# Mechanical behaviour of p(HEMA) hydrogel for disc prosthesis on lumbar spine

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p(HEMA) - poly(hydroxyethyl methacrylate) - is a biocompatible material which in water forms a hydrogel and have multiple biomedical applications. The objective of this paper was to propose a biomaterial based on p(HEMA) for its use as damping element within disc prosthesis on lumbar spine and to analyse its mechanical behaviour. Intervertebral discs from healthy lumbar spine are mainly subjected to compressive stress and they have the capacity to ensure an adequate absorption of mechanical vibrations. In this study the mechanical behaviour of p(HEMA) has been evaluated by compression-decompression experimental tests on dry and wet biomaterial. The compression-decompression tests were performed with four different speeds: $v_1 = 0.05$ *mm/s*, $v_2 = 0.1$ *mm/s*, $v_3 = 0.15$ *mm/s* and $v_4 = 0.2$ *mm/s*. The results of this study highlighted the viscoelastic behaviour and damping capacity of swollen p(HEMA) hydrogel.

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*Keywords*: Biomaterial, Compression, Damping, Hydrogel, p(HEMA)

## 1. Introduction

Polymers based on 2-hydroxyethyl methacrylate (HEMA) (Fig. 1) are hydrophilic materials which after water absorption to an equilibrium state behave as elastic gel and in this form they are known as hydrogels. In dry state, p(HEMA) is a hard and brittle material; after swelling p(HEMA) becomes soft and flexible and can be easily cut with a scalpel or scissors [1-5]. p(HEMA) hydrogel is a biocompatible material successfully used in ophthalmology for contact lenses, artificial lens and artificial cornea, in orthopaedic for artificial cartilage and prosthesis of nucleus pulposus, and also in controlled drug delivery systems [3-5].

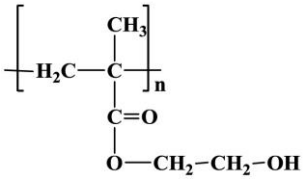

*Fig. 1. Chemical structure of p(HEMA)*

Mechanical properties, oxygen permeability and water content of p(HEMA) hydrogel can be controlled by copolymerization or cross-linking rate. The transparent hydrogels of p(HEMA) allow diffusion of liquids, suggesting the existence of a porous structure [1-4]. Damping capacity of hydrogel (as in case of natural cartilage) is associated with the squeeze effect of the fluid through the material micropores and with the viscous friction within the material [3-5].

Total disc prostheses currently used on lumbar spine are based on ball-and-socket design and are made of CoCrMo-on-CoCrMo (Cobalt-Chromium-Molybdenum alloy) and CoCrMo-on-UHMWPE (ultra high molecular weight polyethylene). These prostheses have good endurance and durability but do not provide the adequate absorption (damping) of mechanical vibrations that may occur in the lumbar spine during daily activities. Also, the metal-on-polyethylene devices do not reproduce the complex viscoelastic behaviour of the natural intervertebral disc. The main stress supported by intervertebral disc or disc prosthesis is the axial compression [6]. Thus, the investigation of mechanical behaviour of the hydrogel should consist mainly in compressive testing.

The purpose of this paper was to study a simple case of vibrational stress – the compression-decompression mechanical behaviour – on a biomaterial based on p(HEMA). Experimental tests on mechanical behaviour of the p(HEMA) material from this study have included compression-decompression tests on dry and wet polymer with four different speeds. In swollen state this biomaterial is proposed as damping element within intervertebral disc prosthesis on lumbar spine.

## 2. Materials and methods

The biopolymer based on HEMA used in this study was in form of preforms of transparent artificial lens (Corneal Industrie, France) (Fig. 2). This biomaterial was chosen because in wet (swollen, hydrated) state is a hydrogel used as artificial cartilage with properties similar to natural cartilage and particularly to natural intervertebral disc.

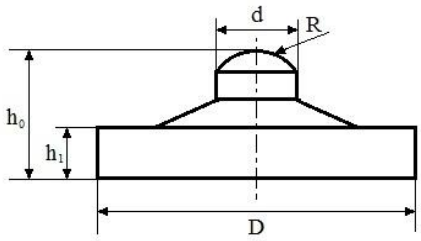


*Fig. 2. p(HEMA) transparent lens – lateral view [7].*

The compactness of these convex lenses is *0.71* [7]; this value is closed to bovine articular cartilage [8]. Maximum degree of swelling for p(HEMA) lenses in distilled water or physiological serum can be reached after *48 h* [4-5]. In this study, the swelling of p(HEMA) lens and its testing were realised in physiological serum.

The dimensions and the masses of p(HEMA) convex lenses in dry and wet state are listed in Tables 1 and 2. The radius of convex lens was determined from surface topographic analysis using the Taylor Hobson roughness tester (Laboratory of Machine Elements, IMMR Department, Faculty of Mechanical Engineering, Iasi, Romania). From these tables it can be concluded that not only the size and mass of hydrogel have changed due to liquid absorption but also the roughness of spherical cap [7].

*Table 1. Dimensions of p(HEMA) lens [7].*

| Material | Height $h_0$ (mm) | Height $h_1$ (mm) | Diameter $D$ (mm) | Diameter $d$ (mm) |
|---|---|---|---|---|
| Dry p(HEMA) | 3.005 | 1.218 | 12.410 | 6.046 |
| Wet p(HEMA) | 3.315 | 1.310 | 13.691 | 6.672 |

*Table 2. Mass, radius and roughness of convex lens [7].*

| Material | Mass $m$ (g) | Radius $R$ (mm) | Roughness $R_a$ (μm) |
|---|---|---|---|
| Dry p(HEMA) | 0.259 | 7.967 | 0.0334 |
| Wet p(HEMA) | 0.338 | 8.775 | 0.0264 |

In order to study the damping capacity of dry and wet p(HEMA) hydrogel, the experimental compressive tests have been realised. The tests have been performed using the CETR UMT-2 microtribometer (Laboratory of Machine Elements, IMMR Department, Faculty of Mechanical Engineering, Iasi, Romania) equipped with a sensor having the working range of *0.2 - 20 N* and a resolution of *1 mN* (Fig. 3). A special cylindrical indenter of *20 mm* diameter (Fig. 3 and Fig. 4) have been realised and used for these compressive tests. Experimental tests have been performed at room temperature of *25° C*.

Compressive experimental tests have included both the compression phase and decompression phase followed by a relaxation time. The maximum height of convex lens was considered as reference for zero position of the carriage; this position was set before each test. The vertical displacement of carriage for the compression phase was from position *0* to *0.5 mm* and for the decompression phase was from *0.5* to *0 mm*.

Compression and decompression have been performed with four different constant speeds: $v_1 = 0.05$ *mm/s*, $v_2 = 0.1$ *mm/s*, $v_3 = 0.15$ *mm/s* and $v_4 = 0.2$ *mm/s*. Constant speed condition is not satisfied at the beginning and the end of the two phases because the carriage acceleration (which depends on size of the loading speed) takes about *0.5 s* (Fig. 5).

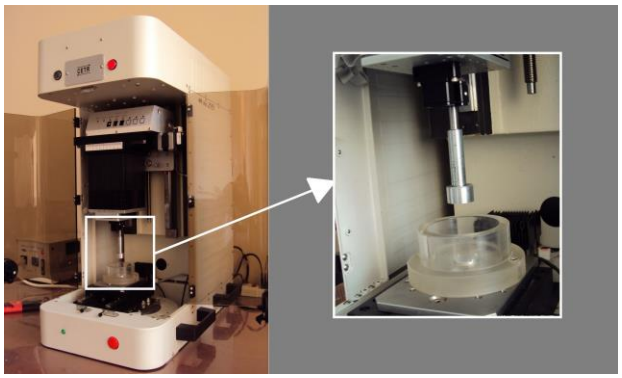

*Fig. 3. Microtribometer CETR UMT-2 adapted with a cylindrical indenter [7].*

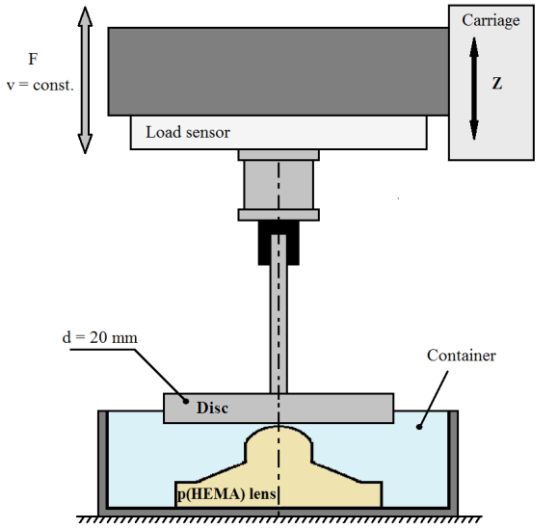


*Fig. 4. Experimental setup.*

## 3. Experimental results

The results of compression-decompression experimental tests which was performed at four different speeds ($v_1 = 0.05$ *mm/s*, $v_2 = 0.1$ *mm/s*, $v_3 = 0.15$ *mm/s* and $v_4 = 0.2$ *mm/s*) and with a maximum deformation of *0.5 mm* on dry and wet p(HEMA) convex samples, are presented in Figs. 5-13. As expected, the load has increased with depth, time and speed during the

compression phase, while the load has decreased with the indenter withdrawal (rising), with time and speed during the decompression phase.

The maximum load at compression of wet p(HEMA) has increased with the four indenter speeds (Fig. 6, Fig. 8 and Fig. 10); while the maximum load at compression of dry p(HEMA) has remained constant at the four different speeds (Fig. 7, Fig. 9 and Fig. 11). These have indicated that the p(HEMA) hydrogel have a significant viscoelastic behaviour and a damping capacity only in swollen state. Also, Fig. 6 and Fig. 7 show that the area enclosed by the hysteresis loop has increased with the loading speed.

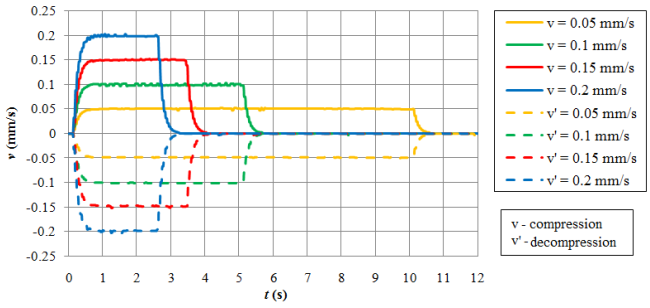


*Fig. 5. Variation in speed with time for wet/dry p(HEMA) lens.*

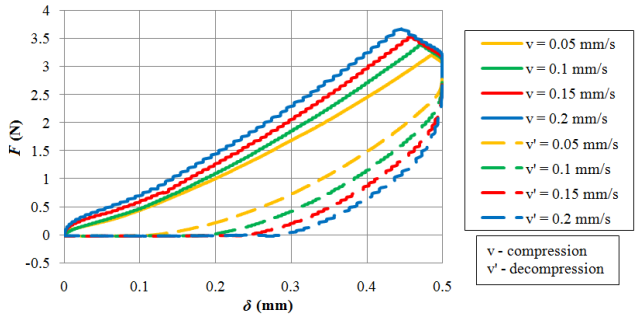


*Fig. 6. Variation in load with depth at different speeds for wet p(HEMA) lens.*

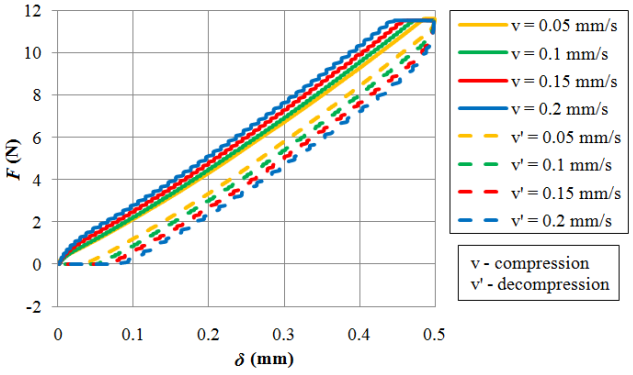


*Fig. 7. Variation in load with depth at different speeds for dry p(HEMA) lens.*

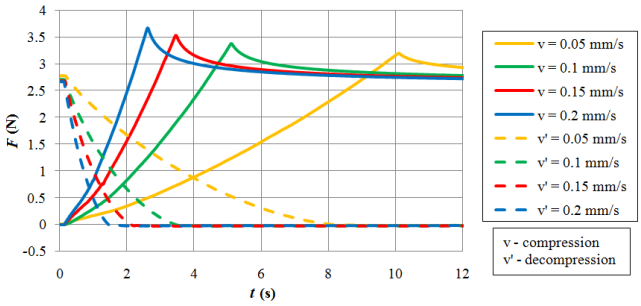


*Fig. 8. Variation in load with time at different speeds for wet p(HEMA) lens.*

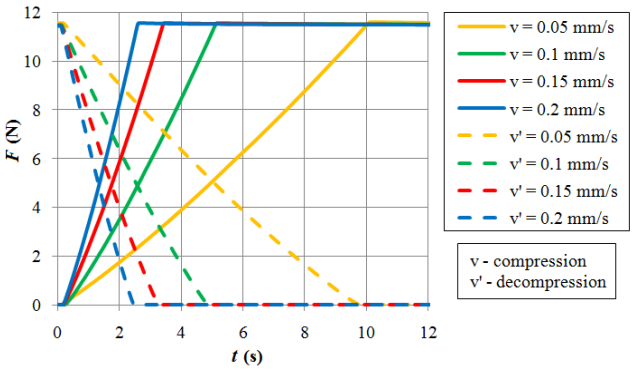


*Fig. 9. Variation in load with time at different speeds or dry p(HEMA) lens.*

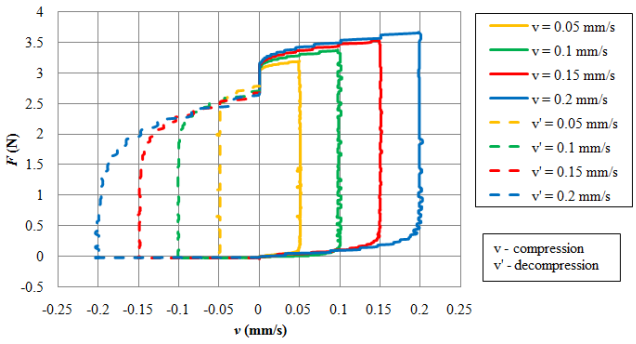


*Fig. 10. Variation in load with speed for wet p(HEMA) lens.*

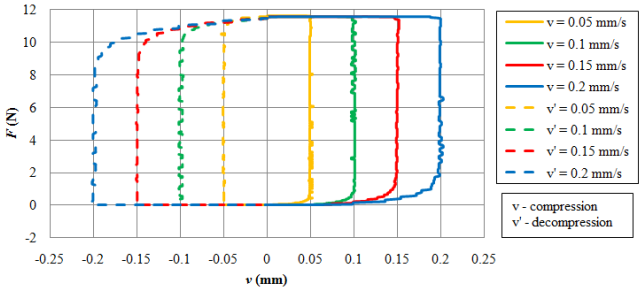


*Fig. 11. Variation in load with speed for dry p(HEMA) lens.*

The total mechanical work of deformation (total deformation energy) at compression-decompression when $\delta = (0 - 0.5)$ *mm* can be determined with the formula:

$$L = \int_0^{0.5} F(\delta) \cdot d\delta \tag{1}$$

p(HEMA) hydrogel from an implanted disc prosthesis will remain only in swollen state due to the existence of the interstitial fluid within the intervertebral space [9]. Thus to investigate de deformation energy, this paper had taken into account only the swollen p(HEMA). In order to determine the deformation energy at compression and decompression of swollen p(HEMA), the sixth order polynomials have been fitted to the data points of load-depth curves. The equations of load-depth curves at compression with the four different speeds for swollen hydrogel are:

$$\begin{aligned}
F_1(\delta) &= -7512.9\delta^6 + 10365\delta^5 - 5343.7\delta^4 + 1261\delta^3 - 126.14\delta^2 + 8.4163\delta + 0.0145 \\
F_2(\delta) &= -12309\delta^6 + 16952\delta^5 - 8765.3\delta^4 + 2094.2\delta^3 - 220.01\delta^2 + 12.932\delta - 0.028 \\
F_3(\delta) &= -14860\delta^6 + 20544\delta^5 - 10727\delta^4 + 2611.7\delta^3 - 286.18\delta^2 + 16.851\delta + 0.0121 \\
F_4(\delta) &= -14381\delta^6 + 19756\delta^5 - 10299\delta^4 + 2521.1\delta^3 - 281.94\delta^2 + 18.142\delta + 0.0218
\end{aligned} \tag{2}$$

The equations of load-depth curves at decompression with the four speeds for swollen hydrogel are:

$$\begin{aligned}
F_1(\delta) &= 1322.1\delta^6 - 1526.4\delta^5 + 586.83\delta^4 - 72.718\delta^3 + 9.8022\delta^2 - 0.5761\delta - 0.0176 \\
F_2(\delta) &= 5300.2\delta^6 - 7133.6\delta^5 + 3504.7\delta^4 - 737.34\delta^3 + 67.767\delta^2 - 2.2172\delta - 0.0169 \\
F_3(\delta) &= 8516.4\delta^6 - 11846\delta^5 + 6134.3\delta^4 - 1413.2\delta^3 + 141.15\delta^2 - 4.7447\delta - 0.0172 \\
F_4(\delta) &= 9492.8\delta^6 - 13158\delta^5 + 6856.9\delta^4 - 1614.6\delta^3 + 165.91\delta^2 - 5.7365\delta - 0.0126
\end{aligned} \tag{3}$$

After the integration of the load-depth curves equations (2)-(3), the total deformation energy at compression and decompression of swollen p(HEMA) with four different speeds has been determined. Fig. 12 shows that the deformation energy at compression has increased with speed, while the deformation energy at decompression has decreased.

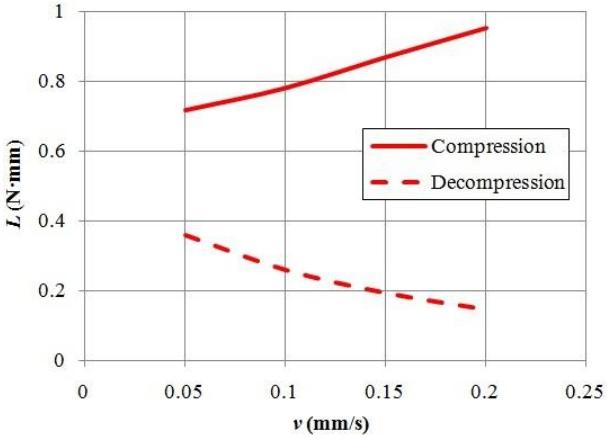


*Fig. 12. The variation of deformation energy with speed for swollen p(HEMA) lens*

The area enclosed by the hysteresis loop (the area between the load-depth curves for compression and decompression) represents the lost energy on a vibration cycle (compression-decompression) or the mechanical work consumed by viscous friction inside the material. This viscous friction is due to the fluid expulsion through the hydrogel micropores. The lost energy can be calculated with the following formula:

$$E = L_{compression} - L_{decompression} \tag{4}$$

Fig. 13 shows that the lost energy on a vibration cycle has increased with the indenter speed.

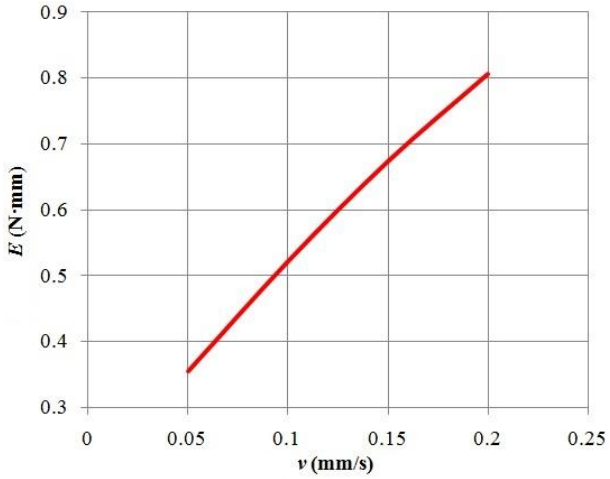


*Fig. 13. The variation of lost energy with speed for swollen p(HEMA) lens*

## 4. Discussions

To validate the experimental results from this study, the ex-poro-hydrodynamic (XPHD) lubrication theory developed by Pascovici's team for sphere-on-plan configuration has been used. Modelling of porous medium in XPHD was done with Kozeny-Carman (KC) law for permeability variation [10,11]:

$$\phi_{KC} = \frac{D(1-C)^3}{C^2} \tag{5}$$

where $\phi$ - material permeability, $C$ - material compactness, $D$ - complex parameter of material: $D = \frac{d_f^2}{16k}$, $d_f$ - fibber diameter of porous material and $k$ is a correction constant, $k = (5 - 10)$.

The analytical model used for validation of experimental results from compression of swollen p(HEMA) was the analytical solution for squeeze effect using the KC law at constant speed, in case of sphere-on-plan configuration, with using of first term from Taylor series approximation [10]:

$$F_{KC} = -\frac{\pi\eta R^2 h_0{}^2 C^2}{2D(1-C)^2}\frac{(1-H)^2}{H}\frac{dH}{dt} \tag{6}$$

where $F_{KC}$ - theoretical load, $\eta$ - viscosity of physiological serum, $R$ - radius of spherical cap, $H$ - adimensional thickness, $H = \frac{h}{h_0} = \frac{h_0 - \delta}{h_0}$ and $h$ - lens thickness, $h_0$ - initial thickness (height), $\delta$ - material deformation (depth) and $v = -\frac{dh}{dt} = -h_0\frac{dH}{dt}$ is speed. Then equation (6) becomes:

$$F_{KC} = \frac{\pi\eta R^2 h_0 C^2 v}{2D(1-C)^2}\frac{(1-H)^2}{H} \tag{7}$$

Since the permeability and the complex parameter vary with the speed [10], to calculate the theoretical load it was considered an average fibber diameter $d_f = 0.35 \cdot 10^{-3}$ *mm*, a complex parameter $D$ of $0.9 \cdot 10^{-9}$ $mm^2$ and $1.5 \cdot 10^{-9}$ $mm^2$ for speed of *0.05 mm/s* and respectively *0.1 mm/s*. The viscosity of physiological serum was $\eta = 1$ *mPa·s* $= 10^{-9}$ $N \cdot s/mm^2$, the initial thickness was $h_0 = 3.315$ *mm*, lens radius $R = 8.775$ *mm*, compactness $C = 0.71$ and the contact configuration was plan-on-deformable sphere where lens geometry was considered completely spherical (Fig. 14).

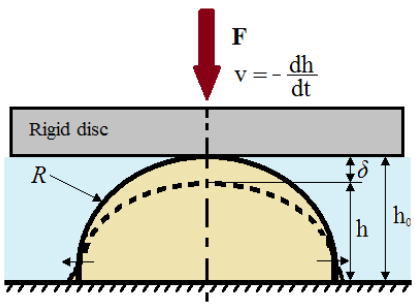


*Fig. 14. The geometry of plane-on-deformable sphere contact for squeeze effect*

Fig. 15 highlights the validation of experimental results with analytical model developed by the Pascovici's team [10,11]. It can be observed that the experimental results from compression with different two speeds were closed to the theoretical.

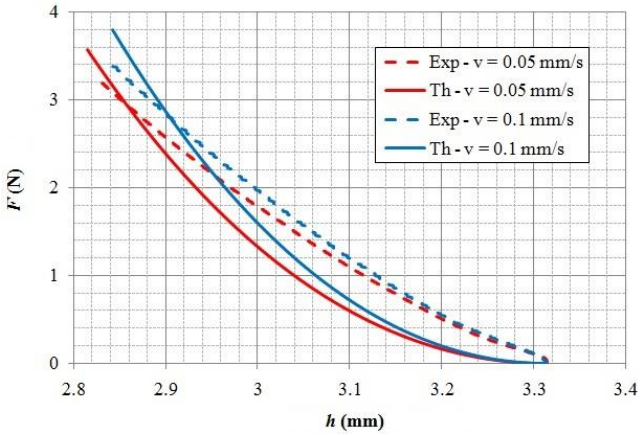


*Fig. 15. Variation in load with height for wet p(HEMA) – comparison of experimental and theoretical results*

It is worth mentioning that the analytical model developed by the Pascovici's team [10,11] was achieved and applied for sphere-on-plane configuration where the sphere was represented by a rigid spherical indenter and the plan by a deformable porous layer. However, in this paper, the surface of deformable porous material that has been tested was spherical while the indenter tip was plan, indicating that this model is valid for both sphere-on-plane and plane-on-sphere configurations.

Considering the obtained results from the present study the authors suggest that these p(HEMA) hydrogels that are used for contact lenses and artificial cartilage can also be used in total lumbar disc arthroplasty as a mechanical vibrations absorber. In comparison with the experimental tests realised in this study where the hydrogel is submerged into a container with physiological serum, in vivo the absorption and expulsion of liquid would be much slower due to the biological enclosed space and viscosity variation of interstitial fluid.

The mechanical and structural characteristics of hydrogels based on p(HEMA) can be controlled to become very close to the natural joint cartilage (particularly to the natural intervertebral disc) which presents a good damping capacity; for example, the values of the elasticity modulus, oxygen permeability and water content can be controlled to be close [3-5,7].

As reported in the literature [12-14], over the past years many efforts have been made in designing novel intervertebral disc prostheses based on p(HEMA) hydrogels, semi-interpenetrating polymer networks (s-IPNs) and composite hydrogels. The interesting results already obtained using p(HEMA) and p(HEMA)-based s-IPNs composite hydrogels as biomimetic intervertebral

disc prostheses have indicated similar viscoelastic properties to natural intervertebral disc.

## 5. Conclusions

Intervertebral discs from healthy lumbar spine are mainly subjected to compression stress and have the capacity to ensure an adequate absorption of mechanical vibrations. Total disc prostheses currently used have good endurance and durability but do not provide adequate damping. The objective of this paper was to test the compression-decompression mechanical behaviour of a hydrogel based on p(HEMA) in order to propose it as intermediate damping element within total disc prosthesis on lumbar spine.

The experimental results showed that the maximum load, which the swollen hydrogel responded at compression-decompression with different testing speeds, has depended on the deformation speed; this can be explained through its significant viscoelastic behaviour which is close related to liquid expulsion and absorption through the material pores; and therefore it suggests a certain damping capacity. Also, the lost energy on a vibration cycle within swollen p(HEMA) increased with speed.

Future research will include additional experimental tests to determine the variation of damping coefficient with speed and to study the fatigue resistance of the p(HEMA) hydrogel.

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