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The determination of mechanical properties of prosthetic liners through experimental and constitutive modelling approaches

Określenie właściwości mechanicznych linerów ortopedycznych z wykorzystaniem eksperymentu i modelowania konstytutywnego

Abstract

The aim of the study was the estimation of the ability of hyperelastic material models for the fitting of experimental data obtained in the tensile testing of silicone liners used in lower-limb prosthetics. Three groups of liners were analysed: I – silicone liner, II – part of the liner in which the silicone has a fabric reinforcement, III – silicone liner with an outer covering material. Both longitudinal and circumferential samples were taken. The Neo-Hookean, Mooney-Rivlin and Ogden parameters of constitutive models of hyperelastic materials were calculated.

Keywords: prosthetic liners, constitutive models, hyperelastic material, tensile test

Streszczenie

Celem badań była ocena przydatności modeli materiałów hipersprężystych do dopasowania danych doświadczalnych uzyskanych w próbie rozciągania dla silikonowych linerów ortopedycznych stosowanych w protezach dolnych. Przeanalizowano trzy grupy: I – liner silikonowy, II – liner silikonowy z wewnętrznym wzmocnieniem, III – liner silikonowy z zewnętrznym wzmocnieniem. Wyróżniono dwa kierunki pobrania próbek: podłużny i obwodowy. Zidentyfikowano parametry określonych funkcji modeli konstytutywnych materiałów hipersprężystych: Neo-Hookean'a, Mooney-Rivlin'a i Ogden'a.

Słowa kluczowe: linery protetyczne, modele konstytutywne, materiał hipersprężysty, próba rozciągania

1. Introduction

When observing the development of prosthetic technology, it can be seen that for a long time, a significant problem has been the biomechanical compatibility of a lower limb prosthetic with the residual limb - together, these constitute a new appendage after an amputation [11]. The direct contact of the patient's living tissues with the materials used for the comprising the prosthesis remains an important issue with regard to creating optimal solutions. A limb, as a new appendage which meets the load-bearing function requires characteristics aimed at eliminating, or at least limiting, the concentration of loads localised in sensitive areas. A prosthetic socket is one of the most important elements of the lower limb prosthesis. It transfers the load to the limb and therefore must meet the strength requirements with respect to the loads for which it is designed to transfer [3].

Today, the closest that is achieved to the ideal solution is to provide prosthetics with innovative sockets and silicone liners. Modifications to the liners has led to the development of a new type of suspension system of the prosthesis which provides comfortable attachment to the person's residual limb, and with care and soothing additions, this has a positively influence on the skin of the limb [2, 14]. Due to the good see above note adhesion of the silicone liner to the residual limb, this causes the reduction of friction presented on the skin; thus, it is often called the 'second skin'. Silicone liners may be used alone in the prosthesis as comfortable inserts or as an element of a total-surface bearing fit suspension for patients with a higher degree of mobility. There are three commonly used solutions using silicone liners in the suspension systems: shuttle lock, suction and vacuum [4, 6]. These solutions provide: comfort; the reduction of vertical movements of the limb in the socket, which stabilizes the walk; increases in the mobility of the person. The person's comfort depends on the precision of the socket and selection of an appropriate fit [2, 13]. Therefore, the modelling of problems of strength as an engineering tool is an important issue that enables the simulation of the behaviour of materials and structures in the selected load conditions. One of the methods of analysis for the selection of parameters is a static tensile test which provides the experimentally obtained material constants. This data provides the possibility to predict the stress characteristics at different levels of deformation. This can be a measure of the usage comfort evaluation of the product and the criterion of durability with regard to repeated attachment and removal. It can also be potentially useful for assessing prosthetic compatibility by providing quantitative information on the similarities and differences of these products.

The aim of this study was to evaluate selected mechanical characteristics obtained during the static tensile testing of longitudinal and circumferential samples taken from two types of the prosthetic liners (M and W). The results of tests were used to identify the parameters of the constitutive equations of selected hyperelastic models (the Neo-Hookean, the Mooney-Rivlin and the Ogden) to determine the material constants of the tested liners. The tested materials exhibit non-linearity and anisotropy; however, the assumption was made that its tensile behaviour can be predicted with the use of hyperelastic material models. From the point of view of clinical applications, the tested liners should show similar mechanical behaviour to skin. For skin,

which is an anisotropic material, hyperelastic material models are used in the modelling of the mechanical behaviour with assumption of its isotropy [16]. We can hypothesise that data from uniaxial stretching tests can be used in the simulation and modelling of the material parameters of orthopaedic liners – this is important from the point of view of obtaining a better match between the properties of the liners and the properties of the skin on the residual limbs. The usefulness of the hyperelastic material models for silicone liners was thus investigated in this study.

2. Material and methods

Samples were prepared from two models of orthopaedic liners produced by the leading manufacturers of prosthetic supplies, MediLiner FIRST® (M) and Willow Wood® Express AKLiner (W) (Fig. 1). The circumference dimension at a distance of 4 cm from the distal point is in the range of 32–47 cm, and the proximal dimension at the proximal end at a distance of 30 cm from distal point is in the range of 44–80 cm. The total length of the liners was approx. 370 mm.
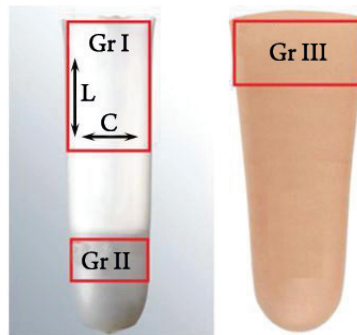

Fig. 1. The locations from where samples were taken: groups I and II from the liner M, group III from the liner (W); the directions of sampling have been marked (for all groups the same)

The first group (I) of samples was cut from the proximal part of the liner (M). The second group (II) of samples was cut from the distal part of the liner, where the silicone had a fabric reinforcement (M+). The third group (III) of samples was cut from the second type of liner (W). For all three groups (I–III), longitudinal (L) and the circumferential (C) samples were taken. The locations and orientations of samples taken are shown in Fig.1. All samples had the same dimensions: length 100 ± 1 mm and the width 10 ± 0.1 mm; however, these were of different thicknesses. The average thickness, the average cross-sectional area and the number of samples are presented as average values with standard deviations in Table 1.

For the prepared samples, the mechanical properties under static tension were determined with the use of the Instron 4465 testing machine with a force sensor of 5 kN. The samples were mounted using flat clamps and they were extended at a speed of 25 mm/min. The test was carried out in room conditions at a temperature of 22 ± 1°C. The measurement base of the samples was 60 ± 1 mm. The calculated values of maximum stress, stretch and energy of deformation are shown as the arithmetic average values with a standard deviation of $(X \pm SD)$.

Table 1. The characteristic parameters of samples

Liner type	M		M+		W	
Group of samples	G I		G II		G III	
Direction of taken	C $n = 4$	L $n = 4$	C $n = 3$	L $n = 3$	C $n = 4$	L $n = 4$
Average thickness (SD) [mm]	2.45 (0.11)	2.39 (0.04)	4.49 (0.13)	5.60 (0.05)	3.99 (0.02)	3.94 (0.03)
Average cross-sectional area [mm²]	24.5	23.9	44.9	56.0	39.9	39.4

For all tested groups of samples, the analysis of experimental data was conducted by plotting the stress (σ) against the stretch (λ). The average experimental curves were then determined and on their basis, model analyses were carried out. In the process of fitting the tension experimental data with the use of constitutive hyperelastic laws, OriginPro 7.5 software was applied.

3. Constitutive material models

To describe the non-linear stress versus stretch relationship of hyperelastic materials, several constitutive models can be employed [1, 2]. These models assume that hyperelastic material behaviour may be characterised by the strain energy density function (W) expressed in terms of the three invariants of the strain Cauchy-Green tensor I_1, I_2, I_3, given as (1):

$$W = f(I_1, I_2, I_3) \tag{1}$$

Hyperelastic materials are special kinds of elastic material. For many engineering materials, including especially vulcanized elastomers, linear elastic models have poor accuracy in predicting the non-linear behaviour of the material.

The dependence of stress-strain for the material is referred to as isotropic non-linear elastic – it is independent on the strain rate. It is assumed that the material is also non-compressible. The response to the load applied on an unfilled silicone often shows the behaviour of a perfectly hyperelastic material. For filled elastomers, as well as for soft tissues, hyperelastic models are also used with the assumption of homogeneity of the material [7, 16].

The three invariants of the stretch tensor I_1, I_2, I_3, are given as (2), (3), (4):

$$I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \tag{2}$$

$$I_2 = \lambda_1^2\lambda_2^2 + \lambda_2^2\lambda_3^2 + \lambda_3^2\lambda_1^2 \tag{3}$$

$$I_3 = \lambda_1^2\lambda_2^2\lambda_3^2 \tag{4}$$

Thus, equation (1) can be given by formula (5), where C_{ijk} are the material constants:

$$W = \sum_{(i+j+k=1)}^{\infty} C_{ijk}(I_1 - 3)^i(I_2 - 3)^j(I_3 - 1)^k \tag{5}$$

Considering the conditions of the uniaxial tension of incompressible materials, the principal stretches λ_i are describe by the formula (6) and (7):

$$\lambda_1^2 \lambda_2^2 \lambda_3^2 = 1 \tag{6}$$

$$\lambda_1 = \lambda, \lambda_2 = \lambda_3 = \frac{1}{\sqrt{\lambda}} \tag{7}$$

therefore, the expression of the invariants of the stretch tensor become simplified and it is written as the principal stretch (8):

$$I_1 = \lambda^2 + 2\frac{1}{\lambda}, I_2 = 2\lambda + \frac{1}{\lambda^2}, I_3 = 1 \tag{8}$$

In the literature, hyperelastic models (Ogden, Money-Rivlin and Neo-Hookean) are widely described in order to determine the relationship of the stress-stretch of hyperelastic materials such as polymers, rubber-like materials or biological tissues [1, 7, 8, 12, 16].

Neo-Hookean material model

For the Neo-Hookean model, the strain energy density function is expressed as the principal stretch function (9), and the stresses are given by equation (10):

$$W = C_1(I_1 - 3) \tag{9}$$

$$\sigma_{11} - \sigma_{33} = 2C_1\left(\lambda^2 - \frac{1}{\lambda}\right) \tag{10}$$

Mooney-Rivlin material model

The strain energy density function for the Mooney-Rivlin model is expressed as the principal stretch function in the form (11):

$$W(\lambda_1, \lambda_2, \lambda_3) = C_1(I_1 - 3) + C_2(I_2 - 3) \tag{11}$$

therefore, for the uniaxial tensile, the stresses can be expressed as (12):

$$\sigma_{11} - \sigma_{33} = 2C_1\left(\lambda^2 - \frac{1}{\lambda}\right) - 2C_2\left(\frac{1}{\lambda_2} - \lambda\right) \tag{12}$$

The necessary and sufficient conditions for to be positive are given by the inequalities $C_1 \geq 0; C_2 \geq 0$ [16].

Ogden material model

The strain energy density function for the Ogden model is given as (13):

$$W(\lambda_1,\lambda_2,\lambda_3)=\frac{2\mu}{\alpha^2}\left(\lambda_1^\alpha+\lambda_2^\alpha+\lambda_3^\alpha-3\right)\tag{13}$$

where α and μ are the material constants: the shear modulus and the strain hardening exponent [2].

For the uniaxial tension conducted by the load in the direction of the long axis, the nominal stress can be given as (14):

$$\sigma_{11}-\sigma_{33}=\frac{2\mu}{\alpha}\left(\lambda^{\alpha-1}-\lambda^{-1(\alpha/2)}\right)\tag{14}$$

In order to compare the quality of the fit of a theoretical model to experimental data, the coefficient of determination R^2 is defined by formula (15):

$$R^2=\frac{\sum_{i=1}^{n}\left(\hat{y}_i-\overline{y}\right)^2}{\sum_{i=1}^{n}\left(y_i-\overline{y}\right)^2}\tag{15}$$

where y_i is the actual value of the variable, is the theoretical value of the variable on the basis of models, is the arithmetic mean average of the experimental value of the variable.

4. Results

On the basis of the tensile test, the stress-stretch curves for both circumferential and longitudinal samples for all three groups were obtained. For each group, the average stress-stretch curves were determined and the analysis of the results was performed. Circumferential samples taken from liners had lower values of maximum stress (Table 2). These values were: for group I (liner M), 0.15 MPa at a stretch of 3.08; for group II (liner M+ with reinforcement in the distal part), about 0.2 MPa at a stretch of 2.88; for group III (liner W) 0.16 MPa at stretch of 3.11. In the case of the longitudinal samples, the values were: for group I (M), the maximum stress was 0.16 MPa at a stretch of 3.23; for group II (M+) it was 0.23 MPa at a stretch of 2.36; in group III (W), the stress reached almost 1.54 MPa and the stretch was 1.57. The higher stress values were obtained for the knitted fabric reinforced silicone in comparison to the silicone without reinforcement for the longitudinal samples (Fig. 2).

Group I showed no significant differences in the stress values between the circumferential and longitudinal samples confirming the homogeneous composition of the liner without reinforcement in the proximal part. The addition of a knitted fabric reinforcement in the distal part of the liner had an influence on the tensile test results, raising the tensile stress and varying the slope of the stress-stretch curve. The orientation of the samples (whether circumferential or longitudinal) in groups with the fabric-reinforced silicon (groups II and III) also significantly influenced the stress-strain curves (Fig. 3). For the circumferential samples, the curve had a lower slope and a larger stretch. In group II, this was about 25% of the value relating to the longitudinal samples; in group III, it was over 40%.

Table 2. The characteristic mechanical parameters of tested liners

Liner type	M		M+		W	
Group of samples	G I		G II		G III	
Direction of taken	C	L	C	L	**C**	**L**
Maximum stress [MPa]	0.15 (0.02)	0.16 (0.01)	0.20 (0.003)	0.23 (0.01)	0.16 (0.04)	1.54 (0.54)
Stretch at maximum stress [−]	3.08 (0.17)	3.23 (0.05)	2.88 (0.11)	2.36 (0.02)	3.11 (0.14)	1.57 (0.54)
Energy of deformation [10^{-3}J] (SD)	252.61 (33.72)	297.62 (37.48)	547.81 (21.98)	400.46 (24.26)	380.93 (80.05)	711.34 (57.05)

a)

b)

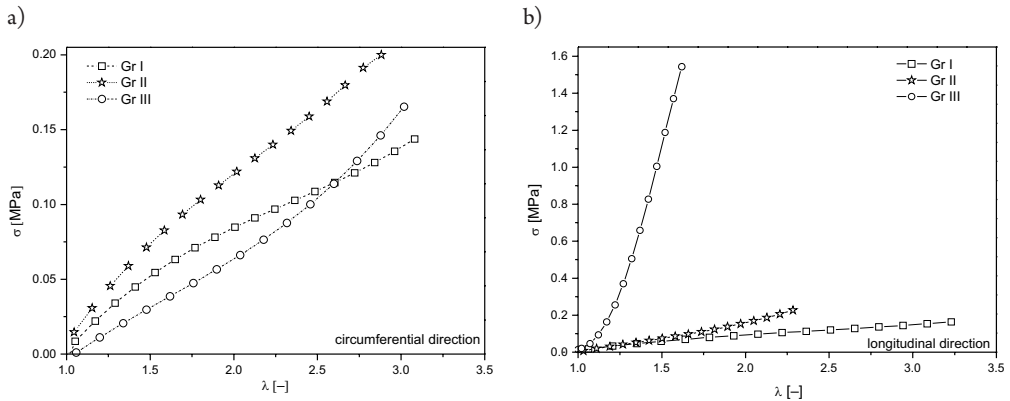Fig. 2. Stress (σ) – stretch (λ) representative curves: a) circumferential, b) longitudinal

This confirmed the importance of the weaving direction in a fabric reinforcement for a silicone liner.

A highly accurate fit can be observed between the models and experimental stress when analysing the constitutive modelling (Fig. 4).

Analysis of the determined material constants in the constitutive modelling revealed strong agreement of values of the shear modulus (μ) obtained by the Mooney-Rivlin and the Ogden models in groups I and II for both circumferential and longitudinal samples and

in group III for circumferential samples (Tables 4 and 5). A strong fit between experimental data and these models is also confirmed in Figure 4. The exception was the shear modulus (μ) for group III in the longitudinal samples. For the Mooney-Rivlin model, it was a value of 0.67 MPa, and for the Ogden model, it was 0.397 MPa. The Neo-Hookean model was the least compatible with the experimental data – the constant μ was almost 2.5 times lower in the first group for both circumferential and longitudinal samples, and in the second group, for circumferential samples (Table 3). Knowing the requirements of orthopaedic surgeons and therapists, an orthopaedic device can be designed in such a way that will support the residual limbs, e.g. by controlled pressure.
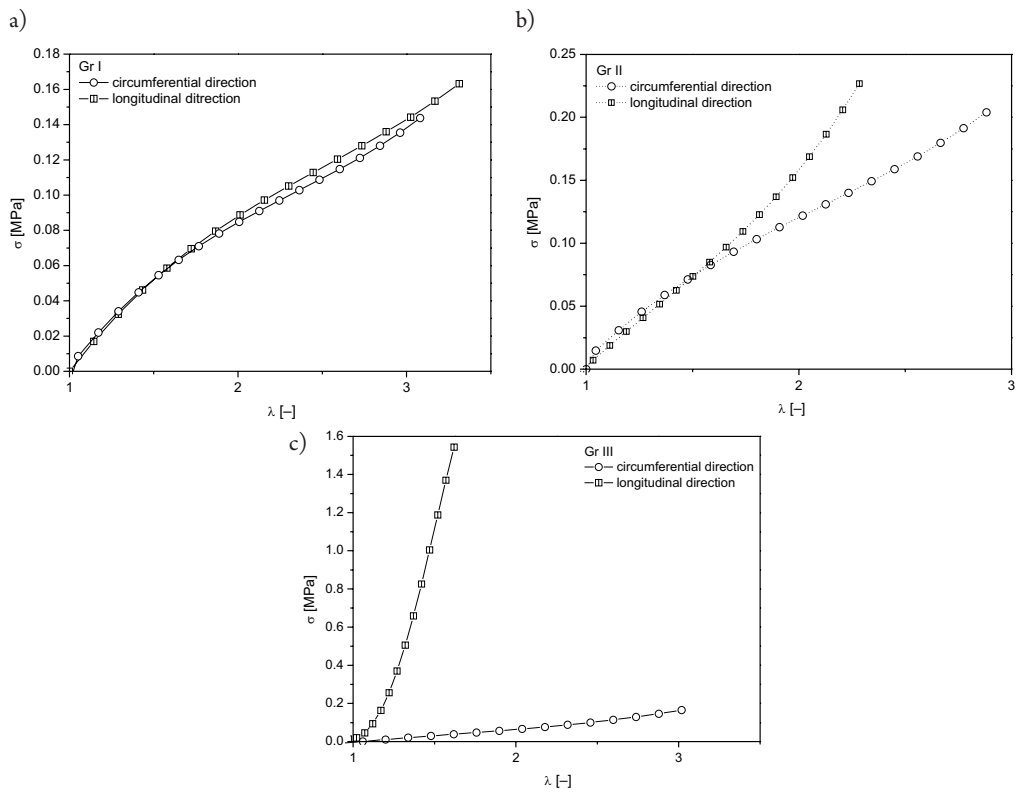
a)

b)

c)

Fig. 3. Comparison of stress-stretch (σ-λ) curves for three groups of samples for both circumferential and longitudinal samples
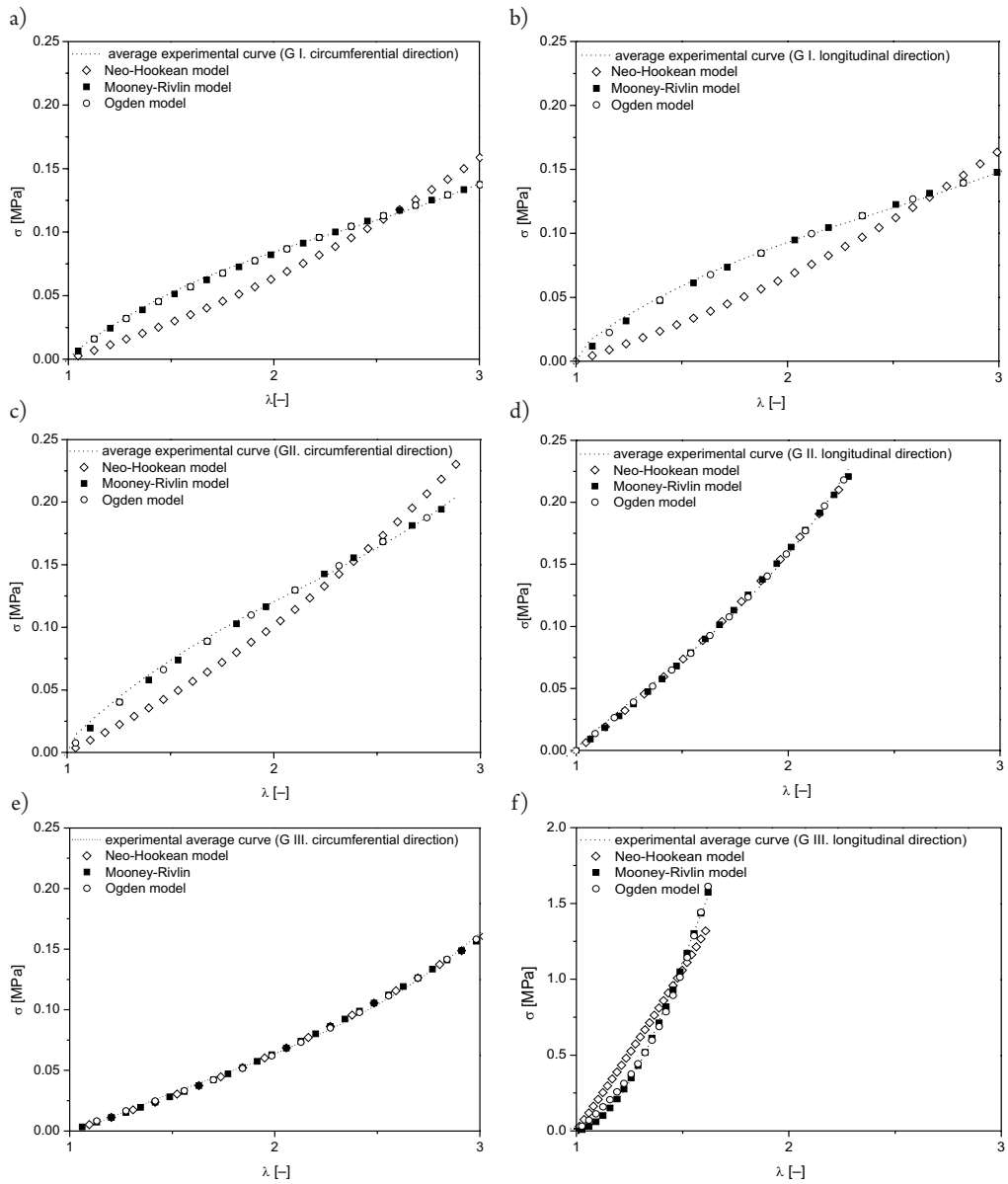
Fig. 4. Fitting to the tension data using three hyperelastic material models

Table 3. Parameter values for the Neo-Hookean model

Group of samples	Gr I (M)		Gr II (M+)		Gr III (W)	
Direction of samples taken	C	L	C	L	C	L
C_1 [MPa]	0.0091 ±0.0004	0.0095 ±0.0005	0.0145 ±0.0006	0.0230 ±0.0001	0.0091 ±0.0001	0.3354 ±0.1637
R^2	0.8394	0.8017	0.9133	0.9981	0.9975	0.9319
μ [MPa]	0.018	0.019	0.029	0.046	0.018	0.671

Table 4. Parameter values for the Mooney-Rivlin model

Group of samples	Gr I (M)		Gr II (M+)		Gr III (W)	
Direction of samples taken	C	L	C	L	C	L
C_1 [MPa]	0	0	0.0014 ±0.0004	0.0230 ±0.001	0.0081 ±0.0004	0.3354 ±0.1878
C_2 [MPa]	0.0238 ±0.0006	0.0259 ±0.0011	0.0320 ±0.0011	0	0.0021 ±0.0001	0
R^2	0.9986	0.9955	0.9979	0.9981	0.9981	0.9318
μ [MPa]	0.048	0.052	0.067	0.046	0.020	0.670

Table 5. Parameter values for the Ogden model

Group of samples	Gr I (M)		Gr II (M+)		Gr III (W)	
Direction of samples taken	C	L	C	L	C	L
μ [MPa]	0.049 ±0.006	0.056 ±0.001	0.067 ±0.001	0.053 ±0.001	0.024 ±0.001	0.397 ±0.030
α	1.9392 ±0.0245	1.8691 ±0.0378	2.1128 ±0.0382	3.4016 ±0.0346	3.1253 ±0.0466	6.3136 ±0.2704
R^2	0.9986	0.9975	0.9978	0.9993	0.9986	0.9923

5. Discussion

Eshragi et al. [4] investigated the pressure distribution in orthopaedic liners for the lower leg and demonstrated that the highest values occur from the anterior, followed by the posterior and the lateral and the medial (0.02–0.12 MPa). Our research indicates that the pressure generated remains in the safe range of loads for the strength of the liners (0.15–1.6 MPa).

Analysis of the results of the experiment demonstrates the non-linear nature of the load-deformation curves. As with the work of Ali et al. [1], similar values of the model constants for the Ogden and the Mooney-Rivlin models were obtained. Comparing the levels of model fit with the experimental data, a strong agreement of both the Mooney-Rivlin and the

Ogden models can be observed for all analysed cases across the whole range of considered stretch [5, 7]. By contrast, the Neo-Hookean model showed a strong fit in only two cases –for liner (M+) for longitudinal samples, and for liner (W) for circumferential samples. These samples of silicone were reinforced with fabric, for which the weaving direction was in line with the direction of the applied load. However, in the case of samples taken from liner (W) for longitudinal samples, a weaker fit was evident for the experimental curve for the Ogden model in the range of stretch between 1 and 1.25. Comparing the obtained values of the model parameters with the study conducted by Martins et al. [7], similar values of C_1 for the Noe-Hookean and C_1 and C_2 for the Mooney-Rivlin model for the group III in the longitudinal direction can be seen. The Ogden model with six parameters (C_1–C_6) makes it difficult to compare values. In the investigation of Ali et al. [1], the application of the constitutive models of hyperelastic materials for rubber was also made. However, they obtained nominal stress values of 25 MPa and nominal strain above 6; furthermore, a visible mismatch in the graph of the Neo-Hookean and the Mooney-Rivlin models can be seen. This makes the results difficult to discuss. However, the weakest fit of the Neo-Hookean model can be confirmed.

Sasso et al. [15] made a comparison and numerical simulations using a multi-parameter model for different compressions and the uniaxial and biaxial tensile load cases (the Ogden with six parameters, the Mooney-Rivlin with five parameters); again, observing the weakest fit with the Noe-Hookean and a better fit with the multi-parameter models.

Shergold et al. [16] used the Mooney-Rivlin and the Ogden models to analyse silicon Sil8800 and B452 as well as pig skin tissue by studying uniaxial stretching and compression at different strain rates (0.004, 0.4, 40. 4000/s^{-1}). They reported: μ = 0.4–7.5 and α of 1.2 for pig skin; μ = 0.4–2.8, α = 3, C_1 = 0.5 and C_2 = 0 for B452; μ = 2.1–8.0, α = 2.5, C_1 = 1.0 and C_2 = 0.9 for Sil8800; C_1 = 0.3 and C_2 = 0 for human skin. They indicated that the value of constant μ depends on the strain rate. It is considered that the value of α is related to the geometric evolution of the collagen network, and is independent on strain rate. In contrast, the resistance to rearrangement by deformation of the collagen fibers influence on value of μ and these deformation mechanisms are sensitive to the strain rate. The results were analysed in a context comparable to human skin parameters [16]. The substitute materials of human skin (rubber-like materials and pig skin tissues) are often used in medical engineering; therefore, knowing the basic mechanical properties of these substitutes is important for the comfort of the person and for strength requirements.

6. Conclusions

The mechanical properties of the rubber-like material can be described by constitutive models of hyperelastic materials, such as the Mooney-Rivlin and the Ogden models, at low, medium and large deformations, despite the fact that they are represented by the energy density function based on the principal strain invariants (the Mooney-Rivlin) and on three principal stretches (the Ogden). For two of the three analysed constitutive models

of hyperelastic materials, a strong fit of tension experimental data (correlation coefficient $R^2 > 0.99$) was obtained. For the Noe-Hookean model, a weaker fit was observed.

The procedure used to optimise the material parameters for three models has been successfully used for silicone materials designed for orthopaedic liners. Silicone liners are made from materials which exhibit a hyperelasticity with elastic stress-strain dependence for high deformation levels – they also exhibit a strong non-linear relationship. Hyperelastic material models can be used to determine the non-linear properties of the RTV-silicone and TPE liners with the assumption that material is homogenous, incompressible and isotropic.

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