartilage constructs, Acta Biomater. 10 (6) (Jun 2014) 2602–2611.

[116] D.J. Huey, J.C. Hu, K.A. Athanasiou, Unlike bone, cartilage regeneration remains elusive, Science 338 (6109) (Nov 16 2012) 917–921.

[117] L. Bian, M. Guvendiren, R.L. Mauck, J.A. Burdick, Hydrogels that mimic developmentally relevant matrix and N-cadherin interactions enhance MSC chondrogenesis, Proc Natl Acad Sci U S A. 110 (25) (Jun 18 2013) 10117–10122.

[118] J.W. Hayami, S.D. Waldman, B.G. Amsden, Chondrocyte generation of cartilage-like tissue following Photoencapsulation in methacrylated polysaccharide solution blends, Macromol. Biosci. 16 (7) (Jul 2016) 1083–1095.

[119] M. Costantini, J. Idaszek, K. Szöke, et al., 3D bioprinting of BM-MSCs-loaded ECM biomimetic hydrogels for in vitro neocartilage formation, Biofabrication 8 (3) (Jul 19 2016) 035002.

[120] E.C. Beck, M. Barragan, T.B. Libeer, et al., Chondroinduction from naturally derived cartilage matrix: a comparison between Devitalized and decellularized cartilage encapsulated in hydrogel Pastes, Tissue Eng Part A 22 (7–8) (Apr 2016) 665–679.

[121] K.L. Spiller, S.A. Maher, A.M. Lowman, Hydrogels for the repair of articular cartilage defects, Tissue Eng Part B Rev 17 (4) (Aug 2011) 281–299.

[122] L.A. Callahan, A.M. Ganios, D.L. McBurney, et al., ECM production of primary human and bovine chondrocytes in hybrid PEG hydrogels containing type I collagen and hyaluronic acid, Biomacromolecules 13 (5) (May 14 2012) 1625–1631.

[123] H. Cao, X. Wang, M. Chen, et al., Childhood cartilage ECM enhances the chondrogenesis of Endogenous cells and Subchondral bone repair of the Unidirectional collagen-dECM scaffolds in combination with Microfracture, ACS Appl. Mater. Interfaces 13 (48) (Dec 8 2021) 57043–57057.

[124] D.O. Visscher, H. Lee, P.P.M. van Zuijlen, et al., A photo-crosslinkable cartilage-derived extracellular matrix bioink for auricular cartilage tissue engineering, Acta Biomater. 121 (Feb 2021) 193–203.

[125] M. Potente, H. Gerhardt, P. Carmeliet, Basic and therapeutic aspects of angiogenesis, Cell 146 (6) (Sep 16 2011) 873–887.

[126] L. de Silva, P.N. Bernal, A. Rosenberg, J. Malda, R. Levato, D. Gawlitta, Biofabricating the vascular tree in engineered bone tissue, Acta Biomater. 156 (Jan 15 2023) 250–268.

[127] H. Jiang, X. Li, T. Chen, et al., Bioprinted vascular tissue: Assessing functions from cellular, tissue to organ levels, Mater Today Bio 23 (Dec 2023) 100846.

[128] M. Vliora, C. Ravelli, E. Grillo, M. Corsini, A.D. Flouris, S. Mitola, The impact of adipokines on vascular networks in adipose tissue, Cytokine Growth Factor Rev. 69 (Feb 2023) 61–72.

[129] W. Jia, P.S. Gungor-Ozkerim, Y.S. Zhang, et al., Direct 3D bioprinting of perfusable vascular constructs using a blend bioink, Biomaterials 106 (Nov 2016) 58–68.

[130] L. De Moor, J. Smet, M. Plovyt, et al., Engineering microvasculature by 3D bioprinting of prevascularized spheroids in photo-crosslinkable gelatin, Biofabrication 13 (4) (Sep 21 2021).

[131] W. Zhu, X. Qu, J. Zhu, et al., Direct 3D bioprinting of prevascularized tissue constructs with complex microarchitecture, Biomaterials 124 (Apr 2017) 106–115.

[132] P. Wang, Y. Sun, X. Shi, H. Shen, H. Ning, H. Liu, 3D printing of tissue engineering scaffolds: a focus on vascular regeneration, Biodes Manuf 4 (2) (2021) 344–378.

[133] E.L. Bastos, F.H. Quina, M.S. Baptista, Endogenous Photosensitizers in human skin, Chem Rev 123 (16) (Aug 23 2023) 9720–9785.

[134] M.M. Severn, A.R. Horswill, Staphylococcus epidermidis and its dual lifestyle in skin health and infection, Nat. Rev. Microbiol. 21 (2) (Feb 2023) 97–111.

[135] A.M. Jorgensen, A. Gorkun, N. Mahajan, et al., Multicellular bioprinted skin facilitates human-like skin architecture in vivo, Sci. Transl. Med. 15 (716) (Oct 4 2023) eadf7547.

[136] T. Baltazar, B. Jiang, A. Moncayo, et al., 3D bioprinting of an implantable xeno-free vascularized human skin graft, Bioeng Transl Med 8 (1) (Jan 2023) e10324.
[137] S. Michael, H. Sorg, C.T. Peck, et al., Tissue engineered skin substitutes created by laser-assisted bioprinting form skin-like structures in the dorsal skin fold chamber in mice, PLoS One 8 (3) (2013) e57741.
[138] M. Falcone, B. De Angelis, F. Pea, et al., Challenges in the management of chronic wound infections, J Glob Antimicrob Resist 26 (Sep 2021) 140–147.
[139] M.H. Kathawala, W.L. Ng, D. Liu, et al., Healing of chronic wounds: an Update of recent developments and future possibilities, Tissue Eng Part B Rev 25 (5) (Oct 2019) 429–444.
[140] D. Simões, S.P. Miguel, M.P. Ribeiro, P. Coutinho, A.G. Mendonça, I.J. Correia, Recent advances on antimicrobial wound dressing: a review, Eur. J. Pharm. Biopharm. 127 (Jun 2018) 130–141.
[141] Z. Li, A. Zheng, Z. Mao, et al., Silk fibroin-gelatin photo-crosslinked 3D-bioprinted hydrogel with MOF-methylene blue nanoparticles for infected wound healing, Int J Bioprint 9 (5) (2023) 773.
[142] B. Chu, J-m He, Z. Wang, et al., Proangiogenic Peptide Nanofiber hydrogel/3D Printed Scaffold for Dermal Regeneration, 424, 2021 128146.
[143] F. Zhou, Y. Hong, R. Liang, et al., Rapid printing of bio-inspired 3D tissue constructs for skin regeneration, Biomaterials 258 (Nov 2020) 120287.
[144] J. Cho, H. Lee, W. Rah, H.J. Chang, Y.S. Yoon, From engineered heart tissue to cardiac organoid, Theranostics 12 (6) (2022) 2758–2772.
[145] G. Iacobellis, Epicardial adipose tissue in contemporary cardiology, Nat. Rev. Cardiol. 19 (9) (Sep 2022) 593–606.
[146] S. Cho, D.E. Discher, K.W. Leong, G. Vunjak-Novakovic, J.C. Wu, Challenges and opportunities for the next generation of cardiovascular tissue engineering, Nat. Methods 19 (9) (Sep 2022) 1064–1071.
[147] R. Zaman, S. Epelman, Resident cardiac macrophages: Heterogeneity and function in health and disease, Immunity 55 (9) (Sep 13 2022) 1549–1563.
[148] Z. Ma, S. Koo, M.A. Finnegan, et al., Three-dimensional filamentous human diseased cardiac tissue model, Biomaterials 35 (5) (Feb 2014) 1367–1377.
[149] S. Anil Kumar, M. Alonzo, S.C. Allen, et al., A visible light-cross-linkable, fibrin-gelatin-based bioprinted construct with human cardiomyocytes and fibroblasts, ACS Biomater. Sci. Eng. 5 (9) (Sep 9 2019) 4551–4563.
[150] K. Zhu, S.R. Shin, T. van Kempen, et al., Gold nanocomposite bioink for printing 3D cardiac constructs, Adv. Funct. Mater. 27 (12) (Mar 24 2017).
[151] A. Mathur, Z. Ma, P. Loskill, S. Jeeawoody, K.E. Healy, In vitro cardiac tissue models: current status and future prospects, Adv. Drug Deliv. Rev. 96 (Jan 15 2016) 203–213.
[152] S. Ostrovidov, S. Salehi, M. Costantini, et al., 3D bioprinting in skeletal muscle tissue engineering, Small 15 (24) (Jun 2019) e1805530.
[153] J.M. Grasman, M.J. Zayas, R.L. Page, G.D. Pins, Biomimetic scaffolds for regeneration of volumetric muscle loss in skeletal muscle injuries, Acta Biomater. 25 (Oct 2015) 2–15.
[154] X. Cui, G. Gao, Y. Qiu, Accelerated myotube formation using bioprinting technology for biosensor applications, Biotechnol. Lett. 35 (3) (Mar 2013) 315–321.
[155] C. Cvetkovic, R. Raman, V. Chan, et al., Three-dimensionally printed biological machines powered by skeletal muscle, Proc Natl Acad Sci U S A 111 (28) (Jul 15 2014) 10125–10130.
[156] Y. Morimoto, H. Onoe, S. Takeuchi, Biohybrid robot powered by an antagonistic pair of skeletal muscle tissues, Sci. Robot. 3 (18) (May 30 2018).
[157] M. Yeo, G. Kim, Three-dimensional Microfibrous bundle structure fabricated using an electric field-assisted/cell printing process for muscle tissue regeneration, ACS Biomater. Sci. Eng. 4 (2) (Feb 12 2018) 728–738.
[158] J.H. Kim, Y.J. Seol, I.K. Ko, et al., 3D bioprinted human skeletal muscle constructs for muscle function Restoration, Sci. Rep. 8 (1) (Aug 17 2018) 12307.
[159] T.A. Dixon, E. Cohen, D.M. Cairns, et al., Bioinspired three-dimensional human neuromuscular junction development in Suspended hydrogel Arrays, Tissue Eng Part C Methods 24 (6) (Jun 2018) 346–359.
[160] K. Yu, X. Zhang, Y. Sun, et al., Printability during projection-based 3D bioprinting, Bioact. Mater. 11 (May 2022) 254–267.
[161] S. Sang, X. Mao, Y. Cao, et al., 3D bioprinting using Synovium-derived MSC-laden photo-cross-linked ECM bioink for cartilage regeneration, ACS Appl. Mater. Interfaces (Feb 13 2023).
[162] A. Zaupa, C. Terraza, P.N. Abarzúa-Illanes, et al., A Psychrophilic GelMA: Breaking Technical and Immunological barriers for multimaterial high-resolution 3D bioprinting, Biomacromolecules 24 (1) (Jan 9 2023) 150–165.
[163] A. Ding, S.J. Lee, S. Ayyagari, R. Tang, C.T. Huynh, E. Alsberg, 4D biofabrication via instantly generated graded hydrogel scaffolds, Bioact. Mater. 7 (Jan 2022) 324–332.
[164] L. Nie, Y. Sun, O.V. Okoro, Y. Deng, G. Jiang, A. Shavandi, Click chemistry for 3D bioprinting, Mater. Horiz. 10 (8) (Jul 31 2023) 2727–2763.
[165] J. Gehlen, W. Qiu, G.N. Schädli, R. Müller, X.H. Qin, Tomographic volumetric bioprinting of heterocellular bone-like tissues in seconds, Acta Biomater. 156 (Jan 15 2023) 49–60.
[166] U. Aizarna-Lopetegui, C. García-Astrain, C. Renero-Lecuna, et al., Remodeling arteries: studying the mechanical properties of 3D-bioprinted hybrid photoresponsive materials, J. Mater. Chem. B 11 (39) (Oct 11 2023) 9431–9442.
[167] Y.J. Shin, R.T. Shafranek, J.H. Tsui, J. Walcott, A. Nelson, D.H. Kim, 3D bioprinting of mechanically tuned bioinks derived from cardiac decellularized extracellular matrix, Acta Biomater. 119 (Jan 1 2021) 75–88.
[168] A. Tong, R. Voronov, A Minireview of microfluidic scaffold materials in tissue engineering, Front. Mol. Biosci. 8 (2021) 783268.
[169] Q. Mei, J. Rao, H.P. Bei, Y. Liu, X. Zhao, 3D bioprinting photo-crosslinkable hydrogels for bone and cartilage repair, Int J Bioprint 7 (3) (2021) 367.
[170] M. Diba, G.L. Koons, M.L. Bedell, A.G. Mikos, 3D printed colloidal biomaterials based on photo-reactive gelatin nanoparticles, Biomaterials 274 (Jul 2021) 120871.
[171] L. Jiang, Y. Wang, Z. Liu, et al., Three-dimensional printing and injectable conductive hydrogels for tissue engineering application, Tissue Eng Part B Rev 25 (5) (Oct 2019) 398–411.
[172] S.R. Shin, H. Bae, J.M. Cha, et al., Carbon nanotube reinforced hybrid microgels as scaffold materials for cell encapsulation, ACS Nano 6 (1) (Jan 24 2012) 362–372.
[173] Y. Zuo, X. Liu, D. Wei, et al., Photo-cross-linkable methacrylated gelatin and hydroxyapatite hybrid hydrogel for modularly engineering biomimetic osteon, ACS Appl. Mater. Interfaces 7 (19) (May 20 2015) 10386–10394.
[174] Y. Li, S. Peng, J.-T. Miao, et al., Isotropic Stereolithography Resin Toughened by Core-Shell Particles, 394, 2020 124873.
[175] D.E. Hagaman, S. Leist, J. Zhou, H.F. Ji, Photoactivated polymeric Bilayer Actuators fabricated via 3D printing, ACS Appl. Mater. Interfaces 10 (32) (Aug 15 2018) 27308–27315.
[176] P. Dorishetty, R. Balu, A. Gelmi, J.P. Mata, N.K. Dutta, N.R. Choudhury, 3D printable Soy/silk hybrid hydrogels for tissue engineering applications, Biomacromolecules 22 (9) (Sep 13 2021) 3668–3678.
[177] S.J. Min, J.S. Lee, H. Nah, et al., Development of photo-crosslinkable platelet lysate-based hydrogels for 3D printing and tissue engineering, Biofabrication 13 (4) (Aug 16 2021).
[178] L. Tytgat, L. Van Damme, M.D.P. Ortega Arevalo, et al., Extrusion-based 3D printing of photo-crosslinkable gelatin and κ-carrageenan hydrogel blends for adipose tissue regeneration, Int. J. Biol. Macromol. 140 (Nov 1 2019) 929–938.
[179] H. Wang, J. Tian, Y. Jiang, et al., A 3D biomimetic optoelectronic scaffold repairs cranial defects, Sci. Adv. 9 (7) (Feb 15 2023) eabq7750.
[180] P. Zhang, H. Wang, P. Wang, et al., Lightweight 3D bioprinting with point by point photocuring, Bioact. Mater. 6 (5) (May 2021) 1402–1412.