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Research Article

In *vitro* degradation studies of polylactide/ polyethylene glycol electrospun fiber blends

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ABSTRACT

Biomaterials such as polylactide play a crucial role in tissue engineering by accelerating the structural and functional rejuvenation of damaged tissues. Polylactide is however, hydrophobic and this has limited its use in soft tissue engineering as it prevents cell adhesion and proliferation. Tensile strength and strain at break of polylactide need to gradually degrade during wound healing to enhance cell adhesion. Polyethylene glycol has been used to improve the hydrophilicity of polylactide but In *Vitro* studies involving strength and ductility degradation of electrospun polylactide/ polyethylene glycol fibers under physiological conditions have not been investigated. Pellets of both polymers were dissolved in di-chloromethane in different polylactide/ polyethylene glycol rations of 80/20 and 60/40 (% wt. / wt.); the solution was electrospun into fiber mats. The samples were immersed in phosphate buffered saline/solution maintained at pH7.4 and 37°C for 8 weeks. Fourier Transform Infrared Spectroscopy informs that shifts in functional group wavenumbers with formation of additional peaks on 1906 and 2067 cm^{-1} in the electrospun fibers indicate degradation. There is an increase in average fiber diameter ranging from 1.07 to 13.04 μm, where addition of 40 wt. % polyethylene glycol possesses the greatest magnitude. Weight loss result shows that polymer blends degrade faster than neat polylactide. Within 8 weeks, tensile strength of neat polylactide fiber degrades from 0.28 to 0.08 MPa while that for polylactide/ polyethylene glycol (80/20) fiber occurs from 0.31 to 0.08 MPa; degradation of polylactide/ polyethylene glycol (60/40) is evident as its tensile strength reduces from 0.27 MPa before immersion to 0.04 MPa after 8 weeks of immersion. Ductility reduction of 94 % occurs with polylactide blended with 40 wt. % polyethylene glycol after 8 weeks while 70 and 50 % are lost for polylactide/ polyethylene glycol (80/20) and neat polylactide respectively. The performances of these fiber blends (in comparison with neat Polylactide fiber) inform that they can be good for wound dressing.

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INTRODUCTION

Biomaterials are very important in tissue engineering because they provide biophysical and biochemical stimuli to guide cells for tissue formation. These materials have been confirmed to also hasten the structural and functional rejuvenation of damage tissues [1-2]. Metals/alloys, ceramics and polymers with their composites (a combination of two or more of these different class of materials) have been used over the years for human body tissues. To aid biomaterials' cellular interactions and regulate their functional behaviours (such as mechanical strength), a three-dimensional (3D) porous structural support needs to be provided. A polymeric material that has been proven to be useful for this device is polylactide / polylactic acid (PLA). The material is taking the lead in the emerging biopolymer market owing to its most attractive manufacturing cost, structure and availability. This biopolymer has been in existence for several decades and it can be sourced from corn [3]. Poor hydrophilicity however, has been a major challenge that has limited its use in tissue applications [4]. Scaffolds made of PLA for instance, have been proven to exhibit unfavourable cell adhesion due to its poor hydrophilicity [5] and this hinders effective wound healing. To improve PLA's hydrophilic tendencies, polyethylene glycol (PEG) can be blended with PLA to form PLA/PEG composite. Polyethylene glycol is a hydrophilic oligomer or polymer synthesized from ethylene oxide which is made up of repeating units of –(O-CH_2-CH_2)-. The ability to attach a variety of reactive functional groups to the terminal sites of PEG polymers has greatly expanded their advantages in tissue engineering applications such as drug delivery and wound healing [6]. Fiber mats/scaffolds of PLA/PEG blend can be processed via electrospinning under the application of electrostatic forces. This often occurs at room temperature where interface of the polymer solution is exposed to an electrical force. This culminates in the ejection of a charged linear spray of solution. While the solvent evaporates, the polymer gathers on a collector (usually aluminum plate or foil) as practiced in earlier studies [7].

Scaffolds for small diameter vascular grafts have been produced by electrospinning blended solutions of polyurethane (PU) and PEG [8]. Addition of equal volumes of N,N dimethylformamide (DMF) to tetrahydrofuran (THF) was used in dissolving PU at a concentration of 10% w/v for 12 h. Varying mass of PEG was added to the PU solution to achieve PU/PEG blend ratios of 90/10, 80/20, 70/30, 60/40 and 50/50 prior to electrospinning. The PU/PEG scaffolds possessed high porosity with improved hydrophylicity which heightened as more contents of PEG were added to PU. Polyethylene glycol provided a plasticizing and hardening effect on PU. Mechanical properties of PU/PEG scaffolds were observed to be similar to human and pig arteries. The researchers concluded that blending 20 and 30 wt. % contents of PEG to PU would produce more promising effects in vascular tissue engineering as they improved cell proliferation and attachments. Electrospinning parameters such as melt solution temperature, applied voltage, collector type, and nozzle collector distance were observed on blending varying contents (wt. %) of PEG with PLA [9]. Granules of PLA was first dissolved in Irganox 1010 after which PEG granules were added to produce PLA/PEG blended melts in ratios 95/5, 90/10, 80/20 and 70/30 wt./wt. At a fixed nozzle-collector distance of 10 cm, applied electrospinning voltage was varied between 20-40 kV; spinning temperature ranged between 170-230 ^{o}C. At fixed values of spinning voltage and temperature, nozzle-collector distance was varied at 4, 10 and 14 cm. Blending 20 and 30 wt. % PEG with PLA produced the finest fiber even at low spinning temperature and electrostatic forces. Reduced nozzle-collector distance culminated in bending instabilities of scaffolds. Investigation of Kruse et al. [10] was aimed at processing and characterizing electrospun yarns. Different blends of PLA/PGA were dissolved in chloroform solutions and poured in a 5 ml syringe, electrospun at the rate of 2.5 ml/h while a distance of 23 cm was maintained between the nozzle tip and the collector. Formation of beads in the spun fibers was attributed to the reduced viscosity of PLA/PEG solution caused by addition of more PEG. Tensile strength of the polymer blend was noted to increase with increasing content of PEG. Results of cytocompatibility test of polymer blends with endothelial cells showed that the electrospun fibers were cytocompatible. To hasten healing in slow or non-healing wounds, platelet-rich growth factor (PRGF, derived from umbilical cold blood) with gelatin fibers were electrospun with poly-L-lactide (PLLA) to form a three layered scaffold [11]. The PLLA nanofiber was made the outer layer while gelatin/PRGF) was made the inner layer. In *vitro* study showed that healing efficacy of PRGF could be achieved (up to 3-5 days) when made to be released from scaffold. Their research maintained that PRGF would not display healing attributes when used alone but until when released from the electrospun fibers. Nanofibers comprising poly (γ-glutamic acid) (γ-PGA) as the core and PLA as the shell material have been produced via coaxial electrospinning [12] for wound healing. The core-shell membrane possessed favourable hydrophobic properties (contact angle at 90^{o}) which enhanced the maintenance of its morphology during degradation. In *vitro* studies showed that the scaffolds displayed good biocompatibility and supported cell proliferations. Hajikhani et al. [13] employed coaxial electrospinning method to fabricate PLA- polyethylene oxide/ Polyvinylpyrrolidone (PLA-PEO/PVP) complex nanofibers which contained collagen and cefazolin as a wound dressing material. Cefazoline, which served as an antimicrobial agent and core was encased in PLA (shell); collagen was included in the nanofiber shell to quicken healing process while PVP- a super-hydrophilic polymer elevated the release of collagen. Results showed that the scaffold fended off the growth of microorganisms such as Escherichia coli, Staphylococcus aureus, and Pseudomonas aeruginosa. Lopresti et al. [14] focussed on the chemical, antimicrobial and physical features of electrospun PLA membranes loaded with carvacrol (CRV) and a commercial nisin formulation (Nis) antimicrobial agents. Mechanical characterization showed that the CRV acted as a plasticizer on PLA while Nis made the membranes brittle as their thermal properties remained unaltered. Carvacrol hindered fiber slipping under tension and elevated its elastic properties by 508 %; Nis on the other hand, also contributed positively to membranes' elastic modulus by 100 % as it acted as solid filler. Nisin formulation made the membranes which was initially hydrophobic, hydrophilic. Nanofibers of PLA have been altered on blending with poly(ethylene glycol)-poly(propyl glycol)-poly(ethylene glycol) (PEG/PPG/PEG) copolymers [15]. The copolymer possessed good water stability within 2 months and this enhanced the water retaining capacity of PLA nanofiber. Increasing the content of PEG/PPG/PEG reduced the porosity of electrospun membrane. These features made the researchers conclude that the hybrid polymer fiber would be good for biomedical applications. Polylactide sourced from polysaccharides of -*Bletilla striata* and *Rosmarini*c acid were electrospun by Zhong et al. [16]. In addition to the polymer fiber's ability to encourage cell proliferation, its tensile properties were characterized with the maximum flexibility. The role of PEG in the enhancement of biological and structural modifications of PLA and chitosan membranes has been studied [17]. A hybrid of the three electrospun polymers showed a moderate rate of degradation, reduced hydrophobicity compared to unblended PEG fiber.

These studies have shown that PLA and PEG are both fit for human tissues (most especially as drug carriers) and a successful strategy for increasing hydrophilicity of PLA is by blending with PEG. Few works however, have been investigated on how this electrospun polymer fiber blend behaves when exposed to buffered solution under physiological conditions. Tensile strength and strain at break of electrospun fibers play key role in their ability to serve as potential sites for cell adhesion during wound healing; these properties need to gradually reduce (with time) for this to be achieved. The strength and ductility degradation of electrospun PLA/PEG fibers in PBS has not been studied. This research thus focuses on the degradation studies of electrospun fibers comprising blends of hydrophobic PLA and hydrophilic PEG in a water-based salt solution at pH 7.4 and 37^{o}C.

MATERIALS AND METHODS

Electrospinning of PLA and PLA/PEG Blended Solutions

Corn starch-sourced polylactide with average molecular weight of 250,000 g/mol was obtained from NATUREWORKS while PEG 6000 was purchased from LOBA CHEMIE PVT. LTD.A total of 4 g pellets (PLA with PEG) was dissolved in 20 ml of 95 % purity dichloromethane (DCM) and mixed on a magnetic stirrer at room temperature until a complete dissolution and homogenous viscous liquid was obtained. The polymer solution (electrospinning solution) was fed into a spinneret (60 ml syringe was used) at room temperature and connected to a voltage source (20 kV). The spinneret was positioned vertically downward towards a stationary aluminum foil fiber collector (20 cm x 20 cm) kept 40 mm away from the tip of a 1.4 mm needle diameter fixed at the end of its (spinneret) orifice. The schematic diagram for the PLA/PEG fiber blend spinning from solution is shown in Figure 1. To arrive at the desired content (wt. %) of each combining polymer (PLA and PEG), the mixing ratios considered are displayed in Table 1.

Preparation of Phosphate Buffered Saline/Solution (Pbs)

This solution is widely used in the manufacturing of cell culture owing to its non-toxic nature (to living cells). As a result of this, it is often used for In *Vitro* studies [18-19]. Dibasic sodium phosphate, Na_2HPO_4 (9.47g) was dissolved in 1 L distilled water. A second reagent, mono basic potassium hydrogen phosphate, KH_2PO_4 (9.08g) was dissolved in 1 L distilled water. The mixture of the two solutions was achieved by adding 161 ml Na_2HPO_4 solution with 39 ml KH_2PO_4 solution and maintained at pH 7.4 and 37 ^{o}C.

Degradation Test

Electrospun fibers were fully soaked in PBS (maintained at pH 7.4 and 37 ^{o}C) for 8 weeks. At the end of each week, each fiber was weighed, oven dried at 50 ^{o}C for 12 h and characterized.

CHARACTERIZATIONS

Functional Groups Determination

Polymer functional groups were determined via Fourier Transform Infrared spectroscopy (FTIR) technique with the use of a PerkinElmer UATR Two spectrometer. Measurement was taken for each polymer in transmittance

Table 1. PLA/PEG blend formulations

S/N	PLA (g)	PEG (g)	PLA/PEG Ratio (wt. %)	Total mass (g)
1	4	0	100	4
2	3.2	0.8	80/20	4
3	2.4	1.6	60/40	4

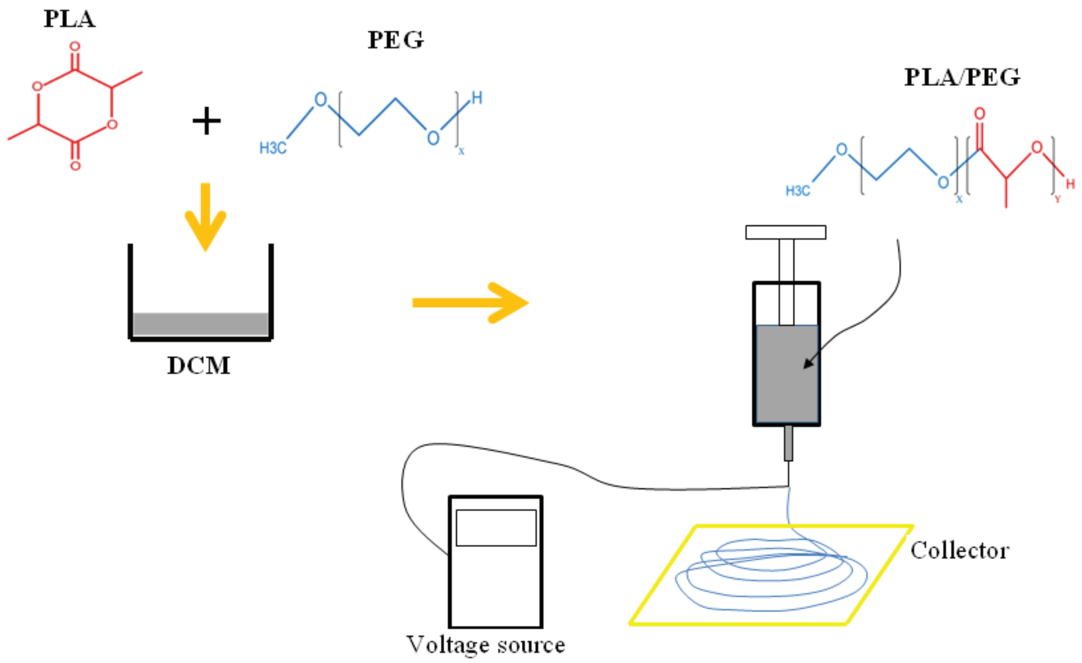

Figure 1. Schematic diagram for PLA/PEG fiber blend spinning from solution.

mode between 500–4000 cm^{-1}. To avoid shifts in baseline, normalisation of spectra were done using OriginLab software.

Scanning Electron Microscope (SEM)

A VEGA 3 TESCAN model variable-pressure scanning electron microscope was used to observe the morphological features of electrospun polymer fibers. Each electrospun fiber was placed on a circular disc lined with carbon and coated with Au for 5 min for electric conduction. Average fiber diameters and pore sizes of samples were determined with the use of Image J software.

Tensile Test

The tensile test of electrospun fibers was determined using the Instron 3369M Tensometer Each fiber was fixed and held firmly at both ends by the gauge as load was applied until the fiber finally failed. This test was done three times for each fiber and the average results were taken. The device recorded maximum stress and final length attained by each fibre prior to fracture; this provided information on the tensile strength (UTS) and ductility of the fibers. Ductility was calculated using equation (1):

$$\text{Ductility (\%)} = (\text{Extension/Initial fibre length}) \times 100 \quad (1)$$

Extension is the difference between the final and initial fibre length.

Weight Loss

After each period of immersion, each electrospun fiber was oven dried at 50 °C for 12 h. The percentage weight loss was used as a measure of degradation. This was calculated using equation (2):

$$W\% = \frac{W_o - W_f}{W_o} x\ 100 \quad (2)$$

Where W_o and W_f are the initial and final dry weights of the fiber.

RESULTS AND DISCUSSION

Polymer Functional Groups

The FTIR spectrum of PEG electrospun fiber is shown in Figure 2a. The OH functional group is found on 3454 cm^{-1}while C-H stretching is represented by 2861 cm^{-1} [20]. Spectra on wavenumbers 1466 and 1334 cm^{-1} are attributed to C-H bending. Deformations of C-H and O-H groups are located on 1243, 1215, 1104 and 937 cm^{-1} [21-22]. Carbonyl group (C=O) stretching of neat electrospun PLA fiber (Fig. 2a) is found on 1724 cm^{-1} and this is ascribed to the vibration of amorphous carbonyl present in the structure; wavenumbers 1459 and 1362 cm^{-1} represent asymmetric and symmetric modes of CH_3 respectively. At 1048 and 1187 cm^{-1}, the bands representing the asymmetric C-O absorption modes exist. In addition, asymmetric and symmetric vibrations also exist on 2993 and 2937 cm^{-1} respectively. Spectra of PLA identified in this study are comparable to the investigations of Singla et al. [23] and Leonés et al. [24]. The FTIR spectra of the blended electrospun fibers (Fig. 2b and c) show that there is a proper mix between the two combining material (PEG and PLA) as PEG functional groups (CH bending) is found at 1449 cm^{-1} and 1359 cm^{-1}.

It is also observed that there exists some similarities in the functional groups between PLA and PEG and this may be responsible for a homogenous mixture each solution blend presented prior to spinning. Degradation of neat PLA fiber after 8 weeks is exemplified by shift in peak at every wavenumber (Fig. 2a). Although each spectrum looks similar, there lie additional peaks at 1906 and 2067 cm^{-1} on the degraded PLA which also indicates that reaction took place during immersion. Reduction/increase in spectra peak intensities have been used to describe degradation of PLA [25,26]. There also exists wavenumber shifts in fiber blends (Fig. 2b and c) with reduced intensities; for instance, the C=O group existing on 1745 cm^{-1} in 80PLA/20PEG shifts to 1738 cm-^{1} (Fig. 2b). The disappearance of C-H functional group at 2866 cm $^{-1}$ in both blends after the 8th week of immersion may imply PEG dissolution. In addition to peak shifts and reduced intensities, degradation of PLA can also be justified by the existence of new peaks at 1906 and 2067 cm^{-1} as exhibited by PLA spectrum after immersion, (Fig. 2a)

Morphological Investigation

Figure 3a-b shows the morphologies of electrospun neat PLA before and after 8 weeks of immersion in PBS. Both structures display cylindrical fibers with few bead formations (indicated in Figure 3) which must have occurred as a result of clogging of PLA solution on the aluminum collector. This is less after 8 weeks soaking (Figure 3b) which implies possible degradation of some PLA beads in PBS. In addition, fiber average diameter of neat PLA increases from 1.07 to 1.35 µm after immersion, which indicates diffusion of PBS solution into PLA's structure. This has also led to an increase in average pore size from 34.3 to 37.4µm. Researchers have reported that there is a linear relationship between fiber diameter and pore size [27]. They concluded that pore size of electrospun fiber mats will increase with the increase in fiber diameter. Moisture (in the air) condensation on the fiber surface was suggested to be a likely mechanism for pores formation on electrospun fiber mats [28].

Increase in average pore size and fiber diameter is observed when 80 wt. % PLA is blended with 20 wt. % PEG (Figure 4a). Before immersion, average pore size of 39.5 µm is calculated while the average fiber diameter is observed to be 2.04 µm. The increase in average fiber diameter is an indication of homogenous mixture between the two polymers during blending. Pores formation is a key feature needed by scaffolds for cell attachments in tissue engineering. It can therefore be said that addition of PEG has helped improve this in PLA. Comparing this to neat PLA fiber (Figure 3b), 7.39 and 45.9 µm are measured for average fiber diameter and pores size respectively. Absorption of PBS is effectively manifested by PEG which may make PLA lose its strength after 8 weeks immersion. Creation of more pores may also be responsible for this, including its weight loss as will be observed in Figure 6.

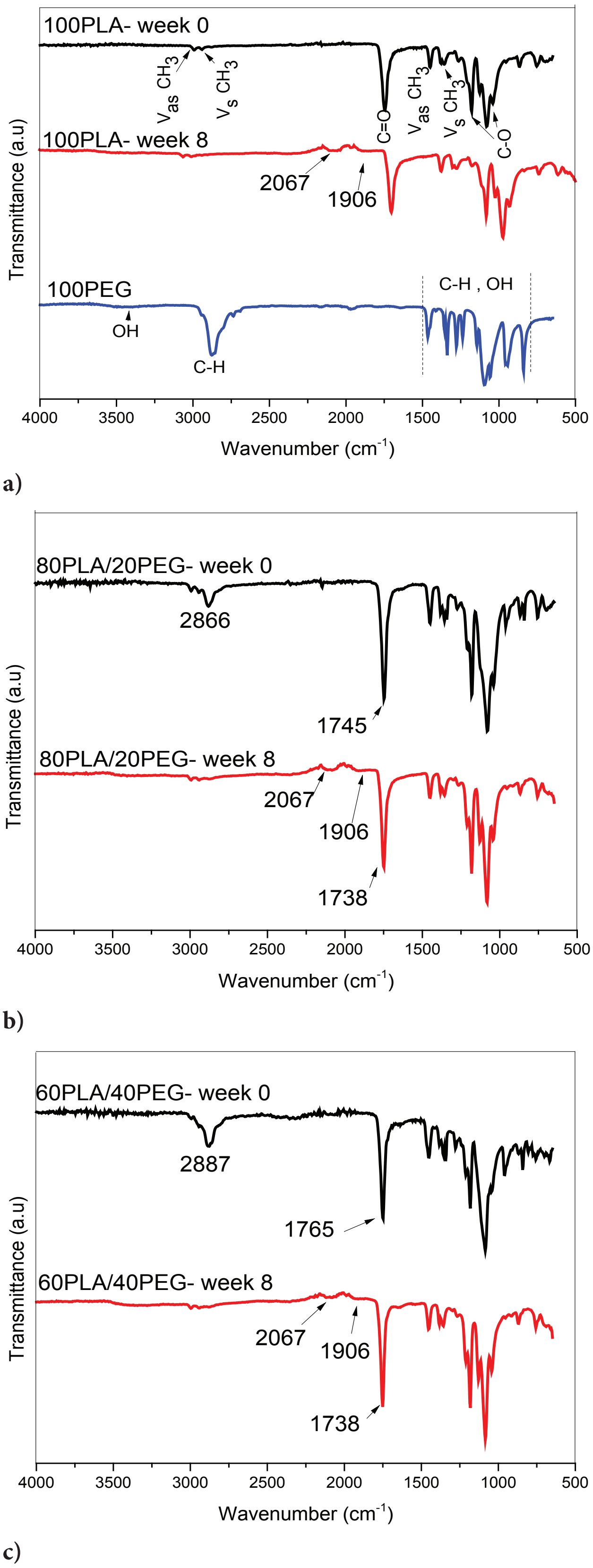

Figure 2. FTIR spectra of (a) 100PLA fiber before(week 0) and after 8 weeks of immersion; 100PEG fiber (b) 80PLA/20PEG fiber (c) 60PLA/40PEG fiber before (week 0) and after 8 weeks of immersion.

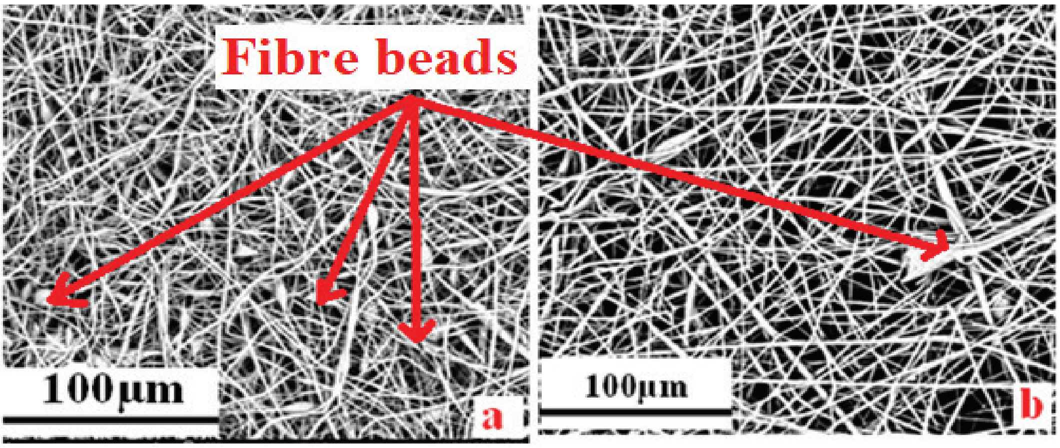

Figure 3. SEM of neat PLA fiber (a) before immersion (b) after 8 weeks immersion in PBS.

A further increase in average fiber diameter of 2.99 µm is measured for 60PLA/40PEG electrospun fiber blend before degradation test (Fig. 5a). After 8 weeks of immersion; the average fiber diameter is raised to 13.04 µm, which is the maximum for all fiber mats processed in this study (Figure 5b). Fiber diameter is uniform (with smooth surface) for the electrospun blend before and after degradation study in PBS. This is an indication that the presence of PEG is instrumental to the formation of a porous structure. Highly porous 3D structures with large surface to volume ratios have been reported to play a key role in antibacterial surfaces [29], scaffolds [30], and wound dressing applications. This makes the production of a porous structure a necessity in biomedical engineering. Degradation of each material is exemplified by increase in fiber diameter after immersion in PBS. This proposition agrees with the findings of Dias et al. [31], who ascribed the diameter increase of their processed electrospun PLA mats after immersion in PBS to hydrolytic degradation. Swelling of PLA/PCL membranes after immersion in distilled water as a result of improved hydrophilicity was characterized by fiber diameter increase [32]. According to the researchers, the diameter increase (as a result of swelling) facilitated cell adhesion and proliferation. A hydrophilic polymer enables solution penetrates through its matrix, which in turn hastens its degradation [33-34]. Asides promoting cell proliferation, enhanced hydrophilicity can culminate in protein adhesion on to the surface of PLA [35], which boosts immunogenicity [36].

Weight Loss in PBS

Figure 6 shows the degradation profiles of LA and PLA/PEG fiber blends in PBS at 37 °C for 8 weeks. The Figure explains that addition of PEG to PLA increases the weight of electrospun PLA/PEG fibers. There is a gradual depreciation in each fiber's weight up to the final time of soaking. This implies that there is a chemical reaction between each polymer and PBS, which leads to loss of weight after drying. Electrospun PLA displays a constant

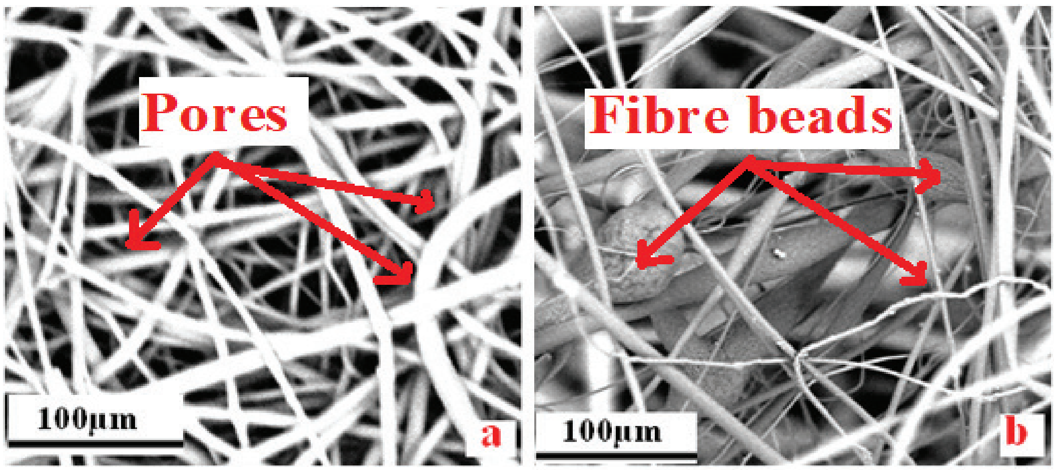

Figure 4. SEM of 80PLA/20PEG fiber: (a) before immersion (b) after 8 weeks immersion in PBS.

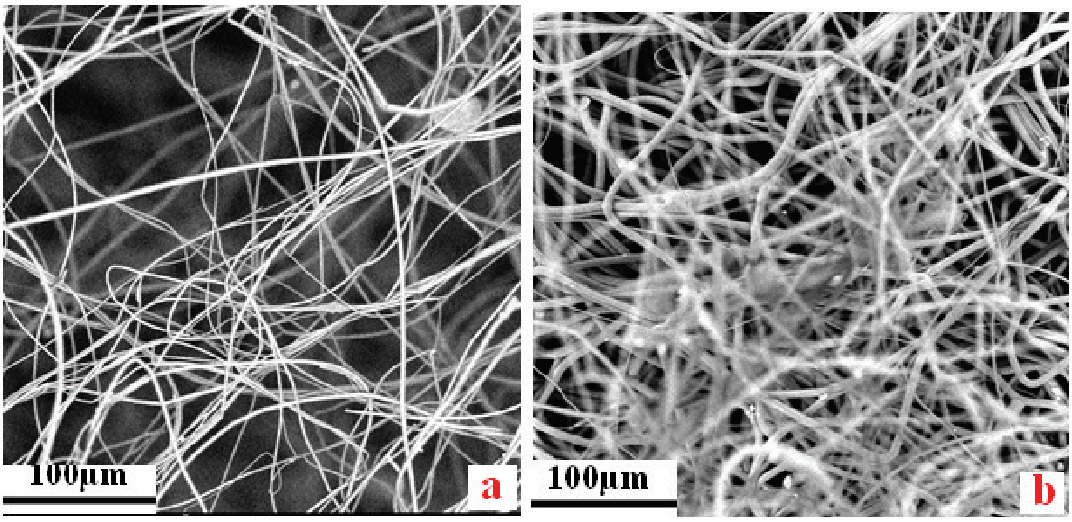

Figure 5. SEM of 60PLA/40PEG fiber: (a) before immersion (b) after 8 weeks immersion in PBS.

weight loss of 0.93 g for 4 weeks after which a gradual loss of weight is recorded for the next 4 weeks. Unlike PLA, blends of 80PLA/20PEG and 60PLA/40 PEG degradation with respect to dry weight loss become obvious after the third week of immersion. This shows that addition of PEG hastens the degradation of PLA. Hydrolytic degradation (surface erosion) of PLA fibres, according to Vaid et al. [37] is fast when the aqueous solution is basic. Reduction in surface changes and fiber characteristics becomes very little as solution pH reduces from 7.4 to 2; this is typical of bulk erosion which is exemplified by a slower degradation. Density functional theory (DFT) calculations employed by the authors affirms that electron withdrawing groups develop positive center at the carbon in carbonyl (C=O) functional group, that ends up creating a nucleophilic attack site by OH^-. Molecular dynamics (MD) simulations also supports the increase in PLA fibre degradation rates in solutions with higher pH values. Degradability and properties of PLA is significantly influenced by hydrolysis while non-catalytic hydrolysis of the polymer entails ester bond cleavage on exposure to water [38]. The researchers confirmed that depending on temperature, highly crystalline PLA exhibits a slower hydrolysis rate compared to amorphous and partially crystalline ones. Vaid et al. [37] however reported that crystallinity of PLA fiber is enhanced by fibre spinning. The hydrolytic degradation produces shorter chains with acid and alcohol terminal groups. The hydrophobic feature of PLA is evident in its degradation profile as its measured dry weight after each week of immersion looks comparable; there is no much difference between the final dry weight (after the eight week) and the dry weight of un immersed electrospun PLA fiber. Plots of 80PLA/20PEG and 60PLA/40 PEG show a much steep profile, indicating fast degradation rate. The physicochemical properties of PLA has been modified by addition of hydrophilic PEG hence, the blends can be useful for tissue engineering applications. Results obtained in this study agree with the works of Hendrick and Frey [4] whose research was aimed at improving the surface hydrophilicity of PLA by addition of PEG homo polymer and PLA-b-PEG co-polymers. Hydrophilicity of PLA was found to be enhanced on addition of PEG homo polymer and PLA-b-PEG co-polymers. Greater increase in water wettability was noticed with PLA-b-PEG co-polymers than PEG homopolymers. Medicated PEG/PLA nanofibrous membranes have been used as a carrier of an antibiotic drug, tetracycline hydrochloride, TCH [39]. The medicated PEG-PLA nanofibers exhibited mechanical barriers which aided the delivery of antibiotics in a sustainable way.

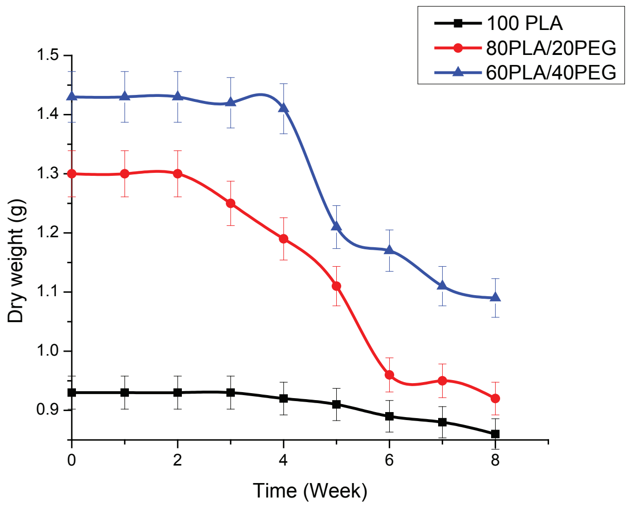

Figure 6. Dry weight loss of PLA and PLA/PEG electrospun fiber blends with time.

Tensile Properties

The maximum strengths of electrospun fibers attained prior to fracture is shown in Figure 7 Prior to immersion, it is observed that 80PLA/20PEG fiber has the greatest magnitude of UTS (0.31 MPa) which is approximately 11% increase from neat PLA fiber (0.27 MPa). An increase in UTS (0.29 MPa) is also noticed when 60 wt. % PLA is blended with 40 wt. % PEG; here, the UTS increase is minimal (approximately 4%). This implies that further additions of PEG to PLA may lead to reduction in UTS. Investigation of Kruse et al. [10] was aimed at processing and characterizing electrospun yarns. Tensile strength of the polymer blend was noted to increase with increasing content of PEG; this was attributed to the reduced viscosity of PLA/PEG solution caused by addition of more PEG. Toncheva et al. [40] produced PLA/PEG electrospun fibers in two ways for mechanical, thermal and biological properties comparisons. The first method was to physically blend PEG to PLA while chemically grafting PEG on PLA entailed the second processing technique. Physical blending and chemical grafting techniques did not modify the wettability of PLA but provided a plasticizing effect, which was more evident in the PLA/PEG blended mats. This was attributed to more of PEG mobility possessed in the blend. The plasticizing effect of PEG used in this study could be responsible for the strength of electrospun PLA fiber; when this effect is much, it may lower the strength. After the first week of immersion, UTS of electrospun polymer fiber blends become lower than neat PLA. This shows that PEG can improve strength of PLA and also, aid its degradation (provided the content of PEG is controlled). Materials such as these can be useful for soft tissues, where considerable gradual depreciation of strength is required for effective healing.

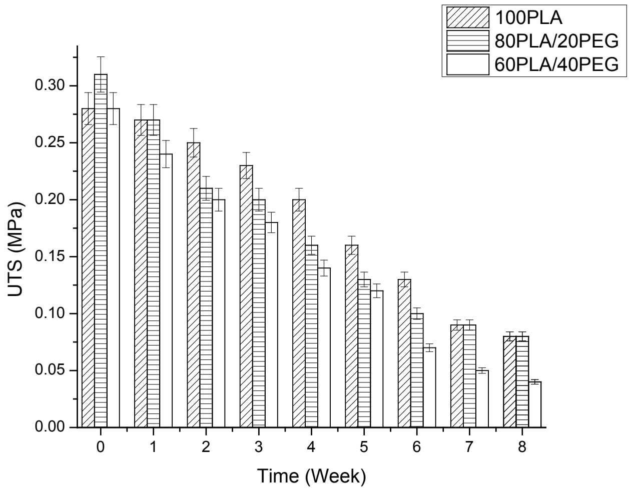

Figure 7. UTS of PLA and PLA/PEG blends with immersion time.

Effect of the PEG on the ductility of PLA is shown in Figure 8. Ductility of electrospun PLA/ PEG blends are higher than neat PLA which is calculated to be1.18% before immersion in PBS. Addition of PEG has thus improved the elastic properties of PLA. Firoozian et all. [41] reported that the ductility of a composite will reduce if its elastic property is dominated by the matrix. Suradi, et al [42] further proved the position that reduced ductility will happen if the volume fraction of the matrix, which is more elastic of the two, reduces. Result from this study reveals that PEG is more ductile than PLA. Blending 60 wt. % PLA with 40 wt. % PEG displays the highest magnitude of ductility before (1.8 %) and at the end of each immersion period. In a bid to evaluate the lifespan of compression-moulded PLA as a structural implant in the human body, Woo and Wee [43] devised a model where its degradation was made to accelerate in PBS at 50, 60 and 70 °C. Degradation was evaluated in terms of its tensile properties (such as tensile strength and strain at break), which was discovered to reduce as time of exposure (in PBS) lapsed. At 50 °C, noticeable decay in the tensile properties commenced after 28 days while this occurred from the 6th and 2nd day at 60 and 70 °C respectively. In this study, it can be inferred that diffusion of fluid (PBS) into fibre's structure as evidenced by its swelling, must have culminated in ester bond cleavage as earlier reported by [38]; this could also be responsible for its ductility loss.

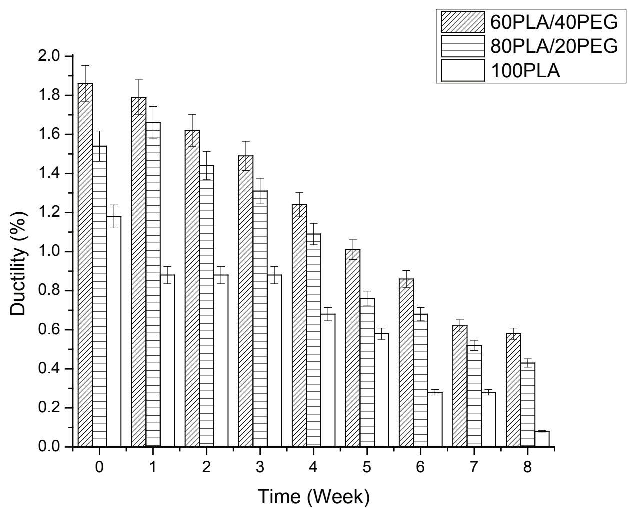

Figure 8. Ductility of PLA and PLA/PEG blends with immersion time.

CONCLUSION

In this study, the degradation of electrospun polylactide and polylactide/ polyethylene glycol fiber blends in phosphate buffered saline/solution at 37 °C and pH 7.4 have been studied. Results from Fourier Transform Infrared Spectroscopy justify that 100 % miscibility exists between polylactide and polyethylene. Degradation is exemplified by shifts in functional group wavenumbers. Formation of additional peaks on 1906 and 2067 cm^{-1} also depicts degradation, including the disappearance of C-H functional group at 2866 cm $^{-1}$ in the fiber blends. Presence of polyethylene glycol is responsible for increase in average fiber diameter and this also widens after 8 weeks of immersion in phosphate buffered saline/solution. Before and after immersion, average fiber diameter increase of 1.07-1.35 µm, 2.04-7.39 µm and 2.99-13.04 µm are measured for neat polylactide, polylactide/ polyethylene glycol (80/20) and (60/40) electrospun fibers respectively. This implies that each fiber expands/swells as it absorbs the solution; Electrospun polylactide fiber maintains a constant dry weight loss for 4 weeks after which a gradual loss is recorded for the remaining 4 weeks. On the other hand, it takes 3 weeks for polylactide/ polyethylene glycol (80/20) and (60/40) electrospun fiber blends to start exhibiting considerable weight loss. After 8 weeks, polylactide/ polyethylene glycol (60/40) fiber loses 23 % of its weight while (80/20) variant loses 30 %; final weight of neat polylactide reduces by 5.3 %. This shows that addition of Polyethylene glycol hastens the degradation of Polylactide. Blending polylactide with polyethylene glycol (40 wt. %) displays the highest magnitude of ductility loss at the end of each immersion period; The electrospun fiber blends can be a good wound dresser as they possess the potential to effectively absorb fluids that are often secreted by wounds prior to complete healing; they can also be used as scaffolds for tissue regeneration. The improved tensile strength and ductility degradation of fibers (compared to neat polylactide fiber) will make them suitable for temporary structural supporting devices for soft tissues. To allow proper healing, these fiber blends will gradually lose their mechanical strengths in a faster way compared to unblended polylactide. Evaluation of degradation products and In *Vivo* studies will be conducted on the electrospun fibers in our future work.

AUTHOR CONTRIBUTIONS

Gbenebor O.P.: Conceptualization, investigation, methodology, data analysis, writing an original draft, and reviewing; **Popoola A.P.I.**: Investigation, methodology, editing and funding acquisition. Authors have read and agreed to the published version of the manuscript.

DATA AVAILABILITY STATEMENT

The authors confirm that the data that supports the findings of this study are available within the article. Raw data that support the finding of this study are available from the corresponding author, upon reasonable request.

CONFLICT OF INTEREST

The author declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

ETHICS

There are no ethical issues with the publication of this manuscript.

STATEMENT ON THE USE OF ARTIFICIAL INTELLIGENCE

Artificial intelligence was not used in the preparation of the article.

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