Design and Selection of Anti-bacterial Polymeric Surgical Sutures

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ABSTRACT

In this work, a formulation was designed to produce multifunctional polymer surgical sutures based on polypropylene (PP) to utilize its mechanical properties. The resulting sutures are characterized by their antibacterial properties, therapeutic release capacity, and biodegradability.

A specialized extruder was designed for the production of these sutures, consisting of an extrusion chamber, a piston, and a water bath for cooling the resulting sutures.

Initially, mixtures of polypropylene and polylactic acid were prepared in three ratios. The results showed that the 1:1 mixture was the best due to its superior mechanical properties and wetting ability. However, due to the incompatibility of the two polymers, a compatibility factor was added in three ratios, and the optimal ratio, 2%, was selected. To accelerate wound healing and improve compatibility, collagen was added in three ratios, and the ratio 5% provided the best balance between mechanical properties, compatibility, and wetting ability. Chitosan was added in three proportions to help stop bleeding and impart antibacterial properties. Polyethylene glycol (PEG) was chosen as the superior of the three proportions because it gave the surgical suture sufficient elongation and ductility for suturing. To enhance the antimicrobial activity, ZnO nanopartilce was added.

Amoxicillin as an antibiotic drug with a broad antibacterial spectrum was loaded. Results proved that drug-loaded suture exhibits superior antibacterial efficiency, highlighting its potential for use in antimicrobial surgical sutures to reduce postoperative infections. Gram-negative bacteria (E. coli) exhibit relatively higher resistance than Gram-positive bacteria (S. aureus) due to the presence of an outer membrane barrier in their cells.

MTT assay results showed that the cytotoxic effect of the coated suture against the human breast cancer cell line (MCF-7) clear signs of cytotoxic damage including cell shrinkage, reduced cell density, membrane irregularities, and partial detachment from the culture surface.

The biocompatibility was evaluated on the normal epithelial cell line (MCF-10). Results demonstrated that the treated cells maintained high viability after exposure to coated suture, indicating that current suture formula exhibited no significant cytotoxic effect on MCF-10 cells.

The photocatalytic activity of the drug-loaded sutures was evaluated through the degradation of methylene blue (MB) under irradiation. After 180 min of irradiation, the degradation efficiency of drug-loaded suture was about 77%, meaning that the loaded suture is quite photo catalytic.

KEYWORDS: *polymeric sutures, knot security, antimicrobial action, mmt assay, photocatalytic activity*

1 INTRODUCTION

One of the biggest challenges within the medical practice is the innovation and improvements in technologies for the closure of wounds or sutures [1]. The general technologies comprise physically perforating materials, for example, staples or sutures. These approaches have several limitations and challenges such as the risk of infections [2], cause continues pain, not always effective and in some cases can result in leakage at the site of closure. To overcome these challenges and limitations polymer-based sutures can be employed to hold body tissues together or ligate blood vessels, after a surgery or accidental injury [3]. Indeed, polymers, natural or synthetic, absorbable or nonabsorbable, have significant potential and, over the years, through the improvement of materials, such as changes in composition, surface alteration and polymer blend, several sutures have been created with excellent physical and mechanical properties [4].

Polypropylene (PP) suggested as a matrix in this formula due to its high strength and good biocompatibility. Because of its lack of adherence to tissue, it is an excellent “pull-out” suture. PP is hydrophobic and has good resistance to many chemical solvents and acids. [5], but using PP alone as a suture faced many difficult. PP is unabsorbable and needs to remove after healing. Also, PP may cause inflammation or infection to some patients due to elicit a foreign body reaction. In high-tension areas, such as scalp and tendons, PP is difficult to be used due to its low friction properties, which lead to knot slippage. Because of its stiffness and low pliable, PP have some challenges to manipulate during suturing. To overcome these shortcomings, it is expected that PP needs to be blended with flexible, biodegradable polymer, such as PLA, which can over time break down within the body. The presence of methyl group in PLA causes it to be hydrophobic with slow hydrolysis rate [6]. PLA, also, improves the flexibility of surgical suture while maintaining the strength of PP matrix which resulted in easier handling and suturing. PLA have low inflammatory response; improving the healing process as well as reducing any complications may associated with suture placement. With PLA, it is easily to control the degradation rate by adjusting PLA-to-PP ratios. PLA addition can allow to the new suture to be used in the high tension areas, due to enhancing the knot security which reduces the slippages risks. Although PLA addition expected to introduce many improvements, but PLA/PP blends will still suffer from many drawbacks, such as the poor inter-facial adhesion and variation in their degradation rates. These will affect the overall mechanical performance, creating complication in manufacturing processes, incomplete absorption resulting in inconsistent product quality, which may not be suitable for some clinical settings. Therefore, Maleic Anhydride Grafted Polypropylene (PP-g- MA) compatibilizer can reduces the PP/PLA phase separation, brittleness, improves the interfacial adhesion, flexibility, elasticity, toughness and the overall performance. It is available, cost-effective, can be incorporate easily during the blending process and can improves the dispersion of the polar Amoxicillin drug within the polymer matrix, leading to a more uniform drug distribution [7].

Designing of polymeric surgical suture considered a multidisciplinary optimization puzzle balancing materials science, microbiology, pharmacology, and biomechanics. The

selection process needs many requirements, such as choosing the right agent/suture for the specific surgery and patient risk, balancing antibacterial power with cell toxicity, achieving sustained controlled release, preventing antibiotic resistance, maintaining mechanical and handling properties, synchronizing degradation with drug release and so on.

The future lies in smart designs like sutures that release agents only in response to infection biomarkers or those using novel, resistance-proof antimicrobial peptides.

The current work attempts to design a formulation for a surgical polymer suture that meets the above requirements, such as good mechanical properties, bacterial resistance, and sustained drug release, in addition to providing patient comfort and ease of surgical procedure.

2 MATERIALS

PP was obtained from Al-Mohand Iranian Company with density of 0.946 g/cm^3 and 1472 MPa Elastic modulus. Milky white granular PLA was obtained from Shanghai Macklin Biochemical Technology Company (China), with molecular weight of 80000 g/mol and 3 mm particle size. Light yellow Collagen was obtained from Shaanxi Sai yang Food Company (China) with 0.33 g/ml bulk density. Off White Chitosan powder was obtained from Shanghai Macklin Biochemical Technology Company (China), with molecular weight of 500000 g/mol. PP-g-MA was obtained from Shaanxi Sai yang Food Company (China), with 1.5 wt.% grafting level and melting point of 160-165 ºC. Zinc oxide was obtained from Shanghai Aladdin Company (China), with particle size less than 100 nm. TiO_2 was obtained from Shanghai Aladdin Company (China), with melting point of 1850 ℃. White to off-white Amoxicillin drug was obtained from local pharmacies, with melting point of 194–196 ℃.

2.1 The locally designed extrude

A locally designed extrusion system was utilized to fabricate polymeric surgical sutures (Figure 1). The device operates on a piston-based extrusion mechanism, where the polymer blend is fed through a hopper into a heated cylinder, transformed into a viscous melt, and subsequently forced through a shaping die to form a continuous filament. The extruded filament is then cooled in a water bath, drawn, and collected via a motor-controlled winding unit. The system comprises essential components including a hopper, piston, heated barrel, die nozzle, cooling unit, and pneumatic control system. Key processing parameters such as barrel temperature, drawing speed, and cooling conditions were carefully optimized to ensure uniform filament quality and reproducibility. Post-extrusion handling was conducted under controlled conditions to preserve the physicochemical integrity of the sutures.

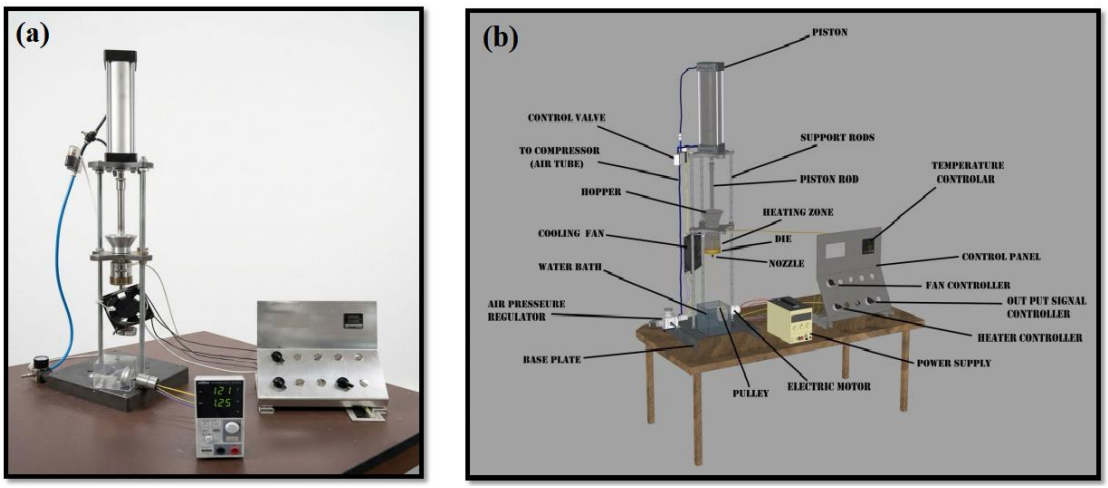

Figure 1: The locally manufactured surgical suture extruder (a) photo and (b) 3D AutoCAD model

2.2 Results analysis

The first goal is to select the desired amount of PLA that be added to PP in terms of results of mechanical properties and wettability, therefore 75% PP: 25% PLA, 50% PP: 50% PLA and 25% PP: 75% PLA blends were tested besides the pure PP suture (Table 1).

The contact angle (CA) is a function of wettability, which relates to suture biocompatibility and its tissue interaction. If CA is high (>90°) the material is hydrophobic (water-repelling), while CA lower than 90° the material be hydrophilic (water-attracting). For surgical sutures, moderate hydrophobicity is often preferred to minimize intense inflammatory reactions.

Tensile strength is critical property for surgical suture, since it determines the force, the suture can withstand before breaking. Maximum deformation and Elongation (TE%) properties are a function of the suture's ductility (how much it can stretch before breaking). A balance is needed, where if it is too low it will be brittle and if it is too high it might not hold tissue securely during suturing. The Young's modulus (E) is a measure of suture stiffness, where a very high modulus means the suture is stiff and can be difficult to handle while a very low modulus refers to too flexible suture.

Table 1: PP/PLA blends properties

PP (wt.%)	PLA (wt.%)	D (mm)	CA (º)	TS (MPa)	Max. Deform. (mm)	TE (%)	E (GPa)
100	0	0.22	125.066	20	581	233	0.66
75	25	0.26	62.769	24	53	21	0.79
50	50	0.21	58.906	27	66	27	2.6
25	75	0.3	41.644	17	20	8.5	0.45

CA results proved that when a water droplet (1.84 D dipole moment) sits on PP (no dipole with 0 D dipole moment), there are no significant dipole-dipole interactions. There is only very weak dispersion forces act, leading to poor wetting and a high contact angle (CA = 125.066°). The water beads up to minimize contact with the low-energy, non-polar surface. CA decreased as PLA portion increased in PP/PLA blends (from 125.066º down to 41.644º).

This is because of high immiscibility between these two polymers, where PLA is inherently more hydrophilic than PP due to ester polar groups (-O-CO-) present in PLA polymer. Indeed, PP contains only nonpolar C-C and C-H bonds where the electron cloud is evenly distributed within these bonds. In cotrast, there is a clear difference in electronegativity along C-O bond within ester group, where Oxygen attracts electrons more strongly, resulting in partial negative (δ-) and partial positive (δ+) charges. Since water molecules are also highly polar, the polar groups on PLA can arises dipole-dipole interactions with water molecules (Like dissolves like), where the dipole moment of ester groups in PLA is 1.5 - 2 D that can energetically "handshake" with the polar water molecules.

Within PLA, generally the bond polarity is high because both C=O (carbonyl) and C-O bonds are highly polar due to the large electronegativity difference between Carbon and Oxygen, which is about 3.44. In PP, the bond polarity is very weak because the electronegativity difference between Carbon (2.55) and Hydrogen (2.20) is very small. This makes the C-H bond only very slightly polar and PP have no dipole moment ; thus cannot "handshake" with water molecules.

This increment in wettability is desired for surgical sutures, where it gives indicators about interactions with the surrounding biological fluids within the human body as well as indicator to its biodegradability.

Results showed that, pure PP suture has extremely high ductility, suffer from low tensile strength, too elasticity and highly hydrophobicity. In contrast, suture composed of (1:1) PP: PLA:

- Has the highest tensile strength (27 MPa), which is a significant advantage for suturing application.
- Offers a good, moderate elongation (27%), suggesting it can absorb some stress without being overly elastic or brittle. On the other hand, formulas containing 25% or 75% PLA have low ductility (21% and 8.5%), making them relatively brittle, while pure PP is extremely ductile (581% elongation), which is unusual and likely too elastic for most surgical applications.
- Has a very high modulus (2.6 GPa), indicating it is quite stiff. While this could be drawback for handling, it is acceptable if the strength is high. The other samples have a more flexible modulus, but their lower strength is a greater disadvantage.
- Has intermediate, more balanced CA (58.906°), which is within the favorable wetting range, suggesting good biocompatibility. In contrast, pure PP suture is highly hydrophobic (CA = 125.066°), which leads to poor tissue integration. Formula with 75% PLA is quite hydrophilic (CA = 41.6°), which might promote a stronger inflammatory response.

1:1 PP:PLA formula provides the best combination of high tensile strength and acceptable levels of ductility and biocompatibility. Although it is stiffer than the others, its superior strength is the most critical factor for a suture's primary function: holding tissue together under stress. The contact angle is also in a favorable range for tissue compatibility.

In order to improve the performance of this formula, a PP-g-MA compatibilizer were added in three ratios (1%, 2% and 4% wt.%) to enhance the miscibility between these two polymers.

Indeed, PP and PLA are inherently immiscible, because PP is non-polar and hydrophobic, while PLA is polar and hydrophilic. Therefore, it is expected that they do not mix during blending and separate into distinct phases. The expected interface between these

phases is weak due to the absence of any chemical bonds or physical entanglements holding them together. Thus, its expected mechanical properties are very poor, where under stress, such as during processing (injection) or mechanical testing (tensile strength), these phases easily pull apart, creating the voids and cracks.

Results showed that 2 wt.% is the best ratio in terms of mechanical and wettability properties (Table 2).

Table 2: Effects of PP-g-MA compatibilizer on the suture performance

PP-g-MA (wt.%)	CA (º)	TS (MPa)	Max. Deform. (mm)	TE (%)	E (GPa)
1	65	30	58	30	0.83
2	62	38	74	36	1.31
4	58	28	56	31	1.43

The DSC curves (Figure 2) proved the success compatibilization process, where it is clear that blend without compatibilizer have two melting points, while the compatibilized blend has single point.

The left image (without compatibilizer) shows a phase-separated and immiscible blend. The two polymers (PP and PLA) are not bonded, leading to large, distinct domains, poor adhesion between phases, and voids/debonding. The large voids and poor adhesion act as stress concentration points. Under tension, a crack will easily propagate along the weak interfaces, leading to premature failure and low strength. The phase-separated structure is prone to cracking when bent or tied into a knot. A knot could easily cause the internal phases to debond and fail. The debonding and large domain structure likely creates a rough surface at the microscopic level. A rough surface would corrode and damage soft tissue as it is pulled through. Micro-pores and cracks can provide a haven for bacterial growth, increasing infection risk. Also, the large, inconsistent phase domains mean the mechanical properties can vary significantly along the length of the suture, leading to unpredictable performance in vivo. Based on the above findings, this structure is unsuitable for a surgical suture.

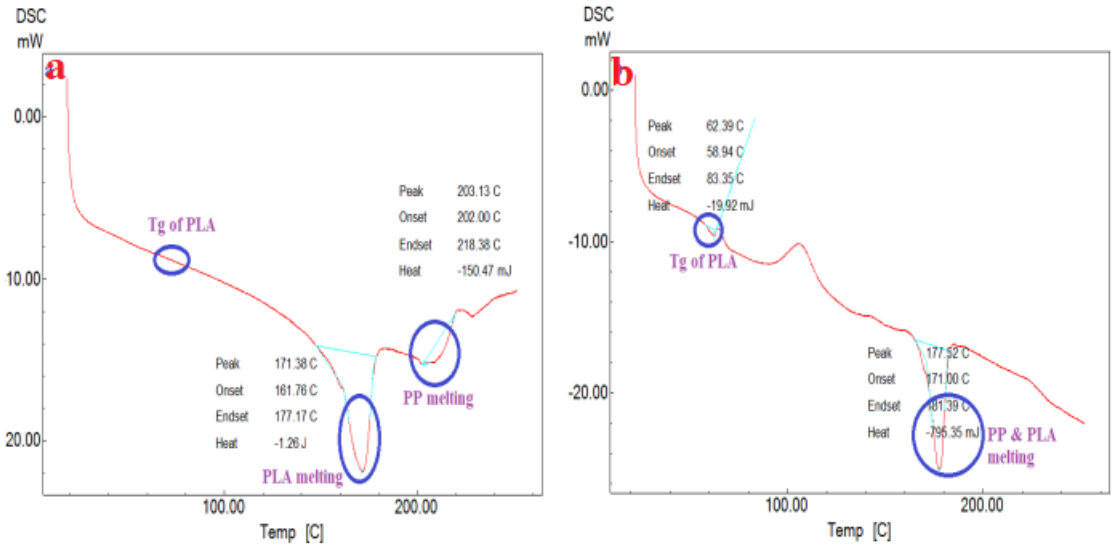

Figure 2: DSC curves of PP/PLA blend (a) without compatibilizer and (b) with 2 wt.% compatibilizer

In contrast, right image (with PP-g-MA compatibilizer) shows a compatibilized, homogeneous blend. It is expected that the compatibilizer acts as a molecular "bridge," creating a fine, interconnected structure with strong adhesion between the phases. Because of its plastic deformation and its high energy absorption, the compatibilized blend can be bent and knotted without internal failure, ensuring the knot holds securely. The fine and uniform morphology translates to a much smoother surface on the suture filament. A smooth surface glides through tissue with minimal resistance (minimizes tissue trauma**)**. A non-porous, continuous surface is less likely to harbor pathogens, thus reduces infection risk. The uniform microstructure ensures that every segment of the suture has the same reliable mechanical properties, which is critical for surgical safety and efficacy. This later structure is essential for meeting surgical suture requirements because the addition of PP-g-MA compatibilizer transforms the blend into a strong, tough, and homogeneous material by creating a finely dispersed morphology with excellent interfacial adhesion, directly fulfilling the critical mechanical and surface property requirements of a surgical suture.

Figure 3 shows the SEM images before and after the compatibilization. It is clear that the compatibilizer significantly improves the miscibility and mechanical integrity of the PP/PLA blend, as evidenced by the more uniform and finer microstructure in the right-hand image compared to the coarse, phase-separated structure in the left.

Without compatibilizer, the image shows larger, distinct phase separation between PP and PLA. The domains appear coarse and unevenly distributed, indicating poor interfacial adhesion. Larger dispersed phases (likely PLA domains in PP matrix) are visible, with sizes ranging up to several micrometers. Rough surface with clear voids or pull-out sites, suggesting weak interfacial bonding and possible brittle failure.

With compatibilizer, the image displays finer, more uniform morphology with reduced phase separation. The blend appears more homogeneous, suggesting improved interfacial adhesion. Domain size is significantly reduced, indicating better dispersion and compatibilization.Smoother fracture surface with fewer voids, indicating enhanced ductility and stronger interfacial bonding.

From the overall compatibility viewpoint, the right image demonstrates that the compatibilizer has effectively reduced phase separation, promoted finer dispersion, and improved interfacial adhesion between PP and PLA phases.

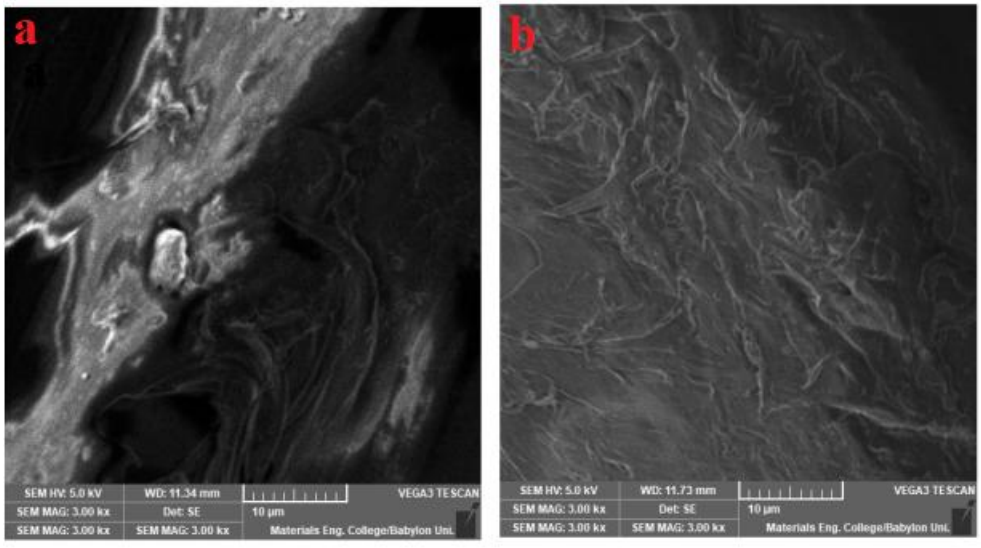

Figure 3: SEM curves of PP/PLA blend (a) without compatibilizer and (b) with 2 wt.% compatibilizer at same magnification power.

To enhance the overall bio-degradation of suture, promote faster wound healing, get better tissue regeneration and reduces the scar tissue formation, Collagen was added by 2.5, 5, 7.5 wt.% ratios. Results showed 5 wt.% collagen addition provides the best balance between mechanical strength, surface wettability and biocomptability (highest tensile strength; 43 MPa, highest elongation; 42% and a moderate contact angle; 49°). Further increment in the collagen content up to 7.5 wt.%, decreases the mechanical properties significantly, where tensile strength reduced to 36 Mpa and the elongation to 34%. This is due to aggregation within the polymer matrix that may compromise the structural integrity of the suture. Also, the angle of contact decreased down to 44º; weakened the mechanical strength and making this composition inappropriate in surgery.

To accelerate the healing action with increasing flexibility and tensile strength, Chitosan was added by 3, 5, 7 wt.% ratios. Chitosan acts as Hemostatic agent**,** which helps in stopping bleeding and as a carrier for therapeutic agents; thus, delivering drugs to the wound sites and released there in controlled manner.

The 5 wt.%. Chitosan addition showed highest tensile strength; 48 Mpa, highest elongation; 47% which showed the suture was more flexible and tougher than before. CA diminished to 42º, which indicates an enhancement of interaction between the suture and the biological tissues within the body. This addition also, gives the polymeric sutures antibacterial activity.

In order to improve the suture processability during manufacturing and tailors its flexibility, PEG was added by 10, 15, 20 wt.% ratios. Results indicate that with 15 wt.% addition, the elongation was greatly enhanced to 55% which was a good sign of increased flexibility and ductility of the surgical suture. This enhancement in elongation is very useful in surgical practice as sutures must be flexible enough to permit they be manipulated, knotted and withstand sudden mechanical stresses without a significant loss in its mechanical strength.

Moreover, the contact angle reduced to 36 º instead of 42 º, which implies that surface hydrophilicity increased. This is possible because PEG is hydrophilic in nature and it contains ether groups that are able to interact with water molecules and biological fluids.

Indeed, the following advantageous can be obtained from PEG addition:

1. Improves the blend compatibility, where its ether oxygen can interact with PLA's ester groups via dipole interactions, while its CH_2 groups have some affinity with PP.
2. PEG can acts as a plasticizer; thus lowering the T_g of PLA, making the blend more flexible, which is crucial for sutures that need handling and knotting.
3. Due to its low-melting point compared with other suture components, PEG can improve the process ability during manufacturing and tailors the flexibility of the final product. Without PEG, the suture might be strong on a molecular level and will not manufacture consistently, too stiff for a surgeon to handle and knot, or would degrade too slowly for its intended purpose.
4. Increases water absorption, which can accelerate the degradation of PLA in vivo. PEG is hydrophilic (water-attracting) and readily dissolves or undergoes hydrolysis. When it does, it creates microscopic pores and channels within the suture structure. This accelerates and controls the degradation process. Body fluids can penetrate deeper into the suture, breaking down the stronger, slower-degrading

polymers more efficiently. This allows manufacturers to design a suture that maintains its strength for a specific period (e.g., 3-4 weeks for tissue healing) and then predictably disappears.
5. Reduces brittleness of PLA without sacrificing strength completely and enhances the overall toughness.
6. Enhancing the biocompatibility, where it is FDA-approved, non-toxic, and reduces inflammatory response.
7. By varying PEG content, one can control the overall degradation.
8. PEG can be sterilized by more than one sterilization method, such as gamma and EtO sterilization.

To improve the antibacterial properties and biological performance of the obtained sutures, zinc oxide nanoparticles (ZnO) was added by 5, 7 wt.% ratios. ZnO nanoparticles have many biomedical applications because of its wide antimicrobial activity, biocompatibility and wound healing.

The 5 wt.%. ZnO nanoparticles addition showed the highest tensile strength; 49 MPa, high elongation; 53% indicating good flexibility and improved surface wettability (contact angle = 34°). In addition, this concentration provides effective antimicrobial functionality without causing significant nanoparticle aggregation within the polymer matrix.

To increase the therapeutic properties of the sutures, as well as to facilitate the controlled delivery of the drug at the wound site, Amoxicillin was chosen as an antibiotic drug because of its broad-spectrum activity against Gram-positive and Gram-negative bacteria (specifically S.aureus and E. coli, which commonly causes surgical infections).

2.3 FESEM and EDX analysis of resultant suture

The FESEM images of un-loaded suture (figure 4, left) demonstrate that suture possesses a continuous and uniform fibrous morphology with well-defined structure, indicating successful fabrication without structural defects. At the selected magnification, the suture surface exhibits noticeable roughness and irregular features, which are advantageous for enhancing surface interaction and potential biological integration. Nanoscale features corresponding to ZnO nanoparticles are clearly observed and appear relatively well-dispersed within the polymer matrix, confirming effective incorporation without significant agglomeration.

The FESEM image of loaded suture (figure 4, right) shows that suture retains a continuous and well-defined fibrous structure after the drug loading process, confirming that the impregnation method did not compromise the structural integrity of the sutures. Compared to the unloaded sample, the surface appears relatively smoother, indicating the formation of a thin drug layer covering the fiber surface. At the same magnification, ZnO nanoparticles are still identifiable; however, their visibility is partially reduced due to surface coating, with slight aggregation observed as a result of drug deposition.

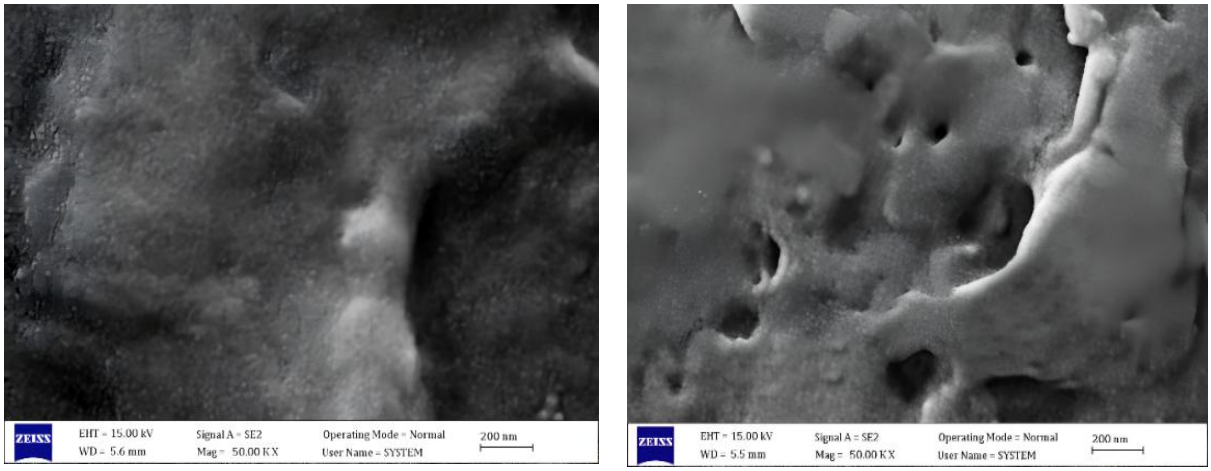

Figure 4: FESEM images of un-loaded (left) and loaded suture (right)

EDX results of un-loaded suture (figure 5, left) confirm the presence of C, O, and Zn, verifying the successful incorporation of ZnO nanoparticles into the polymeric matrix. The dominance of carbon reflects the organic composition of the suture, while the Zn signal provides direct evidence of nanocomposite formation. The absence of impurity elements indicates high purity of the fabricated structure [10].

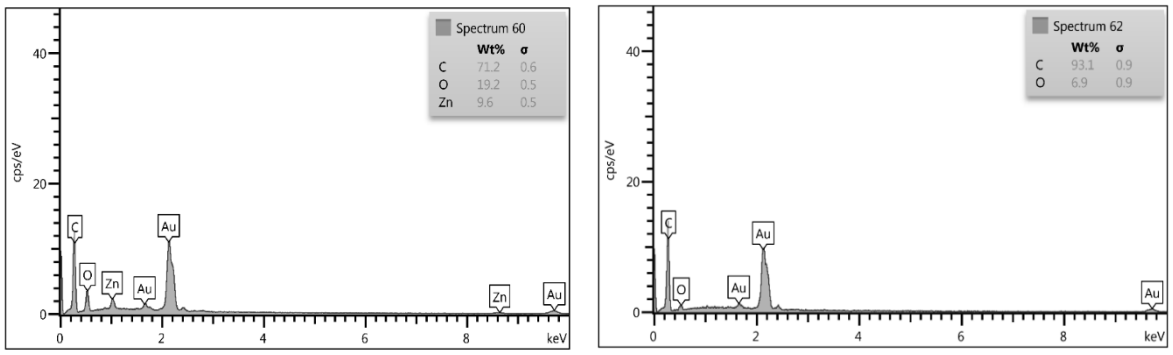

Figure 5: EDX analysis of un-loaded (left) and loaded suture (right)

In contrast, EDX results of loaded suture indicates that the loaded suture is primarily composed of carbon (C) and oxygen (O), confirming the dominance of the organic drug layer on the fiber surface. The reduced visibility of Zn compared to the un loaded sample suggests that the nanoparticles are partially covered by the drug-loading layer. The absence of impurity elements confirms the chemical stability and purity of the loading process [10].

2.4 Characterization of optimized sutures

2.4.1 Anti-bacterial test

As shown in Figure (6), both drug-loaded and un-loaded sutures exhibited clear zones of inhibition against S.aureus and E.coli, indicating effective antibacterial activity. Notably, the drug-loaded sample demonstrated significantly larger inhibition zones compared to the unloaded sample, confirming the successful incorporation and controlled release of amoxicillin within the suture.

The enhanced antibacterial performance can be attributed to the synergistic effect of the composite components. Chitosan provides inherent antimicrobial activity due to its positively charged functional groups, which interact with bacterial cell membranes and disrupt their integrity. Additionally, ZnO nanoparticles contribute to antibacterial activity through the generation of reactive oxygen species (ROS), causing oxidative damage to bacterial cells. The presence of amoxicillin further strengthens the antibacterial effect by enabling localized and sustained drug release.

Furthermore, the variation in sensitivity between S. aureus and E. coli can be explained by differences in their cell wall structures, where Gram-negative bacteria (E. coli) exhibit relatively higher resistance due to the presence of an outer membrane barrier.

Overall, the results confirm that drug-loaded suture exhibits superior antibacterial efficiency, highlighting its potential for use in antimicrobial surgical sutures to reduce postoperative infections [8].

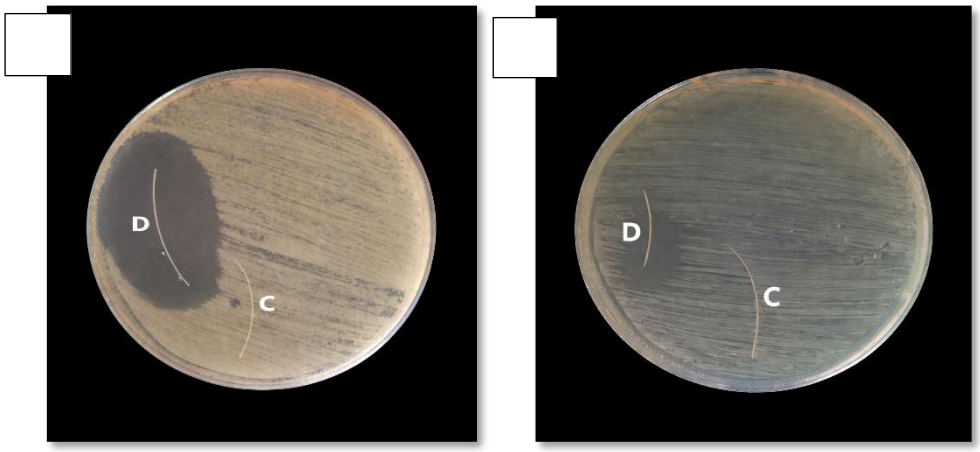

Figure 6: Antibacterial activity of sutures against (A) S. aureus, (B) E. coli.

2.4.2 MTT assay

The cytotoxic effect of the prepared suture was evaluated using the MTT assay against the human breast cancer cell line (MCF-7) as shown in figure (7).

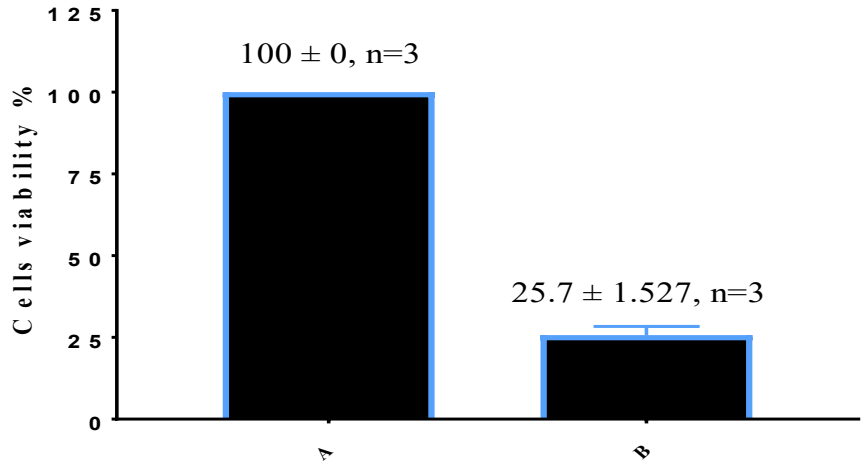

Figure 7: Cytotoxicity effect in MCF-7 cells of: A, control untreated cells. B, treated cells

The obtained results demonstrated noticeable morphological alterations in MCF-7 cells after treatment compared with the untreated control cells as shown in figure (8).

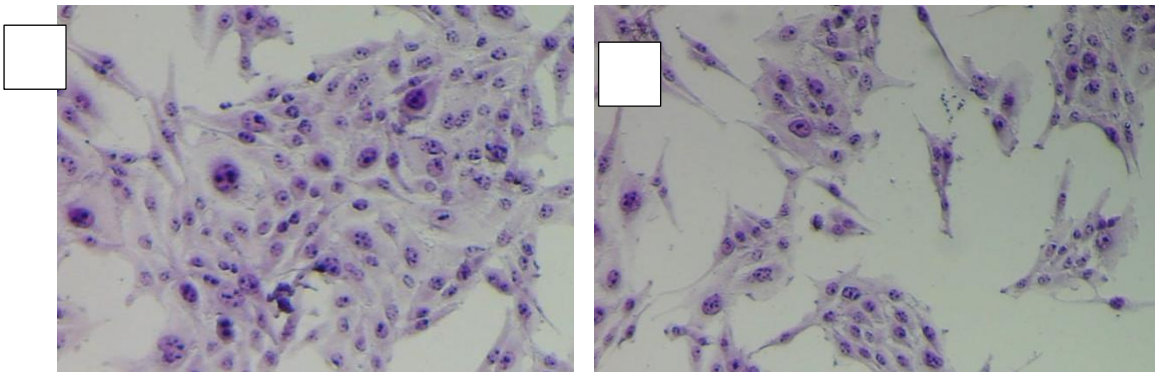

Figure 8: Morphological comparison of MCF-7 cells at 40× magnification. (A) Control untreated cells, (B) Treated cells

The control cells exhibited a normal epithelial morphology with dense and adherent cell populations, indicating healthy and metabolically active cells. In contrast, treated cells showed clear signs of cytotoxic damage including cell shrinkage, reduced cell density, membrane irregularities, and partial detachment from the culture surface.

These morphological changes indicate that the tested samples possess inhibitory effects on cancer cell proliferation. The observed cytotoxic activity may be attributed to the composite composition of the fibers and the presence of bioactive components incorporated within the structure. In particular, the presence of zinc oxide nanoparticles is known to induce oxidative stress in cancer cells through the generation of reactive oxygen species (ROS), which can damage cellular proteins, lipids, and DNA, ultimately leading to apoptosis or inhibition of cell proliferation [9].

2.5 Biocompatibility

The biocompatibility was evaluated using the MTT assay on the normal epithelial cell line (MCF-10). The obtained results demonstrated that the treated cells maintained high viability after exposure to coated suture, indicating that the developed suture formula exhibited no significant cytotoxic effect on MCF-10 cells as shown in figure (9).

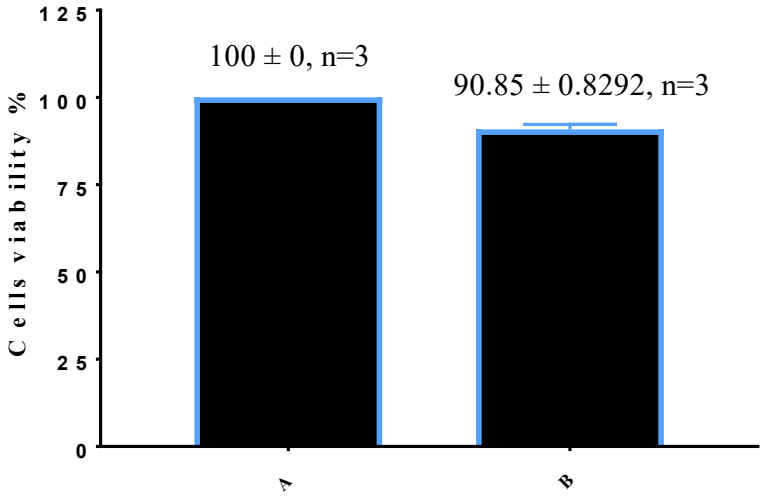

Figure 9: Cytotoxicity effect in MCF-10 cells of: A, control untreated cells. B, treated cells

These findings confirm that the polymeric matrix composed of PP, PLA, PEG, Collagen, Chitosan, and PP-g-MA provides a biologically acceptable environment that supports cellular viability. In addition, the incorporation of ZnO nanoparticles at controlled concentrations did not induce cytotoxicity toward normal cells.

Furthermore, the morphological examination of MCF-10 cells revealed that the treated cells retained their normal epithelial morphology with no observable signs of membrane damage, shrinkage, or detachment compared with untreated control cells as shown in figure (10).These observations further support the excellent cytocompatibility of the developed composite sutures.

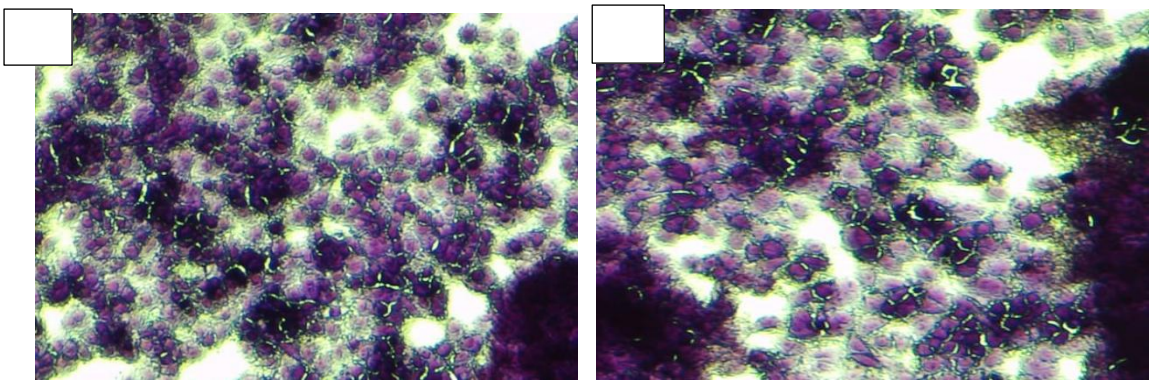

Figure 10: Morphological comparison of MCF-10 cells at 40× magnification. (A) Control untreated cells, (B) Treated cells

The results indicate that the fabricated sutures possess suitable biocompatibility characteristics, making them promising candidates for antibacterial drug-loaded surgical sutures and wound healing applications [9].

2.6 Photocatalytic activity

The photocatalytic activity of the drug-loaded sutures was evaluated through the degradation of methylene blue (MB) under irradiation. The changes in the UV–Vis absorption spectra of MB at different irradiation times are illustrated in figure 11.

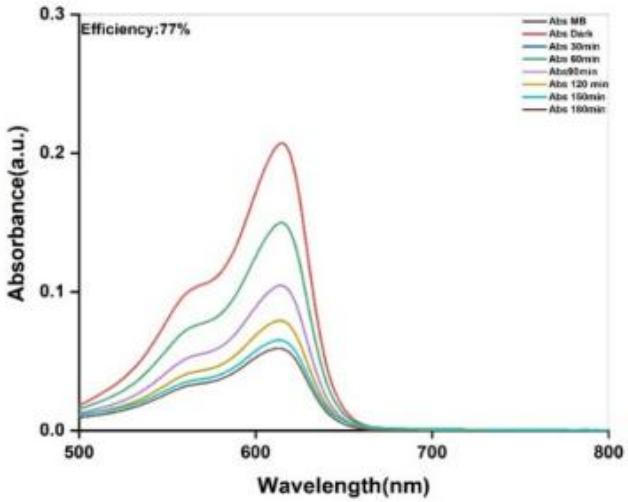

Figure 11: UV–Vis absorption spectra of methylene blue using drug-loaded suture

At the beginning, methylene blue would have a characteristic absorption peak of around 610-620 nm that is related to the electronic transition in the chromophoric structure of the dye molecule. As indicated in figure (11), the intensity of the absorbance of this peak decreases slowly with an increase in the duration of irradiation between 30 to 180 min which shows that the dye molecules continuously degrade when the photocatalyst is present. The dark phase is followed by a weak diminution of the absorbance, which indicates that some fraction of the dye molecules must have been absorbed on the surface of the composite fibers prior to the light irradiation. The main reason behind this adsorption behavior is the surface nature and porous structure of the composite material that increases the contact between the catalyst surface and dye molecules.

When the material is irradiated, the degradation process becomes more severe. The persistence in the decrease of absorbance intensity is an affirmation of the photocatalytic degradation of methylene blue molecules into smaller intermediate molecules and finally to mineralized products like CO_2 and H_2O. After 180 min of irradiation, the degradation efficiency of drug-loaded suture was about 77%, meaning that the composite material made is quite photo catalytic [11].

Figure (12) shows the visual evidence of the degradation process by showing the color of the methylene blue solution changing to a much lighter one with the increase in irradiation time. The color fading is a clear indication to the photo catalytic degradation of the methylene blue in the presence of the loaded suture. The photocatalytic activity that is observed can be ascribed to availability of semiconductor nanoparticles that are spread on the surface of the composite fibers.

Figure 12: Visual color change of methylene blue solution during photocatalytic degradation using drug-loaded suture at different irradiation times (0–180 min).

3 CONCLUSIONS

1. Among other PP:PLA blends, the 1:1 formula provides the best combination of high tensile strength and acceptable levels of ductility and biocompatibility.
2. PP-g-MA compatibilizer has effectively reduced phase separation, promoted finer dispersion, and improved interfacial adhesion between PP and PLA phases.
3. The inclusion of chitosan, collagen, polyethylene glycol, and zinc oxide nanoparticles in surgical sutures gives them an added advantage, making them multifunctional sutures.
4. Drug-loaded suture exhibits superior antibacterial efficiency compared to un-loaded one and Gram-negative bacteria exhibit higher resistance than Gram-positive bacteria.
5. Drug-loaded sutures possess inhibitory effects on cancer cell proliferation.
6. Biocompatibility results demonstrated that the current suture formula exhibited no significant cytotoxic effect on the normal epithelial cell line (MCF-10).
7. The degradation efficiency of drug-loaded suture was about 77%, meaning that the prepared suture is quite photo catalytic

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