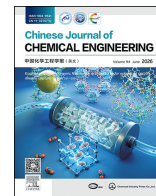




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Full Length Article

Polyurethane-decellularized extracellular matrix electrospun composite fibers as urethral scaffold

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ABSTRACT

Scaffolding is less intrusive for damaged urethral tissue regeneration compared to tissue grafting. The study focuses on polyurethane-decellularized extracellular matrix (PU-dECM) scaffolds electrospun produced from varying PU-dECM concentrations and electrospinning voltages for the production of tunable biomimetic scaffolds as an alternative specialized urethral restoration method. Porcine urethra dECM was produced through immersion-sonication at 0.5% sodium dodecyl sulfate (SDS), followed by solubilization. PU and dECM were dissolved in 7:3 dimethyl sulfoxide: ethanol (DMSO:EtOH) solvent at ratios of 90:10, 85:15, and 80:20 for a total polymer concentration of 9.5%. Following this, electrospinning was conducted at 29–35 kV, 3.75 ml·h⁻¹ flowrate for 90 min. Viscosity decreased while conductivity increased with increasing dECM concentration. Increasing dECM concentration also decreased the Young's moduli, and the concentration-voltage interaction introduced a significant effect on the Young's moduli. The treatments of 80:20–29 kV, 90:10–29 kV, and 90:10–32 kV (PU:dECM ratio-electrospinning voltage) were predicted to have Young's moduli close to that of the human urethra at (0.034 ± 0.01) MPa and were selected as candidate fibers for characterization. Fibers with random orientation and diameters of 1.05–1.41, 1.34–1.51 and 1.30–1.45 μm respectively were produced. Coating of dECM is observed from FTIR spectra. For urethral scaffold applications, the 90:10@29 kV fibers reported the best hydrophilicity at 46.34 ° ± 24.07°.

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1. Introduction

Factors of genetic, environmental, or physical origins can cause damage to tissue or organs in the human body. It can occur that the natural regeneration rate of the body cannot match the extent of such damage. When this happens, interventional reconstruction measures are performed to promote tissue growth, commonly through grafting tissue from autologous, homologous, or heterologous sources [1]. However, these methods face a challenge in the treatment of urological organ damage as grafting of reconstructive tissue can lead to damage on both the grafting source and application sites. These sources also face issues in supply [1,2]. For the urethra, while buccal mucosa grafts are quite successful in treating

urethral strictures, the occurrence of complications after the surgery is still high which requires further surgical treatment [3].

Considering the need to mimic the native extracellular matrix (ECM) to the nanoscale, fibers as biomimetic scaffolds have been of interest as an alternative restoration method. Electrospinning is the prevalent production method due to its simplicity, efficiency, and versatility in producing micro to nanofibers with high surface area to volume ratios whose morphology and mechanical properties can be controlled. The process' customizability enables a high degree of fine-tuning in making a scaffold that is morphologically analogous to the native ECM which leads to high biocompatibility [4]. Among the parameters of interest for the electrospinning process is the applied voltage field, however the effects of varying this parameter remain unclear especially on its effect to the fiber diameter with conflicting reports in Refs. [5,6].

Material selection and blending is important to match the chemical and biological properties of the scaffold to the native ECM. The biodegradability and strength of the scaffold, as well as the electrospinning compatibility are dictated by the base polymer/s selected [7]. Decellularized extracellular matrices (dECM) are ECM

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components of an organ with cellular components removed. dECM that are sourced from heterologous tissue with respect to the damaged organ has higher biocompatibility due to similar compositional makeup of relevant proteins and bioactive molecules such as collagen, elastin, proteoglycans, and glycoproteins [8,9]. This removes the risk of an immune response while also greatly improving a scaffold's biocompatibility through increased cell adhesion, alignment, and proliferation [9–12]. However, pure dECM-based scaffolds face difficulties in implementation due to its low mechanical strength and fast biodegradation [13–17]. This requires supplementation with other polymers that improve said characteristics. Polyurethane (PU) is notable for its mechanical properties, biodegradability, biocompatibility without sacrificing processing ease [18] and rigidity during sterilization of the scaffold [19]. Hence, an appropriate mix of PU to dECM can be performed to tune the properties of the produced fiber for the desired application.

While studies on the applications of PU-dECM scaffolds on several tissue sites such as the heart and blood vessels [20–23], nerve [24,25], and soft tissue [26,27] have been conducted, these are currently limited, and no study has yet to characterize PU-dECM scaffolds targeted for urethral applications. Similarly, while there have been studies that conducted analysis on electrospun PU scaffolds for soft tissue and urethral applications [28,29], no attempt has been made in testing the variation of electrospinning parameters, specifically the applied voltage, for the production of PU-dECM composite scaffolds. Hence, this research aims to develop and characterize the surface morphology, mechanical strength, surface chemical composition, and hydrophilicity of electrospun PU-dECM of varied proportional blends and operating voltage parameters for its potential applications in urethral scaffolding in addressing the need for a specialized restorative method for urethral impairment.

2. Methodology

2.1. Decellularization of porcine urethra and solubilization of porcine urethra dECM

Fresh porcine entrails were collected from white, female adult Duroc pigs weighing 85–100 kg. The porcine urethras were separated from the entrails, connective tissue, and fat. The tubes were opened up, cut into thin slices, and cleaned in 250 ml of distilled water under stirring at 600 r·min⁻¹ for 1 h. The liquid was drained, and the vessel was filled with 300 ml of 0.5% sodium dodecyl sulfate (SDS) solution. This was sonicated at 60 W power with occasional replacement of detergent solution until the tissues turn translucent white. Once white, these were washed with 250 ml of 1% (vol) Triton X-100 for 1 h with stirring at 600 r·min⁻¹. Following this is a wash with distilled water, after which washing with 250 ml of 1X PBS is performed for 1 h under stirring at 600 r·min⁻¹. The decellularized pieces were transferred to a fresh 1X PBS batch for storage at 4 °C until lyophilization for 24 h and grinding. Remaining DNA content through DNA quantification assays (Qubit™ dsDNA Quantification Assay) ranged from 2.30 to 38.4 ng double-strand DNA, indicating 83.42% to 98.08% removal of DNA. GAG quantification was not performed in this study, however decellularization of porcine urethra with 0.5% SDS has been reported to produce dECM with a dry mass composition of 20.8% collagen, 5.4% GAGs, and 18.3% elastin [30]. While sonication is introduced in the process to speed up the decellularization process, it has also been reported that the introduction of sonication does not have a significant effect on the GAG and elastin content of the decellularized ECM of other tissues [31,32].

For the solubilization process, 2.0 g ECM flakes were mixed with 1.0 g pepsin and 100 ml of 0.01 mol·L⁻¹ HCl for 7 h, after which filtration through a syringe pump was performed. This was neutralized under an ice bath at 4 °C with 0.1 mol·L⁻¹ NaOH with balancing of salts through 0.9 ml of 10X PBS and 0.3 ml of 1X PBS. Once neutralized, the pre-hydrogel was lyophilized for 24 h and powdered.

2.2. Polymer blend preparation

PU pellets (poly[4,4'-methylenebis(phenyl isocyanate)-alt-1,4-butanediol/di(propylene glycol)/polycaprolactone], Sigma Aldrich CAS 68084-39-9) and the dECM powder were mixed in the corresponding trials to produce 100 ml solutions with 9.5% total polymer mass per volume in 7:3 dimethyl sulfoxide: ethanol (DMSO:EtOH) solvent. These solutions were stirred at 600 r·min⁻¹ at room temperature until the disappearance of undissolved aggregates. The solution compositions are presented in Table 1.

2.3. Electrospinning process

Electrospinning environmental conditions were monitored through the equipment system (Inovenso NE300 Multinozzle Electrospinning Equipment) for humidity and temperature. Recorded temperatures for the production process ranged from 24.4 °C to 27.7 °C, while recorded humidities ranged from 48% to 65%. The polymer solution was put in a 10 ml syringe pump with a 22-gauge needle, after which the electrospinning voltage was applied with a solution flow rate of 3.75 ml·h⁻¹, a fiber collection time of 90 min, and a tip-to-collector distance of 7.5 cm. The different blends are electrospun at three different voltage levels: 29, 32 and 35 kV. Fibers collected at a rotating mandrel with a 30 cm diameter that is spinning at 130 r·min⁻¹. After electrospinning, the fibers were stored in a humidity-controlled environment at 25% to 35% relative humidity until it was to be used. Before characterization tests, the scaffolds were freeze-dried for 24 h.

2.4. Blend and fiber characterization

The polymer blend viscosity was measured through the use of a digital viscometer (Brookfield Ametek DV2T Viscometer) to obtain values in poise units. An electric conductivity meter (Jenway 4510 Conductivity Meter) was used to measure the solution conductivity. The polymer blends were characterized in duplicate.

Fiber characterization was performed in triplicates. For fiber characterization, a universal testing machine (Zwick Roell Z0.5) with a 0.5 kN load at 500 mm·min⁻¹ speed start position and a grip separation of 30 mm was used for the analysis of the mechanical properties of the scaffolds. 6 cm rectangular cuts of the scaffold were used for the mechanical tests. Selection of the scaffolds for the succeeding characterization steps were performed based on the reported Young's Moduli for human urethra at (0.034 ± 0.010) MPa [33].

Table 1
Polymer blend composition.

PU:dECM ratio	Volume of solvent/ml	Total polymer mass/g	PU mass/g	dECM mass/g
80:20	100	9.5	7.60	1.90
85:15	100	9.5	8.08	1.42
90:10	100	9.5	8.55	0.95
100:0	100	9.5	9.50	0.00

Scanning electron microscopy (JSM IT500HR/LA Sschottky Tip Field Emission SEM) of the fibers was performed in which images of the fibers were analyzed through ImageJ-DiameterJ for fiber diameter measurement. FTIR spectroscopy (PerkinElmer Frontier FT-IR Spectrometer) was performed in comparison to the pure PU and dECM. Water contact angle goniometry (Biolin Scientific Theta Lite Optical Tensiometer) was performed at intervals within 10 s after the placement of the distilled water droplet of 2 μl onto the scaffold samples that are placed horizontally. Three droplet positions were noted per scaffold replicate, from which the average water contact angle was calculated.

3. Results and Discussion

3.1. Polymer blend characterization

The dynamic viscosity of the polymer blends over time are provided in Fig. 1. Hydrolysis of dECM through pepsin digestion allowed it to be dissolved in the DMSO:EtOH system for a homogenous system with polyurethane. DMSO:EtOH are capable of dipole-dipole and hydrogen bonding interactions with functional groups of PU and dECM for its solvation. The PU:dECM solutions exhibited rheopectic behavior. This is ideal for electrospinning because the lower initial apparent viscosity of the solution would require lower electric field strength to overcome surface tension for Taylor cone formation at the nozzle. This decreased initial resistance also promotes Taylor cone elongation towards the collector, aiding smaller diameter fiber production. During jetting, the jet is less likely to break due to the increasing viscosity over time that leads to continuity of fiber streams [34]. As the dECM concentration is increased, the viscosity of the solution decreases due to the shorter chain length of dECM compared to PU. The PU-PU interaction forces are more dominant compared to the PU-dECM and dECM-dECM interaction forces due to their higher concentration. Hence, the van der Waals and dipole-dipole forces between PU chains decrease when dECM concentration is increased as it cannot compensate for the lost PU-PU interactions, leading to lowered viscosity.

The addition of PU to the 7:3 DMSO:EtOH solvent caused a slight increase in the average conductivity of the solution from 0.03 to 0.14 $\mu\text{S}\cdot\text{cm}^{-1}$. The addition of dECM causes a substantial jump in the values of the solution conductivity as seen in Fig. 2. The jump can be attributed to the dECM itself and the salt impurity which is 4.53% (mass) after neutralization and PBS salt addition. dECM can increase the conductivity of the polymer

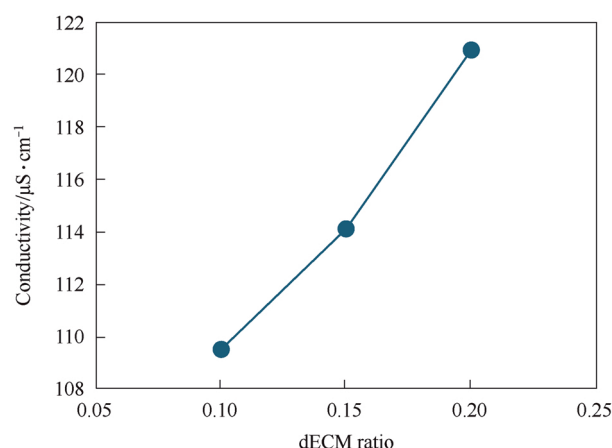


Fig. 2. Average conductivity vs. dECM ratio of the polymer blends.

solution due to the presence of polar groups and quasi-ring structures for π bonds which enable electrophilic interactions [35]. Conductive solutions are preferred as they facilitate a higher degree of interaction between the electric field and the solution in the jetting process. The recorded conductivities are within the range of 10^{-11} – 10^{-1} $\text{S}\cdot\text{m}^{-1}$ as an indicator for conductivity on electrospinning processes [36]. As electrospinning and electrospaying processes are congruent processes that deal with electrohydrodynamic interactions, the factor that primarily dictates the regime to produce particulates, jet as fibers, or produce an intermediate for beaded fibers is dictated by the chain length of the components [36]. Hence, the range provided can be seen as applicable also to electrospinning processes.

3.2. Mechanical properties of PU-dECM fibers and selection of candidate fibers

Presented in Fig. 3 are the stress-strain plots for representative trials of the produced fibers. The summary of the Young's moduli for the fibers is presented in Table 2.

For statistical analysis, natural log transformation was used for the Young's moduli response to provide a better fit of the model and reduce the impact of outliers. The model for Young's moduli response is seen as significant, such that only a 0.01% chance of an F -value of 9.84 can be obtained due to noise attribution. For the

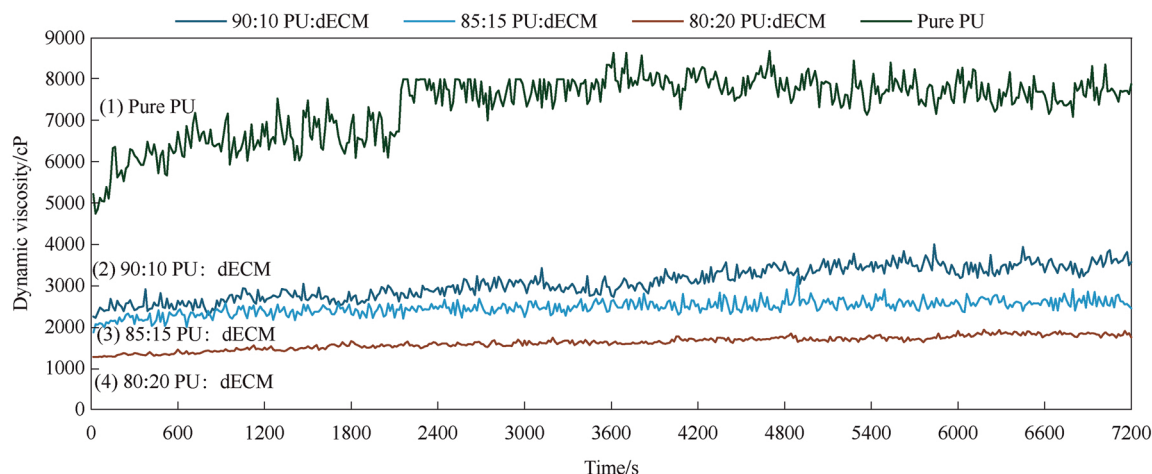


Fig. 1. Dynamic viscosity vs. time plots of the polymer blends.

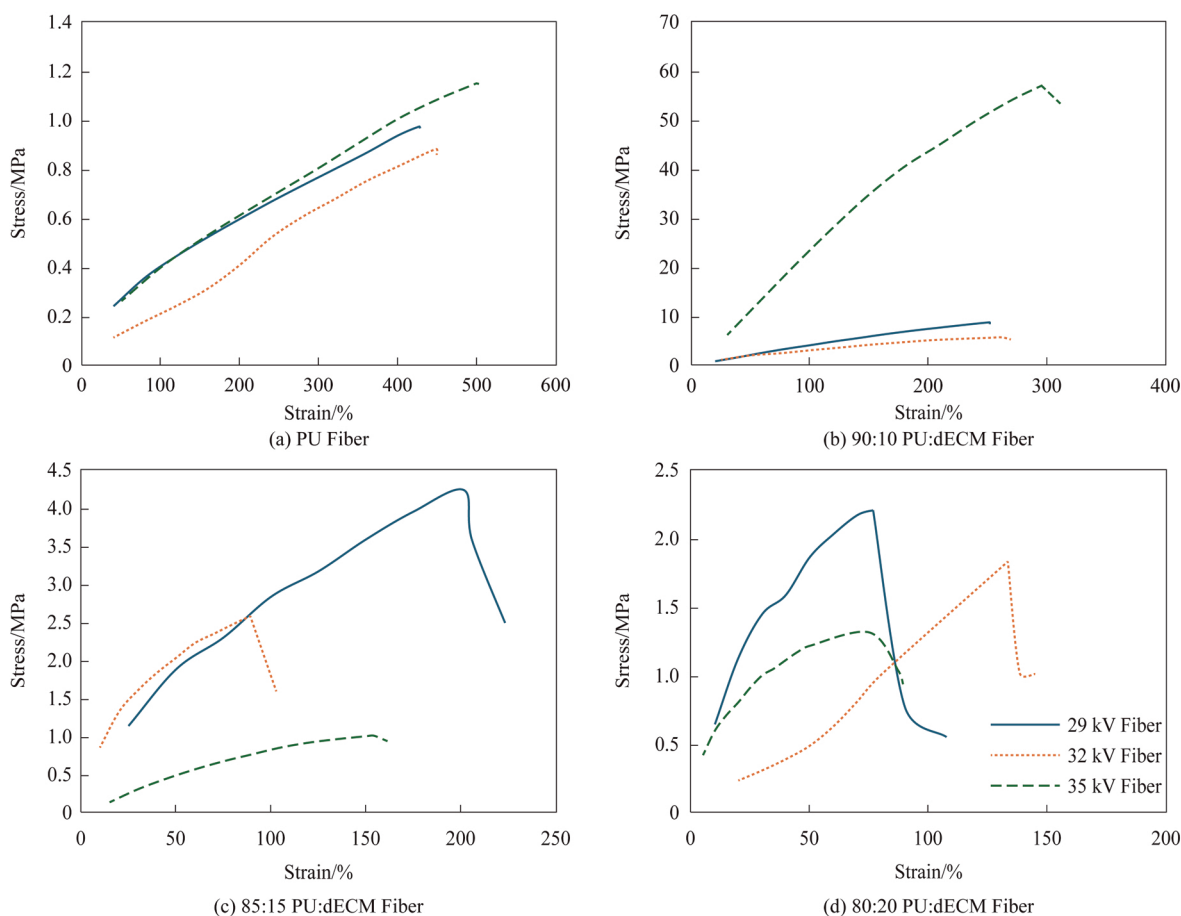


Fig. 3. Stress-strain plots of representatives of the produced PU (a) and PU:dECM fibers (b–d).

Table 2

Summary of Young's Modulus (MPa) of the produced scaffolds.

PU:dECM Concentration	Electrospinning Voltage/kV		
	29	32	35
100:0	0.0020 ± 0.00036	0.0019 ± 0.00015	0.0016 ± 0.00055
90:10	0.0469 ± 0.0293	0.0765 ± 0.0979	0.1429 ± 0.1027
85:15	0.0193 ± 0.0039	0.0200 ± 0.0077	0.0125 ± 0.0104
80:20	0.0472 ± 0.0329	0.0109 ± 0.0044	0.0066 ± 0.0091

Table 3

Predicted and actual Young's modulus response (MPa) of the candidate treatments.

Treatment Name	PU:dECM ratio	Voltage level/kV	Predicted Young's modulus response/MPa	Actual Young's modulus response/MPa
80:20 PU:dECM @ 29 kV	80:20	29	0.040	0.047 ± 0.033
90:10 PU:dECM @ 29 kV	90:10	29	0.041	0.047 ± 0.029
90:10 PU:dECM @ 32 kV	90:10	32	0.042	0.076 ± 0.098

Young's Moduli, dECM concentration and the interaction between dECM concentration and voltage have significant effects, but voltage alone does not present significant effects. The dECM concentration and its interaction with the voltage are generally the main predictors of the mechanical properties of the scaffold. Natural polymers have weak mechanical strengths [37]. At the same total polymer concentrations, increasing the dECM ratio would lead to a lower mechanical strength response. Increasing the dECM ratio also leads to a decrease in the overall viscosity of the polymer blend. Decreased viscosities can lead to higher jet instability and breakage during the electrospinning process, which can further weaken the resulting fibers [34]. As discussed previously in Section 3.1, increasing the dECM ratio also led to increased solution conductivities. This in turn allows the solution to be attracted more to the collector theoretically until the critical conductivity is reached. At the critical conductivity, the balance

between the surface charge density and the tangential electric field is disrupted, and higher conductivities promote the reduction of the tangential electric field which is harmful in the formation of the Taylor cone and only leads to a sharp cone extrusion [38]. Because the solutions tested are assumed to be at the border of the critical conductivity, it is harder to electrospun, thus leading to the formation of spurts of fiber jets and compromise of the continuity and strength of the fibers.

It is desired for tissue engineering applications that the mechanical strength of the fiber is similar to that of the native tissue site. Cell differentiation in scaffolds is promoted when the scaffold mechanical properties are in close agreement with the target site tissue [39]. The Young's modulus of the human urethra tissue was determined from literature to be at (0.034 ± 0.01) MPa [33]. Optimization based on this range was performed, and the candidate treatments are presented in Table 3. These scaffolds are then

selected to proceed for the tests for morphology, surface chemical composition, and hydrophilicity.

3.3. Morphology of candidate PU-dECM fibers

Presented in Fig. 4 are representative SEM images of the selected fiber treatments in Table 3 and its distributions. Shown in Table 4 are the average fiber diameters per replicate for the sample size of 105 fibers per image. It can be observed from Fig. 4 that the fibers obtained generally present uniformity in the fiber diameter size and surface roughness, however knotting of the fibers along with

Table 4

Average fiber diameters (μm) of the candidate treatments.

Treatment	Replicate 1	Replicate 2	Replicate 3
80:20 PU:dECM @ 29 kV	1.41 ± 0.41	1.18 ± 0.54	1.05 ± 0.30
90:10 PU:dECM @ 29 kV	1.51 ± 0.42	1.40 ± 0.33	1.34 ± 0.42
90:10 PU:dECM @ 32 kV	1.30 ± 0.34	1.45 ± 0.42	1.40 ± 0.40

random fiber alignment can be observed. Sudden increases in the fiber diameter are also noted for areas wherein knotting is present. This can be seen as a manifestation of the critical conductivity

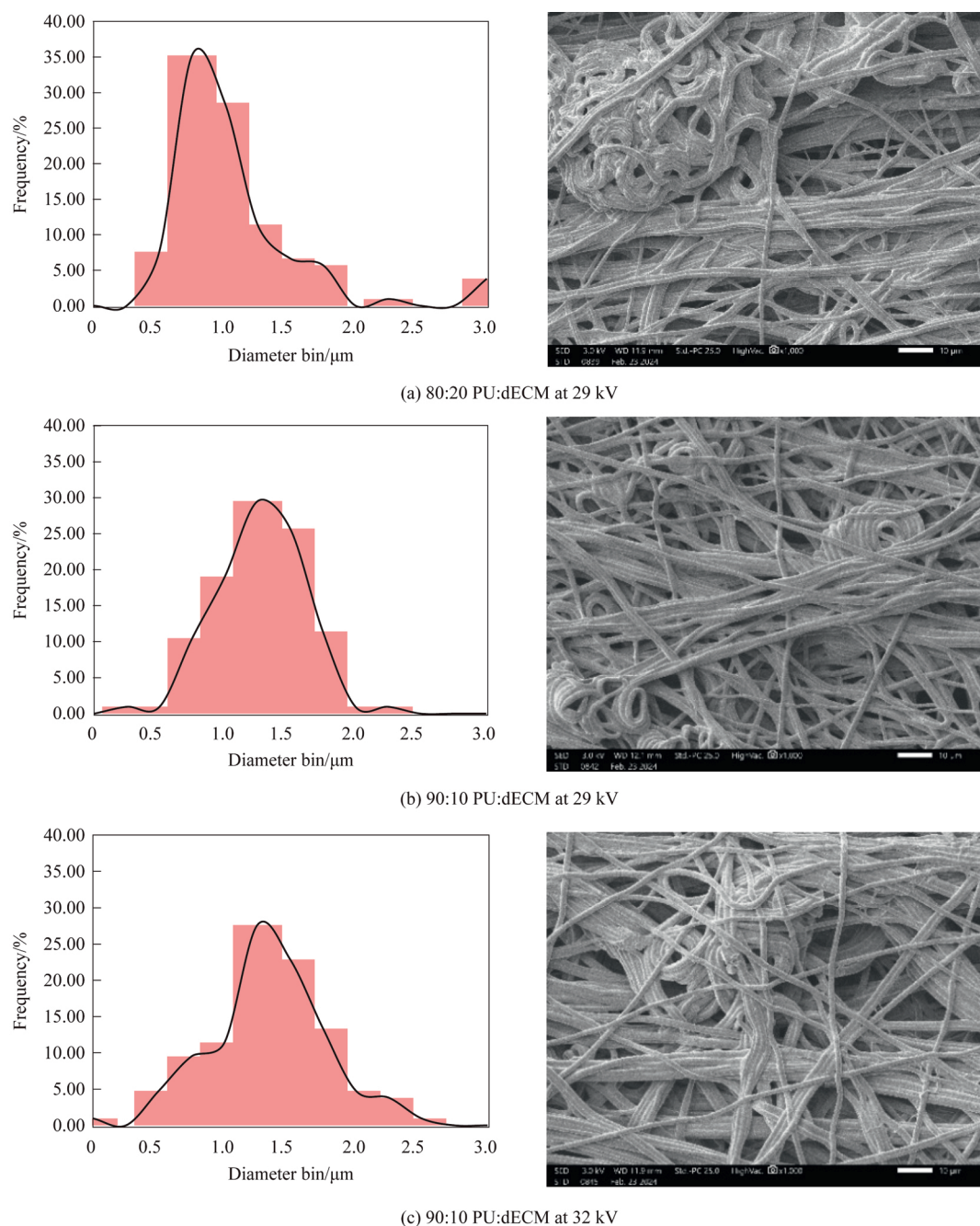


Fig. 4. Representative SEM images of fibers: (a) 80:20 PU:dECM at 29 kV, (b) 90:10 PU:dECM at 29 kV, and (c) 90:10 PU:dECM at 32 kV.

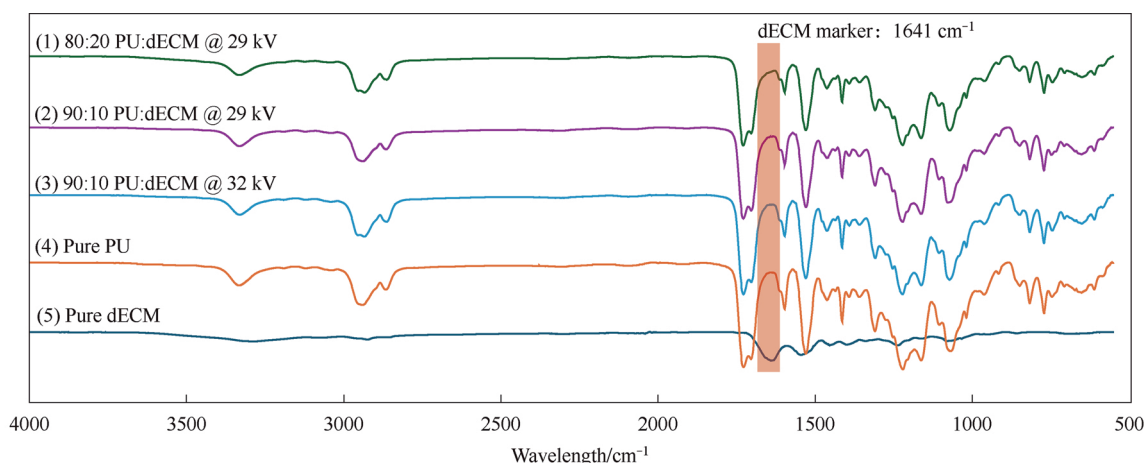


Fig. 5. FTIR spectra for contrasting of fibers with pure substances.

threshold wherein a conical spurting or webbing is observed rather than the production of a singular fine jet. This conical spurting for PU:dECM blends suggests that the regime of the process is between the electrospinning and electrospinning regime for beaded fiber morphology, which can be attributed to the short chain length of the dECM. In this case, there is not enough stretching action as a larger volume of the solution is ejected per spurt, which contributes to the larger fiber diameters in that area.

The produced fibers had average fiber diameters within a micron, with nano and microfibers being noted. Statistically, there is no significant difference between the obtained average fiber diameters for the three candidate treatments. An *F*-value of 2.23 was obtained. This suggests that regardless of the added amount of ECM within 10% to 20% or the change of voltage from 29 kV to 32 kV, the fiber diameter would not have a significant increase. One of the primary goals of tissue engineering scaffolds is the mimicking of the structural make-up of the native extracellular matrix, in which the structural proteins of the ECM range around 50–500 nm in diameter that serves as a good backbone for cell attachment later [40]. In the makeup of the native ECM, collagen fibrils bundle up to become fibers with diameters ranging from 500 to 3000 nm [41]. While the produced PU:dECM fibers did not have average diameters within the 50 to 500 nm range that is usually desirable for tissue scaffolds, they are still within the size range for ECM fibers. As the main function of the native ECM collagen fibers is the structural integrity support of the tissue and the produced scaffolds have similar tensile strengths as that of the human urethra, these shows promise in the potential use of the scaffolds for urethral tissue engineering applications.

3.4. Surface chemical composition of candidate PU-dECM fibers

Fig. 5 presents the representative spectra of each blend to contrast with the pure PU and dECM.

For PU fibers, the following are the characteristic peaks: 3332 cm^{-1} for the N—H stretching of secondary amides as is the case with urethanes, 2941 cm^{-1} for the C—H stretching of alkanes, 1728 cm^{-1} and 1702 cm^{-1} for the strong C=O stretching of esters and ketones, 1530 cm^{-1} for the N—H bending of amines, 1413 cm^{-1} for the O—H bending of alcohols, 1218 cm^{-1} for the C—N stretching of amines, and 1070 cm^{-1} for the stretching of primary alcohols [42]. The alkane presence can be attributed to the polycaprolactone segment, while the ester presence is characteristic to that of the urethane bond. On the other hand, the following

peaks are the characteristic peaks for dECM: 3300 cm^{-1} for the N—H stretching of the amide A band, 1641 cm^{-1} for the N—H stretching and bending of the amide I C=O, 1543 cm^{-1} for the amide II N—H bending vibration coupled to C—N stretching, and 1234 cm^{-1} for the amide III band C—H stretching. These observations fall in line with the literature for the FTIR spectra of dECM [43,44]. The lack of a peak at 1041–1052 cm^{-1} denotes that no residual DMSO solvent is left, indicating its removal during the electrospinning and freeze-drying processes.

The distinct PU markers are the peaks at the wavenumber of 1728 cm^{-1} and 1702 cm^{-1} while the dECM marker is the peak at 1641 cm^{-1} (highlighted red in Fig. 5). These are the expected markers that should appear on the PU:dECM blend fibers to indicate proper mixing of components. It can be observed from Fig. 5 that the fibers fall in close agreement with each other in the markers for the surface chemical composition. When comparing the PU:dECM fibers to the pure substances, it can be observed that the PU:dECM blends follow the peaks of PU. However, it can be observed that the distinct characteristic peak of dECM was not present in the PU:dECM fibers as highlighted in red in Fig. 5. What could have occurred is the coating of dECM by the PU. Despite the solubilization of the dECM, there are still preferential intermolecular and intramolecular interactions between its polar groups compared to PU which is primarily non-polar. This can lead to aggregations of dECM that can be coated. PU, on the other hand, is a larger molecule in comparison and is present at higher concentrations which can interfere with the signal the dECM can provide.

3.5. Hydrophilicity of candidate PU-dECM fibers

Representative images of the replicates are presented in Fig. 6 while a summary of the obtained values is shown in Table 5.

All produced blends are hydrophilic in nature, with water contact angles of less than 90° being obtained. An *F*-value of 4.68 was observed and a significant difference was observed in post-hoc comparison between the 80:20 PU:dECM treatment at 29 kV and the 90:10 PU:dECM treatment at 29 kV. From the baseline of 90:10 PU:dECM produced at 29 kV, increasing the dECM concentration to 20% or the voltage to 32 kV also increased the hydrophilicity.

An increase in the dECM concentration corresponded to an increase in hydrophilicity as the functional groups of dECM promote polar intermolecular interactions with water. While PU has polar functional groups, these are diluted by the non-polar aromatic rings in the isocyanate hard segment and the alkane chain of

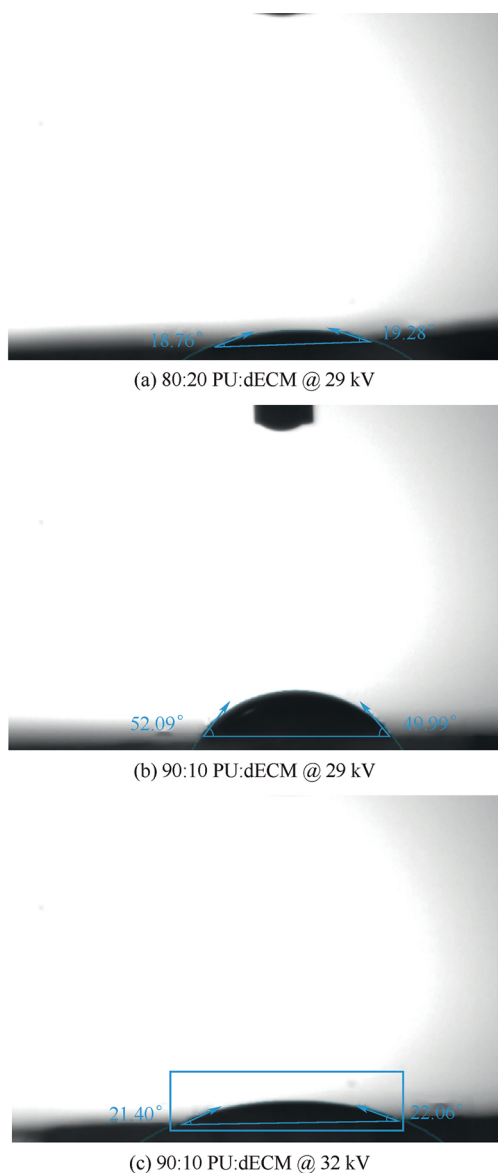


Fig. 6. Representative water contact angle imagery for (a) 80:20 PU:dECM @ 29 kV, (b) 90:10 PU:dECM @ 29 kV, and (c) 90:10 PU:dECM @ 32 kV.

the polycaprolactone soft segment. By introducing dECM, the hydrophilicity is therefore increased. For the voltage effect, with the same resistive load that is the polymer solution, an increase in current accompanies an increase in voltage. As more charge is induced, there is increased repulsion in between PU chains which are filled by dECM. Due to this, an increased voltage can lead to a more even and accessible distribution of dECM across the PU, which increases its hydrophilic effect despite the coating of dECM with PU.

Table 5
Average water contact angle after 10 seconds.

Treatment	Average water contact angle/(°)
80:20 PU:dECM at 29 kV	16.56 ± 14.05
90:10 PU:dECM at 29 kV	46.34 ± 24.07
90:10 PU:dECM at 32 kV	25.64 ± 23.80

It has been reported that being too hydrophilic can be detrimental to the growth of the cells along a tissue site as it prevents proper cellular attachment and interactions with protein bodies. A contact angle between 40° and 60° provided better proliferation compared to more hydrophilic scaffolds in different cell types ranging from ovarian cells, fibroblasts and endothelial cells [45], osseous tissue [46], and urothelium cells [28]. Considering this, the hydrophilicity of the 90:10 PU:dECM scaffold electrospun at 29 kV is within this range.

4. Conclusions

Fibers close to a micron in diameter with random alignment were produced, and three electrospinning voltage-concentration conditions produced fibers that matched the mechanical strength of the human urethra. The fiber diameter is seen as a result of the reduced stretching time during the electrospinning process, and as a result, the desired 50–500 nm diameter threshold was not met. Despite this, the fibers were still able to mimic the diameters of the native collagen fibers at 500–3000 nm in the ECM which can be beneficial in the production of bilayer scaffolds with other nanofibers. The random alignment of the scaffolds can be a point of improvement as aligned scaffolds are preferred. Coating of dECM in PU was observed. Hydrophilicity was enhanced with an accompanying increase in voltage or concentration. However, moderate hydrophilicity is more desirable for tissue scaffold applications. Hence, the 90:10 PU:dECM produced at 29 kV was deemed to be the best parameter setting for urethral applications. Therefore, a polyurethane-decellularized extracellular matrix (PU-dECM) based composite fiber has been successfully fabricated for urethral scaffold applications.

CRediT Authorship Contribution Statement

Piolo Gabriel N. Gervacio: Writing – review & editing, Writing – original draft, Methodology, Investigation, Funding acquisition, Formal analysis, Conceptualization. Joseph Rey H. Sta Agueda: Writing – review & editing, Supervision, Resources, Methodology, Conceptualization. Allan N. Soriano: Writing – review & editing, Supervision, Methodology, Conceptualization. Nathaniel P. Dugos: Writing – review & editing, Supervision, Resources, Methodology, Funding acquisition, Conceptualization.

Availability of Data and Materials

The data from this study can be obtained from the corresponding author upon reasonable request.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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References

- [1] A. Atala, Tissue engineering in urology, *Curr. Urol. Rep.* 2 (1) (2001) 83–92.
- [2] M. Zamani, N. Shakhssalim, S. Ramakrishna, M. Naji, Electrospinning: application and prospects for urologic tissue engineering, *Front. Bioeng. Biotechnol.* 8 (2020) 579925.
- [3] M. Spilotos, N. Sihra, S. Malde, M.H. Pakzad, R. Hamid, J.L. Ockrim, T.J. Greenwell, Buccal mucosal graft urethroplasty in men-risk factors for recurrence and complications: a third referral centre experience in anterior urethroplasty using buccal mucosal graft, *Transl. Androl. Urol.* 6 (3) (2017) 510–516.
- [4] S. Sell, C. Barnes, M. Smith, M. McClure, P. Madurantakam, J. Grant, M. McManus, G. Bowlin, Extracellular matrix regenerated: tissue engineering via electrospun biomimetic nanofibers, *Polym. Int.* 56 (11) (2007) 1349–1360.
- [5] Y. Bagbi, A. Pandey, P.R. Solanki, Electrospun nanofibrous filtration membranes for heavy metals and dye removal, in: *Nanoscale Materials in Water Purification*, Elsevier, 2019, pp. 75–288.
- [6] Q. Wei, D. Tao, Y. Xu, Nanofibers: Principles and Manufacture, . Functional Nanofibers and Their Applications, Elsevier, 2012 pp –21.
- [7] G.C. Ingavle, J.K. Leach, Advancements in electrospinning of polymeric nanofibrous scaffolds for tissue engineering, *Tissue Eng. Part B Rev.* 20 (4) (2014) 277–293.
- [8] C. Kaya, B. Sahin, The role of extracellular matrix proteins in the urinary tract: a literature review, in: *Composition and Function of the Extracellular Matrix in the Human Body*, InTech, 2016.
- [9] J.T. Cao, S.C. Luo, W.L. Huang, X.C. Lu, R.K. Kankala, S.B. Wang, P.Y. Xu, A.Z. Chen, ECM-inspired stem cell secretome sustained releasing composite nanofibrous membranes for accelerated wound healing, *Mater. Today Bio* 34 (2025) 102141.
- [10] B.N. Brown, S.F. Badylak, Extracellular matrix as an inductive scaffold for functional tissue reconstruction, *Transl. Res.* 163 (4) (2014) 268–285.
- [11] S. Sun, J. Qin, Y.F. Zhuang, P.F. Cai, X. Yu, H.S. Wang, X.M. Mo, J.L. Wu, M. El-Newehy, M.M. Abdulhameed, M.Y. Fan, W.H. Qian, B.B. Sun, Development of MgO-loaded PLA/dECM antibacterial nanofibrous membranes for enhanced gingival regeneration, *Biomater. Sci.* 13 (12) (2025) 3354–3366.
- [12] M.J. He, L.S. Li, Y.J. Liu, Z.K. Wu, Y.M. Xu, L. Xiao, K. Luo, X.C. Xu, Decellularized extracellular matrix coupled with polycaprolactone/laponite to construct a biomimetic barrier membrane for bone defect repair, *Int. J. Biol. Macromol.* 276 (Pt 1) (2024) 133775.
- [13] E. Lih, K.W. Park, S.Y. Chun, H. Kim, T.G. Kwon, Y.K. Joong, D.K. Han, Biomimetic porous PLGA scaffolds incorporating decellularized extracellular matrix for kidney tissue regeneration, *ACS Appl. Mater. Interfaces* 8 (33) (2016) 21145–21154.
- [14] E. Lih, W. Park, K.W. Park, S.Y. Chun, H. Kim, Y.K. Joong, T.G. Kwon, J.A. Hubbell, D.K. Han, A bioinspired scaffold with anti-inflammatory magnesium hydroxide and decellularized extracellular matrix for renal tissue regeneration, *ACS Cent. Sci.* 5 (3) (2019) 458–467.
- [15] W.Y. Sun, Y.Y. Yang, L. Wang, H. Tang, L. Zhang, Y.L. She, X.F. Xiao, X.F. Hu, Q. Feng, C. Chen, Utilization of an acellular cartilage matrix-based photocrosslinking hydrogel for tracheal cartilage regeneration and circumferential tracheal repair, *Adv. Funct. Mater.* 33 (32) (2023) 2307406.
- [16] S. Yelleswarapu, A. Dash, S. Chameettachal, F. Pati, 3D bioprinting of tissue constructs employing dual crosslinking of decellularized extracellular matrix hydrogel, *Biomater. Adv.* 152 (2023) 213494.
- [17] W.Z. Fang, M. Yang, Y.W. Jin, K.L. Zhang, Y. Wang, M. Liu, Y.H. Wang, R.X. Yang, Q. Fu, Injectable decellularized extracellular matrix-based bio-ink with excellent biocompatibility for scarless urethra repair, *Gels* 9 (11) (2023) 913.
- [18] Y. Xu, J. Guan, Interaction of cells with polyurethane scaffolds, in: *Advances in Polyurethane Biomaterials*, Elsevier, 2016, pp. 23–542.
- [19] I. Łopianiak, B.A. Butruk-Raszeja, Evaluation of sterilization/disinfection methods of fibrous polyurethane scaffolds designed for tissue engineering applications, *Int. J. Mol. Sci.* 21 (21) (2020) 8092.
- [20] E. Tawagi, T. Ganesh, H.M. Cheng, J.P. Santerre, Synthesis of degradable-polar-hydrophobic-ionic co-polymeric microspheres by membrane emulsion photopolymerization: *in vitro* and *in vivo* studies, *Acta Biomater.* 89 (2019) 279–288.
- [21] C. Alperin, P.W. Zandstra, K.A. Woodhouse, Polyurethane films seeded with embryonic stem cell-derived cardiomyocytes for use in cardiac tissue engineering applications, *Biomaterials* 26 (35) (2005) 7377–7386.
- [22] S.J. Specht, S. Rohringer, P. Hager, C. Grasl, A.M. Schmitt, V.J.C. Pach, K. Ehrmann, S. Baudis, R. Liska, H. Kiss, K.H. Schneider, B.K. Podesser, H. Bergmeister, Decellularized extracellular matrix and polyurethane vascular grafts have positive effects on the inflammatory and pro-thrombotic state of aged endothelial cells, *J. Biomed. Mater. Res.* 113 (1) (2025) e37830.
- [23] I.C.P. Rodrigues, E.S.N. Lopes, K.D. Pereira, S.C. Huber, A.L. Jardini, J.M. Anichino-Bizzacchi, A.D. Luchessi, L.P. Gabriel, Extracellular matrix-derived and low-cost proteins to improve polyurethane-based scaffolds for vascular grafts, *Sci. Rep.* 12 (1) (2022) 5230.
- [24] Y.W. Chen, C.C. Chen, H.Y. Ng, C.W. Lou, Y.S. Chen, M.Y. Shie, Additive manufacturing of nerve decellularized extracellular matrix-contained polyurethane conduits for peripheral nerve regeneration, *Polymers* 11 (10) (2019) 1612.
- [25] Y.C. Wang, J.J. Lin, J.L. Chen, R.C. Liang, Q. Zhang, J.H. Li, M. Shi, L. Li, X.L. He, T. Lan, X.H. Hui, H. Tan, Biodegradable polyurethane-incorporating decellularized spinal cord matrix scaffolds enhance Schwann cell reprogramming to promote peripheral nerve repair, *J. Mater. Chem. B* 11 (10) (2023) 2115–2128.
- [26] L.C. Da, M. Gong, A.J. Chen, Y. Zhang, Y.Z. Huang, Z.J. Guo, S.F. Li, J. Li-Ling, L. Zhang, H.Q. Xie, Composite elastomeric polyurethane scaffolds incorporating small intestinal submucosa for soft tissue engineering, *Acta Biomater.* 59 (2017) 45–57.
- [27] X. Jiang, X.Z. Xiong, Y.X. Lin, Y.R. Lu, J.Q. Cheng, N.S. Cheng, J. Zhang, A composite scaffold fabricated with an acellular matrix and biodegradable polyurethane for the *in vivo* regeneration of pig bile duct defects, *Acta Biomater.* 150 (2022) 238–253.
- [28] Y.Q. Niu, G.C. Liu, C.B. Chen, M. Fu, W. Fu, Z. Zhao, H.M. Xia, F.J. Stadler, Urethral reconstruction using an amphiphilic tissue-engineered autologous polyurethane nanofiber scaffold with rapid vascularization function, *Biomater. Sci.* 8 (8) (2020) 2164–2174.
- [29] Y. Hong, A. Huber, K. Takamari, N.J. Amoroso, R. Hashizume, S.F. Badylak, W.R. Wagner, Mechanical properties and *in vivo* behavior of a biodegradable synthetic polymer microfibre-extracellular matrix hydrogel biohybrid scaffold, *Biomaterials* 32 (13) (2011) 3387–3394.
- [30] I.N. Simões, P. Vale, S. Soker, A. Atala, D. Keller, R. Noiva, S. Carvalho, C. Peleteiro, J.M.S. Cabral, D. Eberli, C.L. da Silva, P.M. Baptista, Acellular urethra bioscaffold: decellularization of whole urethras for tissue engineering applications, *Sci. Rep.* 7 (2017) 41934.
- [31] C.H. Lin, K. Hsia, C.K. Su, C.C. Chen, C.C. Yeh, H. Ma, J.H. Lu, Sonication-assisted method for decellularization of human umbilical artery for small-caliber vascular tissue engineering, *Polymers* 13 (11) (2021) 1699.
- [32] F. Yusof, M. Sha'ban, A. Azhim, Development of decellularized meniscus using closed sonication treatment system: potential scaffolds for orthopedics tissue engineering applications, *Int. J. Nanomed.* 14 (2019) 5491–5502.
- [33] D. Booth, R. Afshari, M. Ghovvati, K. Shariati, R. Sturm, N. Annabi, Advances in 3D bioprinting for urethral tissue reconstruction, *Trends Biotechnol.* 42 (5) (2024) 544–559.
- [34] G.C. da Mata, M.S. Morais, W.P. de Oliveira, M.L. Aguiar, Composition effects on the morphology of PVA/chitosan electrospun nanofibers, *Polymers* 14 (22) (2022) 4856.
- [35] M. Ashoorirad, M. Saviz, A. Fallah, On the electrical properties of collagen macromolecule solutions: role of collagen–water interactions, *J. Mol. Liq.* 300 (2020) 112344.
- [36] S.C. Zhang, C. Campagne, F. Salaün, Influence of solvent selection in the electrospinning process of polycaprolactone, *Appl. Sci.* 9 (3) (2019) 402.
- [37] N. Bhardwaj, S.C. Kundu, Electrospinning: a fascinating fiber fabrication technique, *Biotechnol. Adv.* 28 (3) (2010) 325–347.
- [38] C.J. Angammana, S.H. Jayaram, Analysis of the effects of solution conductivity on electrospinning process and fiber morphology, *IEEE Trans. Ind. Appl.* 47 (3) (2011) 1109–1117.
- [39] S.E. Noriega, G.I. Hasanova, M.J. Schneider, G.F. Larsen, A. Subramanian, Effect of fiber diameter on the spreading, proliferation and differentiation of chondrocytes on electrospun chitosan matrices, *Cells Tissues Organs* 195 (3) (2012) 207–221.
- [40] C.P. Barnes, S.A. Sell, E.D. Boland, D.G. Simpson, G.L. Bowlin, Nanofiber technology: designing the next generation of tissue engineering scaffolds, *Adv. Drug Deliv. Rev.* 59 (14) (2007) 1413–1433.
- [41] J.M. Rhodes, M. Simons, The extracellular matrix and blood vessel formation: not just a scaffold, *J. Cell Mol. Med.* 11 (2) (2007) 176–205.
- [42] G.C. Kodippili, Structural Characterization of Polyurethane Foam and Implications of Aging, University of Nevada, Las Vegas, Master Thesis, 2003.
- [43] V. Moreno-Manzano, D. Zaytseva-Zotova, E. López-Mocholí, Á. Briz-Redón, B. Løkensgard Strand, Á. Serrano-Aroca, Injectable gel form of a decellularized bladder induces adipose-derived stem cell differentiation into smooth muscle cells *in vitro*, *Int. J. Mol. Sci.* 21 (22) (2020) 8608.
- [44] J. Manuel Velazquez Arcos, Real Perspective of Fourier Transforms and Current Developments in Superconductivity, IntechOpen, London, 2021.
- [45] J. Lee, G. Khang, J. Lee, H. Lee, Interaction of different types of cells on polymer surfaces with wettability gradient, *J. Colloid Interface Sci.* 205 (2) (1998) 323–330.
- [46] S.H. Oh, S.G. Kang, E.S. Kim, S.H. Cho, J.H. Lee, Fabrication and characterization of hydrophilic poly(lactic-co-glycolic acid)/poly(vinyl alcohol) blend cell scaffolds by melt-molding particulate-leaching method, *Biomaterials* 24 (22) (2003) 4011–4021.