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Mini Review

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A Mini Review on Used Medical Yarns in the Braiding Production Methods for Biomaterial Applications

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Abstract

Biomaterials are advanced materials that can be implanted within the body, depending on the required design criteria, usually as a result of the production of polymeric-based, and medical yarns by the braiding method. Braiding method is widely used in the production of biomaterials due to its advantages, including a high elastic modulus, high tensile strength, and high dimensional stability properties. The high biological and mechanical properties required in biomaterials depend on the yarn type, yarn count (dtex), braid yarn count, braid ply number, braid angle (°), and braiding structure (2-D, or 3-D) such as some braiding production process parameters. When UHMWPE, PET, HT PET, and PP multifilament yarns are used, they increase the mechanical values in the axial direction in biomaterials. The addition of ELA yarn significantly increases the maximum percentage of breaking elongation. When PGA, PLGA, and PDO multifilament yarns are used, the resorption period is between 2 and 12 months. When PA 6, PA 6.6, HDPE, UHMWPE, PP, PET, PVDF, and PTFE multifilament yarns are used, the resorption period is permanent. In order to shorten the rehabilitation period, biological chemicals such as CHI, COL, and HA can be coated on these structures by impregnation method. Afterwards, the sterilization process must be applied, and FDA approval must be also obtained. Finally, they are widely used in artificial surgical suture, artificial ligament, artificial tendon, and artificial vein applications.

Keywords:
biomaterials; yarns; braiding; biomaterial applications; vascular grafts

1. Introduction

The main purpose of this theoretical mini-review study is to determine the effects of the braiding production method used in biomaterials for the usage areas of medical yarns in braiding structured biomaterials, considering the types of yarns used and the associated process parameters. Braiding method used in the production of biomaterials provides higher elasticity modulus, higher tensile strength, and higher dimensional stability compared to foam/sponge, non-woven surfaces, weaving, and knitting methods [1].

Furthermore, when a 3-D braiding structure is employed, an increase in yarn count (dtex), braid yarn count, and braid ply number, coupled with a decrease in braid angle (°), enhances the mechanical properties in the axial direction [2–13]. Moreover, maximum mechanical values are observed when using yarn counts of 1112 dtex, 1670 dtex, and 3330 dtex [14–21]. When UHMWPE, PET, HT PET, and PP multifilament yarns, which have high elasticity modulus, maximum breaking force, and high mechanical properties in the axial direction, are used, the mechanical values in the axial direction also increase [7–13,22,23]. When ELA yarn is added, the maximum percent breaking elongation value increases as well [8].

When thermoplastic structured PGA, PLGA, and PDO multifilament yarns with low thermal properties are used, the resorption period is between 2 and 12 months.

Moreover, when thermoplastic structured PA 6, PA 6.6, HDPE, UHMWPE, PP, PET, PVDF, and PTFE multifilament yarns with relatively higher thermal properties are used, the resorption time is permanent [7–13,22,23].

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Biological chemicals such as CHI, COL, and HA add biological functionality to these yarns by impregnation method. This positively influences the development of the tissue-implant interface, thereby shortening the rehabilitation period. The resulting interface is biocompatible, non-toxic, non-carcinogenic, non-allergenic, and does not cause infection, symptoms, or trauma [1–20,22–30]. These yarns are generally used in artificial surgical sutures, artificial ligaments, artificial tendons and, artificial vein applications [7–13].

2. Biomaterials

Biomaterials have been officially approved by the FDA of the USA in accordance with internationally valid ISO and ASTM standards, and their main functions are to protect the damaged areas of the body (such as heart, vessel, ligament, tendon, surgical yarn, graft, brain, eye, teeth, hair, skin, neck, spine, hip, cartilage, knee, arm, leg, foot, joint, rib cage, lung, liver, kidney, stomach, intestine, pancreas, genital area, plate or screw) for a certain period of time. They are artificial biological materials that support the development of the cell/tissue-implant interface. They are produced based on biomechanical design criteria, sterilized, and manufactured using various production methods from natural or synthetic raw materials to meet the specific needs of the body [22,24,25,27,29]. Biomaterial applications began with the application of surgical suture in 4000 BC for the first time in world history [1,26]. In another source, surgical suture was used as a biomaterial application for the first time in Egypt in the 1100s BC [18]. Moreover, its value has been increasing in the modern era since the 1960s [23,25]. Chemical sterilization is generally applied to biomaterials with alcohol-based chemicals [1]. Although the design criteria of biomaterials vary depending on the specific needs of the damaged area within the body, they generally include high modulus of elasticity, low Poisson's ratio, high radial strength, high tensile (rupture, or fracture) strength, high compressive strength, high bending strength, high impact strength, high shear strength and high fatigue strength. Additionally, they require low creep strength, high force carrying capacity, appropriate elasticity, cheapness, fiber distribution, low density (lightness), easy raw material availability, and ease of production and implantation. Other factors include suitable yarn type, suitable yarn count, appropriate yarn cross-section, appropriate yarn structure, appropriate yarn coefficient, appropriate twist amount, appropriate fabric type, appropriate fabric construction, appropriate fabric density values, production method angle, appropriate geometric shape, appropriate porosity, appropriate porosity size, high and homogeneous porosity distribution. Further, biomaterials should exhibit appropriate length, appropriate diameter, high biocompatibility, biodegradability, sometimes high bioinertness depending on the intrabody site, surface roughness, high wear resistance, high corrosion resistance, high chemical resistance, high pH resistance, blood pressure resistance, high dimensional stability, high tissue-implant interface bond, high cell-tissue interface development support, fibroin-based protein support, cell encapsulation support, being able to carry drugs, closing wounds, supporting tissue integration, and supporting cell/tissue adhesion. Moreover, ideal biomaterials should prevent blood clotting, infections, and adverse symptoms; support rapid rehabilitation; and not cause trauma or immunological reactions. They should also be blood-compatible, osteogenic, angiogenic, non-thrombogenic, antimicrobial, antitoxic, anticancer, antitumor, and antipathogenic [1–20,22]. Biomaterials can be produced primarily from polymeric, metallic, ceramic, composite materials and cell-tissue based materials [24]. Polymeric materials are chitosan (CHI), collagen (COL), alginate (ALG), gelatin (GEL), heparin (UFH), hyaluronic acid (HA), chondroitin sulfate (GAG), N-butyl cyanoacrylate, N-butyl cyanoacrylate, agarose (AGA), carbon (C), glass (G), polyparaphenyleneterephthalamide (PPD-T), cotton (CO), silk (SI), spider silk (SSI), linen (LI), hemp (FL), jute (J), ramie (R), viscose rayon (CMD), lyocell (LYO), elastane (EL), ultra-high molecular weight polyethylene (UHMWPE), polyacetal (PAc), polyamide (PA), polycarbonate (PC), polyethylene (PE), polypropylene (PP), polyurethane (PU), polystyrene (PS), palmitoylethanolamide (PEA), polyethylene terephthalate (PET), polytriethylene terephthalate (PTT), polybutylene terephthalate (PBT), polybutylene succinate (PBS), polydioxanone (PDS), plastic optical fibers (POF), polydimethylsiloxane (PDMS), polyhydroxyalkanoates (PHA), polyhydroxybutyrate (PHB), poly(3-hydroxybutyrate-co-3-hydroxyvalerate) (PHBV), polybutylene adipateterephthalate (PBAT), polyethyleneglycol (PEG), polyvinylpyrrolidone (PVP), polyvinylalcohol (PVA), polycaprolactone (PCL), polylactic acid (PLA), polyglycolic acid (PGA), polyglactin (PGLA), poly L-lactic acid (PLLA), poly lactic-co-glycolic acid (PLGA), polyaryletherketone (PAEK), polyetheretherketone (PEEK), polyetherketoneketone (PEKK), polyvinylidenedi-fluoride (PVDF), polytetrafluoroethylene (PTFE), polymethylmethacrylate (PMMA), poly(2-hydroxyethyl meth-acrylate) (PHEMA) and polysiloxane materials [1–20,22–30]. Metallic materials are stainless steel (SS), titanium (Ti), titanium alloys (TiO_2, Ti-6Al-4V, Ti-6Al-7Nb, Ti-15Mo-5Zr-3Al), cobalt-chromium (Co-Cr) alloys (Co-Cr-Mo, Co-Cr-W-Ni, Co-Ni-Cr-Mo, Co-Cr-Ni-Mo-Fe, Co-Ni-Cr-Mo-W-Fe), NiTi, tantalum (Ta)), and tungsten (W) materials [3,17,24,25].

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Ceramic materials are mostly calcium phosphate (CaP) based hydroxyapatite (HA), tricalcium phosphate (TCP), high purity alumina (Al), aluminum zirconia rich alumina (ZTA), yttria stabilized zirconia (YSZ), and bioactive glass (BG) materials [24–26]. Composite materials are materials produced by combining polymeric, metallic and ceramic materials [20,21,23–26]. In biomaterial applications, carbon and glass yarns are commonly used with thermal processing (curing) in combination with epoxy, vinyl ester, and unsaturated polyester resins [20]. The production methods of biomaterials vary depending on the type of raw material used such as polymeric, metallic, ceramic or composite materials. Their production methods are monofilament or multi-filament yarns, braiding, weaving, knitting, non-woven surface, embroidery, sol-gel, electrospinning, coating, plasma, various casting methods, various infusion methods, CVD, and 3-D printing methods [1–30]. Produced forms of biomaterials are nanorods, rods, cylindrical structures, plates, biofilms, foams, sponges, cheesecloths, yarns, ropes, fabrics, gels, and solid materials with the desired profile geometry [1–30]. Effective production process parameters in the production methods of biomaterials are concentration, atom types, interatomic distance, bond types and bond strengths, polymerization degree, molecular weight, molecular chain configuration and conformation, glass transition temperature, crystallization temperature, melting temperature, decomposition temperature, temperature, time, pressure and environmental conditions [1–20, 22–30]. The applications of biomaterials are orthopedic, heart, vein, ligament, tendon, surgical yarn, graft, brain, eye, tooth, hair, skin, neck, spine, hip, cartilage, knee, arm, leg, foot, joint, rib cage, lung, liver, kidney, stomach, intestine, pancreas, genital area, plate and screws [1–3,14–19,22–26,29,30]. Application areas of biomaterials are presented in Table 1 [26].

3. Used Yarns Properties in Biomaterial Applications

Various technical properties of some polymeric-based yarns commonly used in biomaterials are presented in Table 2 [22,23].

4. Used Textile Production Methods in Biomaterial Applications

The textile structures used in biomaterials are the most commonly used textile structures are braiding, weaving and knitting textile structures [4,5,7–11,13,14,23–30].

5. Used Braiding Production Methods in Biomaterial Applications

Braiding structures are produced by the mutual sinusoidal movement of braid beds positioned opposite each other at 180° of at least two yarn groups. Moreover, the unit cell structure is more closed and rigid compared to all other textile production methods except weaving [1–6,16,18–23,27,30]. Braiding structures can be produced by vertical or horizontal production methods in various geometries, especially round, square, rectangular, hexagonal, H, I, J, and T cross-section profiles [2–5,20,21]. The hexagonal profile enables the production of the densest braiding structures as it ensures that the braiding formation point with braid coils is reduced to a minimum distance [2]. Moreover, although 1 × 1 (diamond), 2 × 2 (regular) and 3 × 3 (hercules) braiding constructions are widely used, they can be also produced as 2-D, or 3-D braiding structures with 5 × 5 braid beds, and different braiding constructions [2–13,20,21]. For 3-D braiding structures, either a center yarn or center yarns are added, or a mandrel with various geometric profiles that can be solid or empty is added [2–4]. 3-D braiding structures have higher yarn density and tensile strength values compared to 2D braiding structures [14,25–28]. The most effective production parameters in the production of braiding structures are unit cell cage geometry, number of braid yarns, number of center yarns, fiber and yarn diameters, yarn type, yarn number, cross-section of the yarn, number of filaments in the yarn and twist amount, yarn coefficient, yarn-count, surface area of the yarn intersection, yarn route length, yarn tension, covering factor, braid angle, braid bed geometry, revolution and speed of the head gears in the braid bed and draft ratio of the gears [2–13,20,21]. The yarn types used in braiding structures are PA, PE, PP, PET, PAN, PCL, PLLA, ePTFE, C, G, PPD-T, UHMWPE, NiTi, and Al_2O_3 [2–13,20,21]. The yarn numbers used in braiding structures can be used in yarn counts of 1112 dtex, 1240 dtex, 1670 dtex, 2200 dtex, 3300 dtex, 3400 dtex, and 10,000 dtex [2,7–13]. Carbon yarn applications are excluded. 12k filament count is used in biomaterial applications of carbon yarn [2,4,20]. Although the number of braid yarns is commonly produced in the range of 3 to 48 braid yarns, it can also be produced in the number of braid yarns of 24, 32, 48, 64, 80, 96, or 144 depending on the desired biomechanical performance value in the biomaterial application field [2–13]. Braiding structures are widely used in artificial surgical suture, artificial ligament, artificial tendon, artificial vessel and artificial bone applications [2,3,7–13]. Although it is known that braiding structures can theoretically be produced with

Table 1: Application areas of biomaterials [26].

Non-Implanted	Extracorporeal Implants	Healthcare/Hygiene Products
wound dressing bandage plaster lint-free wadding	artificial kidney artificial liver artificial lung	surgical gown surgical sheath bedding clothes baby diaper/baby sheet clothes/ handkerchief surgical socks
Soft tissue implants		
PP and PET ePTFE PLA, PGA, PCL and PDO Chitin PMMA and Silicon COL	warp knitting nonwoven surfaces nonwoven surfaces nonwoven surfaces nonwoven surfaces nonwoven surfaces	abdominal wall hernia treatment hernia treatment artificial skin eye contact lens and artificial cornea eye contact lens and artificial cornea
Orthopedic Implants		
Silicon and PE PE, PP, PES, PTFE, SI and COL PES, C and COL LDPE CHI	weaving and braiding weaving and braiding braiding braiding nonwoven surfaces	artificial joint/bone artificial tendon artificial ligament artificial cartilage bone regeneration
Cardiovascular grafts		
PES and PTFE PLA, PLGA and COL PES, PU and PGA	weaving and knitting braiding nonwoven surfaces	vascular graft heart valve and blood vessel heart valve and blood vessel
Sutures		
COL, PLA and PGA PES, PA, PE, PP and PTFE	monofilament braiding	biodegradable surgical sutures non-biodegradable surgical sutures

braid angles between 1° and 179° in the production of biomaterials, it is generally common and recommended to produce them with braid angles of 30°, 45°, and 60° from experimental studies [2–6,20,21]. The 45° braid angle provides maximum tensile strength and yarn density, while the 60° braid angle provides minimum labor and maximum production speed in biomaterial applications of braiding structures [4,6,20]. Moreover, braiding structures are produced in the diameter range of 5 mm to 24 mm, especially for artificial vascular applications [2].

6. Used Weaving Production Methods in Biomaterial Applications

Woven structures are produced by placing at least two yarn groups (warp and weft yarns) against each other at an angle of 90° and combining them with a beat-up motion. Moreover, the unit cell structure is more closed and rigid compared to all other textile production methods [1,16,22,23,27,28,30].

7. Used Knitting Production Methods in Biomaterial Applications

Knitted structures are primarily divided into warp and weft knitting methods. In warp knitting, they are produced by feeding each yarn to the warp knitting machine separately and creating a closed chain loop structure depending on the loop form of the needle bed. Moreover, the unit cell structure is more closed and less elastic compared to weft knitting [1,16,22,23,27,28,30]. In weft knitting, the yarn is fed to the circular knitting machine from a single point, where it forms an integral loop structure based on the loop configuration of the needle bed. Moreover, the unit cell structure is more open and flexible [1,16,22,23,27,28,30].

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Table 2: Various technical properties of some polymeric-based yarns commonly used in biomaterials [22,23].

Polymer Type	Chemical Structure and Physical	Production Method and Application	Applications	Resorption Time
PA 6	thermoplastic, hydrophobic, T_g = 45 °C, T_m = 220 °C and % 50 crystallinity	melt spinning production method in monofilament or multifilament yarn form	surgical sutures applications	permanent
PA 6.6	thermoplastic, hydrophobic, T_g = 50 °C, T_m = 265 °C and 50% crystallinity	melt spin-ning pro-duction method in monofila-ment or multifila-ment yarn form	surgical sutures applications	permanent
HDPE	thermoplastic, hydrophobic, T_g = 50 °C, T_m = 125 °C and crystallinity up to 85%	melt spinning method and reinforcement of polymer composites	orthopedic and craniofacial applications	permanent
UHMWPE	thermoplastic, hydrophobic, T_m = between 140 °C and 150 °C, above 95% crystallinity, high modulus of elasticity tensile strength	sol-gel spinning method and very high tensile strength	ligament and force bearing applications	permanent
PP	thermoplastic, hydrophobic, isotactic, T_m = between 165 °C and 175 °C and also crystallinity between 40 and 46%	nonwoven surfaces with melt spinning method and monofilament, hollow, spunbond and melt-blown production methods	surgical sutures, hernia treatment, mesh and blood filter applications	permanent
PET	thermoplastic, hydrophobic, T_g = between 65 °C and 105 °C, T_m = 265 °C and crystallinity up to 100%	melt spinning method and weaving, knitting and braiding applications in monofilament or multifilament yarn form	surgical sutures, hernia treatment, ligament, heart valve, vascular graft and endo-vascular graft applications	permanent
PVDF	thermoplastic, hydrophobic, T_m = between 165 °C and 175 °C and crystallinity between 52 and 66%	melt spinning production method in monofilament or multifilament yarn form	surgical sutures, vascular graft, and endo-vascular graft, heart valve, embolic vena cava filter, ligament applications	permanent
PTFE	thermoplastic, hydrophobic, T_g = 45 °C, T_m = 325 °C and crystallinity between 50 and 75%	melt spinning production method in monofilament or multifilament yarn form	surgical sutures	permanent
PCL	thermoplastic, in rubber form at room temperature, T_g = −65 °C, T_m = between 58 °C and 63 °C	melt spinning production method in monofilament or multifilament yarn form and electro-spinning production method in nanofiber form	tissue scaffolding applications	above 24 months

PLA	thermoplastic, T_g = between 60 °C and 65 °C, T_m = between 173 °C and 178 °C	melt spinning production method in monofilament or multifilament yarn form and electro-spinning production method in nanofiber form	graft, drug delivery and tissue scaffolding applications	above 24 months
PGA	thermoplastic, T_g = between 40 °C and 45 °C, T_m = 225 °C and crystallinity between 45% and 55%	melt spinning production method in monofilament or multifilament yarn form and electro-spinning production method in nanofiber form	surgical sutures, mesh and tissue scaffold applications	between 2 and 12 months
PDO	thermoplastic, T_g = between −10 °C and 0 °C, T_m = between 110 °C and 115 °C and 55% crystallinity	melt spinning production method in monofilament or multifilament yarn form and electro-spinning in nanofiber form	surgical sutures, intramedullary pins; binding clip applications	between 6 and 9 months
PLGA	thermoplastic, T_g = between 35 °C and 60 °C, T_m = between 120 °C and 200 °C	melt spinning production method in monofilament or multifilament yarn form	surgical sutures, vascular graft, and endo-vascular graft applications	between 2 and 5 months
SI	wet spinning production method, suitable for bicomponent fiber production, hydrophobic, 70% crystallinity, low elastic recovery behavior	wet production method for removing the sericin part of the raw fiber secreted by the Bombyx mori silkworm.	surgical sutures	between 1 and 2 years
CMD	dry or wet spinning production method, hydrophobic, T_d = 250 °C and 33% crystallinity	it is produced in the form of multifilament yarn from alkaline cellulose xanthate solution and generally by wet spinning production method.	wound dressing applications	permanent

As the yarn count increases, the fabric density values increase, the skipping behavior of the yarns forming the fabric is minimized, and the yarn intersection points increase, the pore size of the biomaterials decreases, the number of pores increases, and cell adhesion and proliferation increases. Shear (shear) strength decreases while tensile strength increases [1,16,18,19,22,23,26,29,30]. Comparative biomechanical technical analysis of textile structures used in biomaterials is presented in Table 3 [1].

8. Used Yarns in Biomaterial Applications

Various yarns produced by the braiding production method in biomaterial applications and their technical properties have been compiled from literature sources [7–13]. In an artificial surgical suture application, raw samples were produced with SI as yarn type, 124 tex as yarn count, single-layer (x1), 1 × 1 (diamond) as braiding construction, 16 braid yarns and 2-D braiding structure. After this, they were coated with CHI concentration (c%) for its 1, 2, and 3, at used 100 °C temperature for 10 min. According to experimental results, the dynamic friction coefficient was lower in CHI-coated samples compared to non-CHI-coated samples. As the CHI concentration increased, tensile strength, knot strength and anti microbial activity increased [7]. In an artificial ACL ligament application, raw samples were produced with HT PET, and EL as yarn types, 1670 dtex, 2200 dtex, and 3300 dtex for HT PET as yarn count, and 420 dtex, 960 dtex, and

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Table 3: Comparative biomechanical technical analysis of textile structures used in biomaterials [1].

Technical Specifications	Foam/Sponge	Nonwoven Surfaces	Woven	Braiding	Knitted
Elasticity modulus	Low	Low	High	High	Medium
Tensile strength	Low	Low	High	High	Low
Dimensional stability	Good	From weak to good	Excellent	Excellent	From weak to good
Drapeness	Low	Good	From weak to good	Low	Excellent
Other technical features	Isotropic behavior	Isotropic behavior	Anisotropic, high fiber parallelism, fiber normally weak	Anisotropic, high biomechanical properties in the axial direction and low in the radial direction	Tendency from isotropic to anisotropic behavior

1280 dtex for EL as yarn count, four-ply (x4), 1 × 1 (diamond) as braiding construction, 1 center yarn and 16 braid yarns, from 40° to 61° braid angle, and 3-D braiding structure. According to experimental results, as the braid angle increased, axial tensile strength values decreased, while the percentage of elongation at break increased. Moreover, these values were much more evident in samples with EL compared to samples with only HT PET. While the axial tensile strength values decreased for both HT PET, and EL yarn types as the yarn count increased, the percent elongation at break values remained almost unchanged for the HT PET yarn but increased for the EL yarn type [8]. In an artificial Achilles tendon application, raw samples were produced with PET, and PP as yarn types, 1112 dtex for PET as yarn count and 1200 dtex for PP as yarn count, double-layer (x2), 1 × 1 (diamond) and 2 × 2 (regular) as braiding constructions, 1 center yarn and 16 braid yarns, 10° to 60° braid angle, and 3-D braiding structure. According to experimental results, as the braid angle increased, the braid diameter decreased. Axial tensile strength, modulus of elasticity, maximum breaking force, and maximum percent elongation at break increased as the number of yarns increased. Moreover, PET as yarn type had all these biomechanical values higher than PP as yarn type. PET as yarn type creep, and fatigue strength values were higher than PP as yarn type [9]. In an artificial ACL ligament application, raw samples were produced with UHMWPE, PPD-T, and HT PET as yarn types, 445 dtex for UHMWPE as yarn count and 1670 dtex for PPD-T, and HT PET as yarn count, single-layer (x1), 1 × 1 (diamond) and 2 × 2 (regular) as braiding constructions, 1 center yarn and 16 braid yarns, 45° braid angle, and 3-D braiding structure. CHI (c = 2%), and GA (c = 25%) chemicals were first dissolved in solution at pH between 5 and 5.5, 70 °C temperature, and 3 hours as a biofinishing process, then they were absorbed and coated for 0.5 h. A biological in-vitro environment was then created using PBS solution. Finally, the materials were sterilized through EtO sterilization. FT-IR chemical technical analysis was performed. According to experimental results, the most common bond formations were observed as HT PET > PPD-T > UHMWPE, respectively. It was reported that the reasons for this were yarn types, chemical structures of the yarns, bond types, molecular weights, crystallinity, reactivity, ease of diffusion and diffusion rate [10]. In an artificial surgical suture application, raw samples were produced with PA 6.6 as yarn type, 44 dtex as yarn count, 23 as filament count, single, double and triple ply (x1, x2 and x3), 1 × 1 (diamond) as braiding construction, 1, 2, and 3 center yarn, and omly 12 and 16 braids, 350 t/m, 500 t/m, and 650 t/m as twist amounts and 3-D braiding structure. According to experimental results, as the number of braid yarns, number of yarn layers and twist amount increased, braid diameter, tensile strength and knot strength values increased. Node activity first decreased, then increased, and finally decreased again. Deformation recovery (elastic recovery) behavior varied [11].

In an artificial ligament and tendon application, raw samples were produced with PCL, and UHMWPE as yarn types, 110 deniers as yarn count and 32 as filament count for PCL and 100 denier 82 as filament count for UHMWPE, PCL/UHMWPE ratio 100/0, 75/25, 50/50, 25/75, and 0/100 ratios, respectively. They had also single-layer (x1), 1 × 1 (diamond) as braiding construction, 4 + 4 and 8 braid yarns and a 2-D braiding structure. According

to experimental results, as the PCL/UHMWPE ratio went from 100/0 to 0/100, the modulus of elasticity, maximum breaking force, axial tensile strength, creep strength, fatigue strength and knot strength values increased, but the maximum percent elongation at break values decreased. The general result was that the PCL/UHMWPE with 2-D braiding structures for their 50/50 and 25/75 ratios were reported to have optimum mechanical results [12]. In an artificial bone application, raw samples were produced with PLA as yarn type, 75,150 and 300 deniers as yarn numbers, 36 as filament count, double, three and four layers (x2, x3 and x4), 1 × 1 (diamond) as braiding construction, 16 braid yarns, and 2-D braiding structure. Then, they were impregnated and coated with 3% concentration of CHI at pH 7 at a temperature range of 20 °C to 22 °C for 6 hours. According to experimental results, it was determined that 150 denier 4-ply PLA yarn had the maximum elasticity modulus, maximum breaking force and axial maximum tensile strength values. Moreover, it was observed that thermal degradation began after 70 days. It was also determined that CHI-coated PLA samples produced less Ca^{+2} ion. It was reported that the reason for this was that CHI vibrated the carbonyl group on the PLA structure at a wavelength of 1730 cm^{-1}, thus accelerating its conversion to Ca^{+2} [13].

9. Conclusions

Biomaterials have increased their importance and use in the body since the 1960s until today. The raw materials used in biomaterials are generally structures in the form of yarns produced by the 3-D braiding production method. In addition to polymeric materials, they can be used in metallic, ceramic and composite materials. The materials produced in this way are typically used in applications such as artificial surgical sutures, artificial ligaments, artificial tendons, artificial blood vessels, and artificial bone. UHMWPE, PPD-T, HT PET, PET, PP, PLA, and PCL yarns are widely used. It has been determined from experimental studies that they are affected by yarn's chemical, yarn's mechanical, yarn's thermal and yarn's biological properties, yarn type, yarn count, number of yarn layers, twist amount, center and braid yarn count, braid angle, braiding construction and biofinishing process for biomechanical performance values in biomaterials. After production, it must be subjected to a sterilization process. Chemical sterilization, particularly using alcohol-based chemicals, is the most commonly employed sterilization method. When C, G, UHMWPE, PPD-T, HT PET, and PET yarns are used to have high values for biomechanical performance values in biomaterials. When the high biological performance values are desired in biomaterials. It is recommended to use PLA, PGA, PCL, and PLGA yarns. The biomaterials produced should not cause various complications, and symptoms within the body, and should support the rapid rehabilitation process. Official approval is required from the FDA in US.

List of Abbreviations

AGA	Agarose
Al	High Purity Alumina
ALG	Alginat
BG	Bioactive Glass
C	Carbon
CaP	Calcium Phosphate
CHI	Chitosan
CMD	Viscose Rayon
CO	Cotton
Co-Cr	Cobalt-Chrome
COL	Collagen
EL	Elastane
G	Glass
GAG	Chondroitin Sulfate
GEL	Gelatin
HA	Hyaluronic Acid
HAp	Hydroxyapatite
HE	Hemp
J	Jute
LI	Linen
LYO	Lyocell
NiTi	Nitinol
PA	Polyamide
Pac	Polyacetal
PAEK	Polyaryletherketone
PBAT	Polybutyleneadipateterephthalate
PBS	Polybutylenesuccinate
PBT	Polybutyleneterephthalate
PC	Polycarbonate
PCL	Polycaprolactone
PDMS	Polydimethylsiloxane
PDS	Polydioxanone
PE	Polyethylene
PEA	Palmitoylethanolamide
PEEK	Polyetheretherketone
PEG	Polyethyleneglycol
PEKK	Polyetherketoneketone
PET	Polyethylene Terephthalate
PGA	Polyglycolic Acid
PGLA	Polyglactin
PHA	Polyhydroxyalkanoates
PHB	Polyhydroxybutyrate
PHBV	Poly(3-hydroxybutyrate-co-3-hydroxyvalerate)
PHEMA	Poly(2-hydroxyethyl methacrylate)
PLA	Polylactic Acid

PLGA	Poly Lactic-co-glycolic Acid
PLLA	Poly L-lactic Acid
PMMA	Polymethylmethacrylate
POF	Plastic Optical Fibers
PP	Polypropylene
PPD-T	Polyparaphenyleneterephthalamide
PS	Polystyrene
PTFE	Polytetrafluoroethylene
PTT	Polytriethyleneterephthalate
PU	Polyurethane
PVA	Polyvinylalcohol
PVDF	Polyvinylidenedifluoride
PVP	Polyvinylpyrrolidone
R	Ramie
SI	Silk
SS	Stainless Steel
SSI	Spider Silk
Ta	Tantalum
TCP	Tricalcium Phosphate
Ti	Titanium
UFH	Heparin
UHMWPE	Ultra High Molecular Weight Polyethylene
W	Tungsten
YSZ	Yttria Stabilized Zirconia
ZTA	Aluminum Zirconia Rich Alumina

Author Contributions

The author is solely responsible for conceptualization, methodology, validation, formal analysis, investigation, resources, data curation, writing-original draft preparation, writing-review and editing, project administration. The author has read and agreed to the published version of the manuscript.

Conflicts of Interest

The author declares no conflicts of interest.

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AI Declaration

The authors confirm that no AI tools were used to generate any content of this manuscript. The authors take full responsibility for the accuracy, originality, integrity, and scientific content of the manuscript.

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