Journal of the Mechanical Behavior of Biomedical Materials 81 (2018) 168–172

Contents lists available at ScienceDirect

Journal of the Mechanical Behavior of Biomedical Materials

journal homepage: www.elsevier.com/locate/jmbbm

Effects of sterilization on the mechanical properties of poly(methyl methacrylate) based personalized medical devices

T.J.A.G. Münker[a,*], S.E.C.M. van de Vijfeijken[b], C.S. Mulder[a], V. Vespasiano[a], A.G. Becking[b], C.J. Kleverlaan[a], On behalf of the CranioSafe Group

[a] Dept. of Dental Material Sciences, Academic Centre for Dentistry Amsterdam (ACTA), Gustav Mahlerlaan 3004, 1081 LA Amsterdam, The Netherlands
[b] Dept. of Oral and Maxillofacial Surgery, Academic Medical Center (AMC), Meibergdreef 9, 1105 AZ Amsterdam, The Netherlands

CranioSafe Group

A.G. Becking[a], L. Dubois[a], L.H.E. Karssemakers[a], D.M.J. Milstein[a], S.E.C.M. van de Vijfeijken[a], P.R.A.M. Depauw[b], F.W.A. Hoefnagels[c], W.P. Vandertop[c], C.J. Kleverlaan[d], T.J.A.G. Münker[d], T.J.J. Maal[e], E. Nout[f], M. Riool[g], S.A.J. Zaat[g]

[a] Department of Oral and Maxillofacial Surgery, Academic Medical Center, University of Amsterdam, The Netherlands
[b] Department of Neurosurgery, Elisabeth-Tweesteden Hospital, Tilburg, The Netherlands
[c] Neurosurgical Center Amsterdam, Academic Medical Center, University of Amsterdam, The Netherlands
[d] Department of Dental Material Sciences, Academic Centre for Dentistry Amsterdam, The Netherlands
[e] 3D Laboratory of Oral and Maxillofacial Surgery, Academic Medical Center, University of Amsterdam, The Netherlands
[f] Department of Oral and Maxillofacial Surgery, Elisabeth-Tweesteden Hospital, Tilburg, The Netherlands
[g] Department of Medical Microbiology, Academic Medical Center, Amsterdam Infection and Immunity Institute, University of Amsterdam, The Netherlands

ARTICLE INFO

Keywords:
PMMA
Sterilization
Personalized medical device
PSI
Pre-treatment

ABSTRACT

Background: Nowadays, personalized medical devices are frequently used for patients. Due to the manufacturing procedure sterilization is required. How different sterilization methods affect the mechanical behavior of these devices is largely unknown.
Materials and methods: Three poly(methyl methacrylate) (PMMA) based materials (Vertex Self-Curing, Palacos R + G, and NextDent C&B MFH) were sterilized with different sterilization methods: ethylene oxide, hydrogen peroxide gas plasma, autoclavation, and γ-irradiation. Mechanical properties were determined by testing the flexural strength, flexural modulus, fracture toughness, and impact strength.
Results: The flexural strength of all materials was significantly higher after γ-irradiation compared to the control and other sterilization methods, as tested in a wet environment. NextDent C&B MFH showed the highest flexural and impact strength, Palacos R + G showed the highest maximum stress intensity factor and total fracture work.
Conclusion: Autoclave sterilization is not suitable for the sterilization of PMMA-based materials. Ethylene oxide, hydrogen peroxide gas plasma, and γ-irradiation appear to be suitable techniques to sterilize PMMA-based personalized medical devices.

1. Introduction

Poly(methyl methacrylate) (PMMA) has been widely used in different fields of healthcare. It is used as bone cement for fixation of knee and hip implants in orthopedics, as the base of dental prosthesis, for cranial reconstruction in neurosurgery, and for many other medical devices (Leggat et al., 2009). PMMA is light, radiolucent, cost efficient, and easy to use. However, it is associated with complications such as infection (Zanotti et al., 2016). The exothermic polymerization of PMMA can cause burn injuries if applied directly onto tissues and there are indications that residual monomers are toxic to the body (Leggat et al., 2009).

The mechanical properties of personalized medical devices are essential for long-term survival. These properties may be affected by

* Corresponding author.
E-mail address: t.j.a.g.munker@acta.nl (T.J.A.G. Münker).

https://doi.org/10.1016/j.jmbbm.2018.01.033
Received 5 January 2018; Received in revised form 25 January 2018; Accepted 28 January 2018
Available online 31 January 2018

storage time, pre-treatment, sterilization, and the location of the inserted medical device in the body. PMMA demonstrates increased flexibility in a liquid environment compared to a dry environment, and storage at 37 °C makes PMMA less resistant to fracture than storage at 21 °C (Hailey et al., 1994).

The most common sterilization methods for medical applications are ethylene oxide gas (EtO), hydrogen peroxide gas plasma (HPGP), autoclavation, and γ-irradiation (Yavuz et al., 2016). These sterilization methods are important as PMMA-based medical devices are not only prepared by powder and liquid mixing in the operating room, but prefabricated 3D-printed methacrylate-based materials and *ex vivo* polymerization are also used (Abdo Filho et al., 2011; Hassan et al., 2017; Sharavanan et al., 2015). The advantage of 3D-printing is a better control on the shape and material properties of the medical device. Manufacturing the medical device before surgery reduces surgical times and removes limitations to the environmental conditions during polymerization, enabling optimizations that may lead to better clinical outcomes. However, the device then needs to be sterilized, this presents a challenge to retain optimal material behavior.

The sterilization of PMMA powder is usually performed by γ-irradiation, except for Palacos, which is sterilized using EtO (Lewis, 1997). The liquid MMA monomer is sterilized through membrane filtration (Harper et al., 1997; Lewis, 1997, 1999; Lewis and Mladsi, 1998). γ-irradiation of PMMA results in chain scission, detectable through a decrease in molecular weight (Graham et al., 2000; Harper et al., 1997; Lee et al., 1999; Lewis and Mladsi, 1998). This directly influences mechanical properties such as fracture toughness, fatigue, and flexural strength (Graham et al., 2000; Harper et al., 1997; Lewis, 1999).

The effect of autoclave, EtO, and hydrogen peroxide (H_2O_2) sterilization on the chemical structure and surface morphology of PMMA is previously described (Yavuz et al., 2016). However, it is still unknown how these sterilization methods affect mechanical properties of cured PMMA. Therefore, the aim of this study is to investigate the effect of sterilization methods: EtO, HPGP, autoclavation, and γ-irradiation on the mechanical properties of PMMA-based personalized medical devices.

2. Materials and methods

The effects of sterilization with EtO, HPGP, autoclavation, and γ-irradiation on the mechanical properties of PMMA-based personalized medical devices were investigated (Table 1). Since the mechanical properties of the PMMA-based materials may vary depending on the application, three different types were investigated: Vertex Self-Curing, Palacos R + G, and NextDent C&B MFH (Table 2).

For each material the flexural strength, flexural modulus, fracture toughness, and impact strength were determined after sterilization and compared to the unsterilized control. All test methods for determining the mechanical properties were taken from the appropriate standards, e.g. ISO 20795–1:2013 and ISO 179–1:2010 (Standardisation, 2010, 2013).

Palacos R + G (Heraeus, Hanau, Germany) and Vertex Self-Curing (Vertex-Dental, Soesterberg, The Netherlands) were hand mixed and prepared according to the manufacturer's instructions. These specimens were molded using a stainless-steel mold. Curing of Vertex Self-Curing followed in a water-filled pressure cooker for ten minutes at 55 °C and 2.5 bar.

NextDent C&B MFH (NextDent, Soesterberg, The Netherlands) was 3D printed in a horizontal direction with a Rapidshape D30 (Rapidshape, Heimsheim, Germany) based on digital light processing (DLP). These specimens were washed in ethanol twice (three minutes and two minutes, respectively) under ultrasonic vibrations and dried for ten minutes prior to a 30 min post-cure in a LC3D-PrintBox (NextDent, Soesterberg, The Netherlands).

All specimens were wet grinded with standard metallographic grinding paper (P500, P1000 and P1200) and visually inspected for a smooth surface without porosities and irregularities. Sterilization was performed seven to ten days post-polymerization and the specimens were stored at least 72 h under standard laboratory climate conditions (22 ± 1 °C and 50 ± 2% humidity).

Table 1
Specifications of the sterilization methods (autoclavation, ethylene oxide (EtO), hydrogen peroxide gas plasma (HPGP), and γ-irradiation).

Sterilization Technique	Specifications	ISO norm
Autoclavation	121 °C for 16 min or 134 °C for 3.5 min	17665:2006
EtO	–	11135:2014
HPGP	Sterrad	11737:2006
γ-irradiation	26.4 – 29.4 kGy from Cobalt − 60	11137–1:2015

2.1. Flexural strength and flexural modulus

Eighteen series of ten rectangular specimens (64.0 ± 1.0 × 10.0 ± 0.2 × 3.3 ± 0.2 mm), one per material and sterilization method, were produced. The width and height of the specimens were measured by dial caliper before sterilization. After sterilization and prior to testing, the specimens were immersed in a water bath at 37.0 ± 1.0 °C for 50 ± 2 h. The flexural strength was tested in a water bath at 37.0 ± 1.0 °C, using a three-point-bending test (supporting bars span of 50.0 ± 0.1 mm) in a universal testing machine (Mecmesin Imperial 1000, West Sussex, UK) with a crosshead speed of 5.0 mm/min. Each specimen was tested until fracture or until the maximum curvature was reached. To calculate the ultimate flexural strength, σ, and the flexural modulus, E, Eqs. (1 and 2) were used.

$$\sigma = \frac{3Fl}{2bh^2} \tag{1}$$

$$E = \frac{F_1 l^3}{4bh^3 d} \tag{2}$$

where F is the maximum load exerted [N], l is the distance between the supports [mm], b is the width and h is the height of the specimen [mm], F_1 is the load at a point in the straight line portion of the load/displacement curve [N], and d is the deflection at load F_1 [mm].

2.2. Fracture toughness

Eighteen series of ten rectangular specimens (39.0 × 8.0 ± 0.2 × 4.0 ± 0.2 mm), one per material and sterilization method, were produced. The specimens were notched on the centerline with a sawing blade to a depth of 3.0 ± 0.2 mm. A pre-crack was made with a sharp blade with a thickness of 0.55 mm to a depth of 100 − 400 µm. An optical microscope was used to check the depth of the pre-crack. The width and height of each specimen was measured with a dial caliper. After sterilization and prior to testing the specimens were immersed in a water bath at 37 ± 1.0 °C for 7d ± 2 h, followed by a water bath at 23.0 ± 1.0 °C for 60 ± 15 min. The fracture toughness was measured using a three-point bending test (supporting bars span of 32.0 ± 0.1 mm) under dry conditions using the universal testing machine with a crosshead speed of 1.0 mm/min. The specimens were loaded until fracture. The maximum stress intensity factor, K_{max}, in MPa $m^{1/2}$ was calculated with Eq. (3).

$$K_{max} = \frac{f P_{max} l_t}{(b_t h_t^{3/2})} \times \sqrt{10^{-3}} \tag{3}$$

where P_{max} is the maximum load exerted on the specimen [N], h_t is the height and b_t is the width of the specimen [mm], l_t is the span [mm], and f is a geometrical function, dependent on x in Eq. (4), where a is the crack length consisting of the notch and the pre-crack [mm].