Chapter

Intraocular Lens (IOL) Materials

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Abstract

In 1949, first intraocular lens (IOL) insertion after cataract surgery was performed by Sir Harold Ridley, in London. Only in the 1970s, the IOL insertion after cataract surgery began to be a standard procedure. The material the first IOL-s were composed of was polymethyl methacrylate (PMMA). The PMMA is a rigid material and the corneal incision had to be at least as big as the IOLs optic and it became its biggest disadvantage in the cataract surgery. The main goal of modern cataract surgery is as smallest incision possible, so the IOL-s had to be flexible and therefore foldable. This goal was achieved by improvements in the IOL design and materials that made them foldable. First foldable IOL-s were made of hydrogel but they were unstable and the development of the first silicone IOL-s overcame that problem. Foldable silicone IOL-s were first implanted in 1978 by Kai-yi Zhou. Foldable IOL's benefits are its compatibility with a small incision surgery that is self-sealing procedure and the possibility of insertion by a single-use applicators that made the surgery safer. In the future, we can expect some new, different and innovative approaches in the IOL design and materials.

1. Introduction

Intraocular lenses (IOL) are implanted in the eye in order to treat refractive errors produced by extraction of the lens as a standard procedure in cataract surgery.

IOL is designed and composed of optic—central part, and the haptics—side structures that keep the lens inside the capsular bag.

The first intraocular lens was inserted in 1949 after cataract surgery by Sir Harold Ridley in St Thomas Hospital in London [1]. The material the first IOLs were composed of was polymethyl methacrylate (PMMA). It was a rigid nonfoldable material making the placement of the IOL challenging [2]. In the 1970s, the new lighter posterior chamber IOLs were designed and had propylene haptics for better stabilization and ciliary sulcus fixation and the IOL insertion after cataract surgery began to be a standard procedure.

In the early 1980s, Epstein began to use lenses made of silicone with the intention to make them foldable. That way they could be inserted into the eye through the small incisions of 3 mm and less compared to 5–7 mm incisions needed for nonfoldable IOLs insertion [3, 4]. The practice of IOL implantation was revolutionized in 1984 when Thomas Mazzocco began folding and implanting the plate haptic silicone IOLs [5].

Current materials used for IOL optics are of two types—acrylic and silicone. Acrylic materials can be rigid (PMMA) and foldable made of hydrophobic acrylic materials (AcrySof - Alcon Laboratories, Sensar – Advanced Medical Optics – AMO) and hydrophilic acrylics (Centerflex, Akreos).

Each foldable acrylic lens design is made from a different copolymer acrylic with a different refractive index, glass transition temperature, water content, mechanical properties and other attributes.

Hydrophobic acrylic lenses and silicone lenses have very low water content (less than 1%). But there are hydrophobic acrylic materials with higher water content about 4% also available. Hydrophilic acrylic lenses are made from copolymers with higher water content ranging from 18 to 38%.

The first silicone material that was used in the industry of IOLs was polydimethylsiloxane, with refractive index of 1.41 while the new silicone materials have higher refractive indexes.

Refractive index in foldable acrylics is 1.47 or greater, and for silicone lenses is lower—1.41 and higher. Therefore acrylic lenses are thinner than silicone ones with the same refractive power.

2. Materials

2.1 Biocompatibility

The biocompatibility of a material is dependent of a biological response to a foreign body material and it depends on the design and the material of the implant. The material should be chemically inert, physically stable, noncarcinogenic, non-allergenic, capable of fabrication in the required form, and have no foreign body reaction [6]. Materials used in ophthalmology should also be optically transparent for long period of time, have a high resolving power or refractive index, and should block ultraviolet rays.

The reaction of lens epithelial cells and the capsule to IOL material and design is capsular biocompatibility. The uvea's reaction to the IOL is uveal biocompatibility [7]. During cataract surgery the blood-aqueous barrier is disrupted and proteins and cells are released in the aqueous humor. Proteins then adsorb on the IOL surface and this will influence subsequent cellular reactions on the IOL [8].

3. Glistenings

Glistenings are a phenomenon caused by penetration of aqueous humor into the IOL material causing vacuole formation in the IOLs optic [9].

Glistenings are fluid-filled microvacuoles that form within the IOL optic when the lens is in an aqueous environment. They can be observed with any type of IOL more often in association with hydrophobic acrylic lenses.

Factors that may influence the formation of glistenings include IOL material, manufacturing technique and packaging and also the associated conditions of the eye-glaucoma, conditions leading to breakdown of the blood-aqueous barrier and use of ocular medications.

Some theories refer glistenings as a cavitation within the IOL from slow moving hydrophilic impurities within the IOL. An osmotic pressure difference between the aqueous solution within a cavity and the external media in which the lens is immersed leads to growth of the cavity [10].

Glistening develop over time and indicate a dynamic process within the lens/eye system. Causes and long-term outcomes are not entirely clear [11].

Hydrophobic acrylic IOL have the highest degree of lens glistening in comparison to the silicone and the HSM-PMMA IOL 11.3–13.4 years after surgery. The HSM-PMMA IOL had almost no lens glistenings. Lens glistening do not interfere with the dioptric power of the hydrophobic acrylic lens IOL [12].

4. Hydrophobicity and hygroscopy

Hydrophobicity is a measure of material's tendency to separate itself from water. Every material has its measurable hydrophobicity that is graded using contact-angle measurements and it is a surface property [13–15]. It ranges from only a few degrees for almost perfectly hydrophilic surfaces, such as bare silica glass prepared with dangling hydroxyl groups [16] to almost 180° for super-hydrophobic surfaces [14].

Hydrophobicity is highly dependent of the material's chemistry since the oxygen–hydrogen bonds in water are highly polar. Partial electric charges on the atoms tend to be attracted to opposite charges. That way water dissolves salts and is attracted to materials that also have partially charged bonds. Polymers consist primarily of nonpolar carbon–carbon and carbon–hydrogen bonds, which is why they are not generally hydrophilic and is attracted to materials with partially charged bonds.

Hygroscopy explains a material's tendency to absorb and hold water. A highly hygroscopic material draws water into itself. In ophthalmology the hydrophobicity has been used to describe both the surface and interior of IOLs. The interaction of an IOL's surface with water is a measure of hydrophobicity and the ability of IOLs to draw water into their interior a hygroscopy.

5. Polymethyl methacrylate

The first IOL, implanted in 1949, was made of PMMA. There have been reports of original lenses implanted by Ridley remaining perfectly clear and centered for more than 28 years [3]. There were also reports of some spontaneous dislocations into the vitreous [5].

It is a rigid, nonfoldable material with less than 1% water content and therefore hydrophobic. PMMA IOLs are usually single pieced, large and therefore nowadays rarely used. They have a refractive index of 1.49 and usual optic diameter 5–7 mm. They are s too rigid to fold and therefore the lens cannot pass through the small incisions used phacoemulsification.

6. Silicone

Silicon IOLs were designed to allow implanting through the incision smaller than the optics diameter. Implantation of silicone IOLs was introduced in 1984 [17]. Silicone is a hydrophobic material of refractive index 1.41–1.46 and the optic diameter of 5.5–6.5 mm. Models are three-piece design with PMMA, polyvinyl difluoride (PVDF) and polyamide haptics. The problem with silicone is an abrupt opening in the anterior chamber following implantation which may cause rupture of the posterior capsule.

Silicone IOL-s suspected to favor bacterial adhesion and therefore having the higher risk of postoperative infections [18]. Silicone oil droplets adhere well to silicone IOL in patients with silicone oil tamponade used in retinal detachment or diabetic retinopathy surgery [19]. Therefore silicon IOL should not be implanted in highly myopic eyes in risk of retinal detachment.

Nowadays the silicone IOLs are less frequently used because they are not suitable for microincision cataract surgery (MICS).

There are also a light adjustable lens-two component silicone IOL where power is adjusted after implantation with UV-exposure in use [20, 21].

Glistenings can happen with silicone optics while the aqueous humor can penetrate the silicon material [12].

7. Hydrophobic foldable acrylic

Acrylic hydrophobic IOLs are modern foldable IOLs most widely used nowadays. They are designed of copolymers of acrylate and methacrylate derived from PMMA. The intention of the new design is to make the IOL foldable. They can be manipulated during the surgery and always turning back to its original shape [22] in a short period of time. First implanted IOL was in year 1993. Hydrophobic Foldable Acrylic can be of three piece and one piece design, with optic diameter 5.5–7 mm, and overall length 12–13 mm, transparent or colored—yellow. Refractive index can be 1.44–1.55.

Single and multi-piece hydrophobic IOLs can be implanted through small incision, not lover than 2.2 mm and have to be positioned properly since they have low self-centering ability. PCO is significantly lower than in PMMA IOLs but generally a bit higher for hydrophobic acrylic lenses compared with silicone [23].

They have higher incidence of photopsias than other acrylic IOLs because of high refractive index and low anterior curvatures and some of them develop glistenings since some are easily penetrated by aqueous humor but are not always clinically relevant unless when are dense or multifocal [24]. New materials of IOLs are prehydrated to equilibrium and will not accept further water, they are hydrophobic with the contact angle with water that of hydrophobic acrylic and are packaged in BSS to absorb the eventual water content before implantation [25].

8. Hydrophilic foldable acrylic

Hydrophilic foldable acrylic is a combination of hydroxyethylmethacrylate (polyHEMA) and hydrophilic acrylic monomer [26] material and it was introduced in 1980 with several modifications since. The IOLs made of this materials are usually single pieced and designed for capsular bag implantation. Refractive index of the material is 1.43, with water content ranging from 18 to 34% [27, 28].

They are soft, compressible with excellent biocompatibility for its hydrophilic surface. They can be implanted through a small incisions, lower than 2 mm and therefore ideal for MICS [29]. The folding of poly-HEMA chains depends on the level of hydration, and so the physical and optical properties of the polymer change as a function of water content. As the lenses hydrate, they absorb water and become soft and transparent.

The main disadvantage is the higher rate of optic opacification than in other materials and lower resistance for capsular bag contraction [30, 31].

9. The future of IOL s materials and designs

Considering the new knowledge and technological improvements and achievements, we can expect the new materials and designs of IOLs. In order to improve biocompatibility and refractive quality we expect some changes in shape of the IOLs (discoid, plate-lamellar, ball shaped) and therefore some novelties in implantation possibilities. The new neuro-ophthalmological knowledge and knowledge about adaptation and perception, industries based on robotic approach and innovations give us the right to expect some new and completely different IOLs in their shape, materials and functioning principle [32, 33]. In conclusion, in the future, we can expect some new, different and innovative approaches in the IOLs design and materials and refractive ophthalmology.

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