24th COBEM - 2017

24th ABCM International Congress of Mechanical Engineering
December 3-8, 2017, Curitiba, PR, Brazil

COBEM-2017-1682
AN ANALYSIS OF DEFORMATION IN AN ARTERIOVENOUS FISTULA MODEL

Jonhattan Ferreira Rangel
Kleiber Lima de Bessa
Federal University of Rio Grande do Norte. Department of Mechanical Engineering, Center of Technology, Avenida Senador Salgado Filho, 300 – Lagoa Nova, CEP 59078-970 – Natal/RN
jferreirarangel01@gmail.com
klbessa@ct.ufrn.br

Willyam Brito de Almeida Santos
willyambas@gmail.com

Sandi Itamar Schafer De Souza
sandi@ufrnet.br

Valquíria Bomfim Fernandes
valbfernandes@hotmail.com

Abstract. *The number of Patients with chronic renal failure (CRF) has have been rising drastically in recent years. Given the need for treatment of these patients arteriovenous fistula (AVF) is recognized by several researchers as being the best access for treatment by hemodialysis. After the creation of the fistula, there is an increase in the blood flow in this region, resulting in new loads on the walls of the vessels, which may cause some pathological problems. Through mathematical modeling this work related the properties between a real AVF and a silicone one in order to obtain an equivalent thickness in the artificial fistula whose deformations are proportional to the ones suffered by the real vessel. The treatment and cleaning of the computerized tomography of a patient with fistula, the 3D printing and the preparation of the molds for the manufacture of fistulas in paraffin and later the fistulas in silicone were carried out. After the tests, we obtained the deformations for three different regions of the AVF and compared with the results obtained by mathematical models. The initial experimental results, for some regions, were approximated to the results of some theoretical models, however due to the disparity between the theoretical and experimental result, there is still the need to improve the methodology.*

Keywords: *Chronic renal failure, arteriovenous fistula, artificial blood vessels, stress-strain*

1. INTRODUCTION

In recent years there has been a significant increase in the number of patients with chronic renal failure undergoing hemodialysis (Amorim *et al.*, 2013; Noubiap, 2015; Pippias, 2015). The hemodialysis treatment can only be maintained for a long period of time primarily by maintaining vascular access (Sivanesan *et al.*, 1999). Several papers suggest arteriovenous fistula as the best vascular access for patients on hemodialysis treatment, because it is a durable access and with low complication rates (Krzanowski *et al.*, 2011; Briones *et al.*, 2010; Akoh, 2009).

However, with the creation of the fistula there is an increase in blood flow in the vessels that compose it, resulting in new loadings in them. The values of the shear stresses and high pressure may cause structural damage to the vessel walls. Studies have also shown that up to 60% of AVF do not mature to the point of use (Dember, 2011).

The elasticity of the arteries has been studied and correlated with the maturation of the AVF (Sorace *et al.*, 2011; Sorace *et al.*, 2012; Paulson, 2014). The arterial composition has been shown to be a strong indicator of vascular disease, cardiovascular disease and renal failure. The arterial hardening is caused by a change in the ratio of collagen to elastin in the extracellular matrix of the arterial tunica media (Faury, 2001).

The shear stress acting on endothelial cells regulates its proliferation, gene expression, lipid composition and metabolism (Chien, 2007; Johnson *et al.*, 2011). Changes in blood pressure may create an acute or chronic mechanical stimulus in the form of circumferential stretching of these cells. Some studies have shown that the mechanical stimuli of

different intensities is detected by mechanoreceptors on the cell surface, allowing the conversion of external mechanical stimuli into biochemical signals in the cell. The stimulation associated with physiological functioning is important in the maintenance of vascular homeostasis. Nevertheless, the high pressure that occurs with hypertension exposes the cells to excessive mechanical load which can lead to pathological consequences (Jufri *et al.*, 2015; Kaunas *et al.*, 2005).

The present work analyzed the deformations in the wall of the artificial fistula due to the application of a certain pressure range, trying to relate the mathematical and physical modeling between the real and silicone fistulas.

2. METHODOLOGY

The study used computer tomography images of a patient with arteriovenous fistula. From this, silicone fistulas were manufactured and mechanical deformations were measured after application of certain loads.

2.1 Data Acquisition and Processing

The medical images were obtained by computed tomography of a patient with AVF provided by Onofre Lopes University Hospital of the Federal University of Rio Grande do Norte. They were reconstructed and treated using InVesalius software (CTI, Campinas, São Paulo, Brazil).

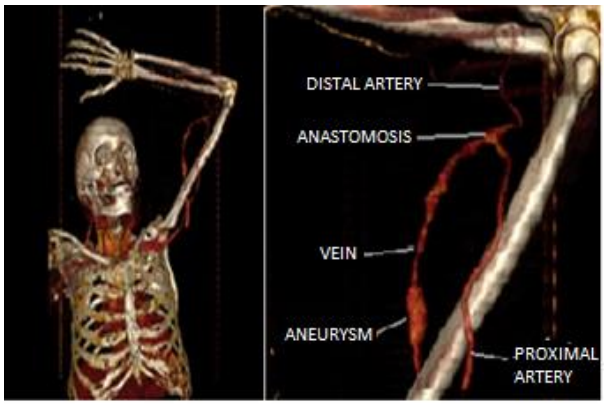

Figure 1. Three-dimensional reconstruction of AVF

Computed tomography is a noninvasive method of medical imaging in which a part of the body is sectioned and examined using a three-dimensional scanner. Due to the high number of sections the reconstruction in a 3D model required a high computational processing power, thus, it was decided to reduce the number of sections and execute the three-dimensional reconstruction.

When decreasing the number of sections, problems occurred in the virtual mesh. Open points in the structure and surface irregularities were corrected with the help of Autodesk Meshmixer software (San Rafael, CA, USA).

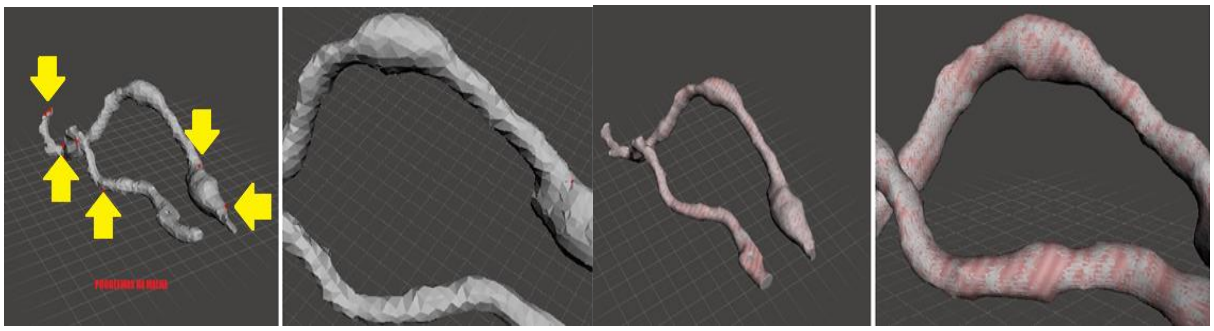

Figure 2. Mesh Correction

2.2 3D printing

After the process of treating and improving the surface of the virtual model, the 3D impression of the solid model of the AVF was executed. The spacing of 0.1 mm between the print threads was chosen, thus achieving a better surface quality of the model.

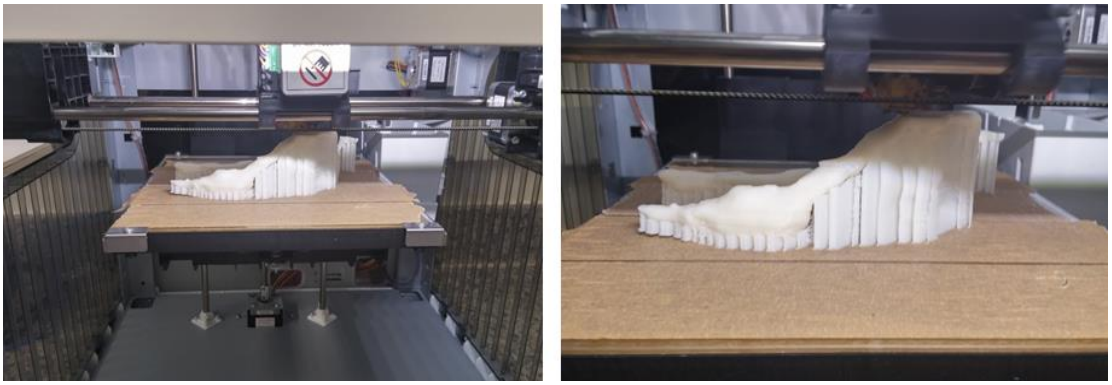

Figure 3. 3D Printing

2.3 Manufacture of molds

A bipartite mold was fabricated from the solid model of the AVF, which served as the basis for the manufacture of paraffin fistulas. See the figure below:

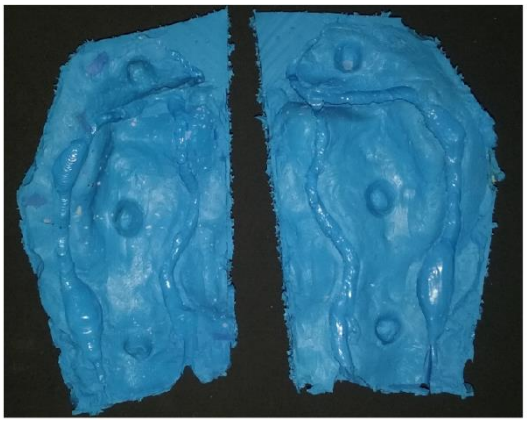

Figure 4. Bipartite mold

The mold was fabricated from elastomeric material in order to facilitate the demolding and manufacturing of paraffin AVFs.

2.4 Mathematical modeling

Due to the flow and pressure conditions present in the blood vessels, longitudinal and circumferential deformations develop therein. The forces that balance these deformations develop internally in the wall of the vessel, where T is the force used to balance the circumferential deformations and F_r to balance the longitudinal deformations. Observe the equations:

$$T = p_i \cdot r_i \tag{1}$$

$$F_R = F_T + p_i \cdot \pi \cdot r_i \tag{2}$$

Considering the blood vessel as a straight tube of circular cross section and applying the Young-Laplace equation for cylindrical tubes gives the following equations for stress:

$$\bar{\sigma}_{\theta\theta} = \frac{p_i \cdot r_i}{h} \tag{3}$$

$$\bar{\sigma}_{zz} = \frac{F_T}{\pi \cdot h \cdot (r_e - r_i)} + \frac{p_i \cdot r_i}{2 \cdot h} \tag{4}$$

Where p_i is the internal pressure in Pa developed inside the vessels, r_i the internal radius, r_e the external radius and h the thickness of the vessel wall, in meters.

Considering that the deformations occur in the elastic region, we can adopt the following equation:

$$\sigma = E \cdot \varepsilon \tag{5}$$

Assuming that the artificial blood vessels have the same mechanical characteristics when submitted to the real loads, we obtain the following equation for the AVF thickness.

$$h_{silicone} = \frac{E_{vessel}}{E_{silicone}} \cdot h_{vessel} \tag{6}$$

The mean values for the modulus of elasticity and thickness of vessel wall were obtained from Sorace *et. al.* (2012), E_{vessel} = 138,4 kPa e h_{vessel} = 228μm (n = 75 patients).

The following equations were used for the calculation of the theoretical values for the deformations, having as parameters the applied pressure and the geometry of the model.

Lamé solution: Thick-walled pressure vessels:

$$\bar{\sigma}_{\theta\theta} = \frac{\left(\frac{b^2}{r^2}+1\right)}{\left(\frac{b^2}{a^2}-1\right)} \cdot p_i + \frac{\left(\frac{b^2}{a^2}+\frac{b^2}{r^2}\right)}{\left(\frac{b^2}{a^2}-1\right)} \cdot p_0 \tag{7}$$

$$\bar{\sigma}_{rr} = -\frac{\left(\frac{b^2}{r^2}-1\right)}{\left(\frac{b^2}{a^2}-1\right)} \cdot p_i - \frac{\left(\frac{b^2}{a^2}-\frac{b^2}{r^2}\right)}{\left(\frac{b^2}{a^2}-1\right)} \cdot p_0 \tag{8}$$

Where b is the distance from the center of the vessel to the outer surface, a is the distance from the center to the inner surface, r the median radius, or the distance from the center to the desired point of tension. The consideration for using the analysis for thick walled vessels is that D / h <10.

The deformations were also analyzed by the method used in thin-walled vessels, which closely resembles the method demonstrated by Fung (1993), as follows:

$$\bar{\sigma}_{\theta\theta} = \frac{p_i \cdot r_i}{h} \tag{9}$$

$$\bar{\sigma}_{rr} = \frac{p_i \cdot r_i}{2 \cdot h} \tag{10}$$

2.5 Manufacture of silicone AVF

The artificial fistula was fabricated by means of painting silicone layers on the paraffin models of fistulas, in order to maintain a constant thickness in all regions of the fistula, thus guaranteeing the structural homogeneity of the vessel.

After painting and checking the thickness, the fistulas were allowed to cure and the paraffin was removed, leaving only the silicone vessel.

2.6 Deformation analysis

The analysis of the deformation was done in the Laboratory of Metrology (Technological Nucleus - UFRN). The silicone AVF was connected to a calibration module and pressures were applied through a syringe and the deformations were measured with a caliper. Observe the image below:

Figure 05: Workbench

The sections selected for analysis of the deformation were as follows: lesser aneurysm (I), greater aneurysm (II) and in the artery (III). These sections were chosen because they are where the greatest deformations were expected to be present, both mathematically and physiologically. Note the image:

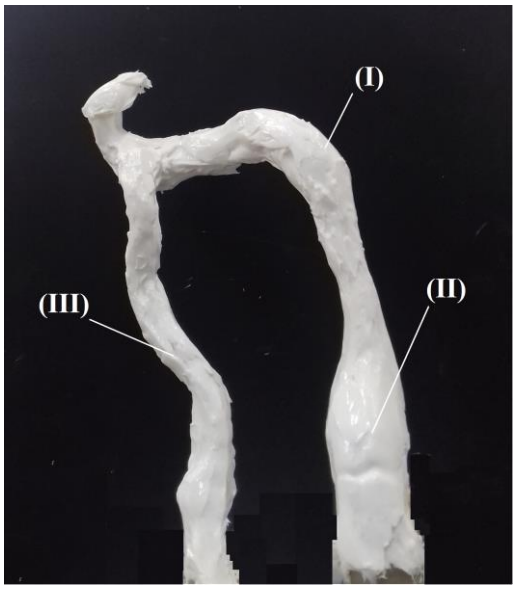

Figure 6. Sections of analysis

3. RESULTS AND DISCUSSION

The thickness to which the silicone fistulas should be manufactured was initially determined. For this analysis, the system was considered isentropic and with small deformations. The values assumed for the calculation were mean values obtained from a population of patients with Arteriovenous Fistula.

The following values were used for the material (silicone) and the real vessel: E_{vessel} = 138,4 kPa, h_{vessel} = 228 μm from Sorace *et. al.* (2012) (number of patients n = 75 patients), and $E_{silicone}$ = 7,5 MPa (Coser, 2009). For analysis, it was considered that the diameters and pressure in the real vessel and in the silicone vessel were the same. Thus, the mean value for the wall thickness of the silicone vessel was obtained:

$h_{vessel\_silicone}$ = 4,20736 μm

After cleansing and improvement of the mesh, execution of the printing and fabrication of the mold, we obtained the AVF in paraffin. Paraffin was chosen because it would be easy to remove after the silicone coating, thus facilitating the manufacture of silicone fistulas.

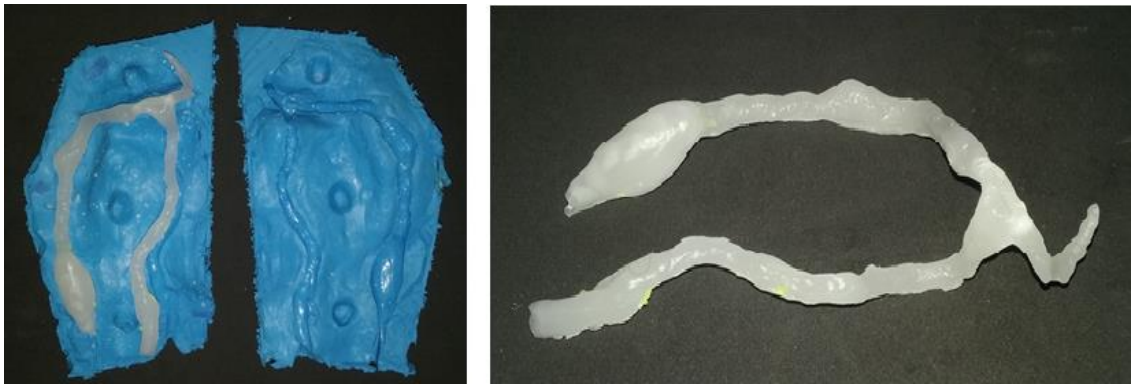

Figure 7. Manufacture of paraffin fistulas

The making of the fistula with the pre-determined thickness was not possible with the available equipment, since the micrometric thickness would require extremely high precision. It was then proposed to make a fistula with a thickness of 2 mm, the construction of which would be more plausible.

The creation of another bipartite template with an increase of 2 mm in the diameter of the entire AVF was then attempted. It was not successful because the cure time of the silicone was too high inside the mold and the mold material did not allow its placement inside a furnace to accelerate the curing process of the fistula.

It was then decided to create the fistula by means of painting. The silicone paint was spread over the entire AVF, seeking to maintain homogeneity in all sections. The continuous section was not achieved due to the difficulty of the painting process, the geometry of the fistula and the high viscosity of the silicone. Observe the manufactured fistula:

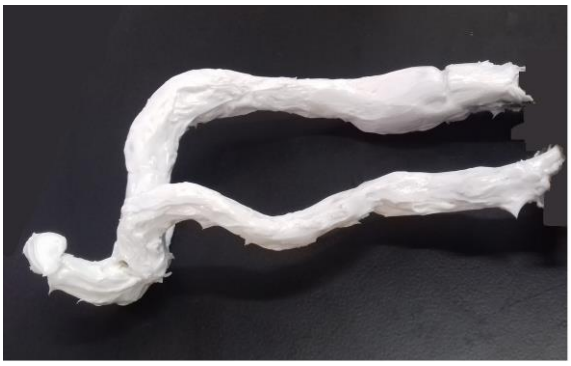

Figure 8. Silicone arteriovenous fistula

From the tests, the following results for the deformations were obtained:

Table 1. Comparison between experimental and theoretical results based on the methods of Thick Walls, thin-walled vessels and the description of Fung (1993) for deformation in the smallest aneurism (section (I)).

Pressure load (mmHg)	Experimental Deformation	Theoretical Deformation – Thick Walls	Theoretical Deformation – Thin Walls	Theoretical Deformation – (Fung (1993))
20	0.005914243	0.03390584	0.02847551	0.033640186
40	0.010349926	0.035714632	0.029205652	0.034693642
60	0.020206999	0.038224011	0.029935793	0.036025672
80	0.021192706	0.03994829	0.030665934	0.037029617
100	0.0340069	0.042956271	0.031396076	0.038526884
120	0.048792509	0.046540482	0.032126217	0.040210265

Table 2. Comparison between experimental and theoretical results based on the methods of Thick Walls, thin-walled vessels and the description of Fung (1993) for deformation in the largest aneurism (section (II)).

Pressure load (mmHg)	Experimental Deformation	Theoretical Deformation – Thick Walls	Theoretical Deformation – Thin Walls	Theoretical Deformation – (Fung (1993))
20	0.013888889	0.034930656	0.052502009	0.045561909
40	0.015681004	0.036253144	0.053848214	0.046837899
60	0.017921147	0.037629883	0.055194419	0.048146882
80	0.035394265	0.040557722	0.056540625	0.050424136
100	0.042114695	0.042491547	0.05788683	0.052059018

Table 3. Comparison between experimental and theoretical results based on the methods of Thick Walls, thin-walled vessels and the description of Fung (1993) for deformation in the artery (section (III)).

Pressure load (mmHg)	Experimental Deformation	Theoretical Deformation – Thick Walls	Theoretical Deformation – Thin Walls	Theoretical Deformation – (Fung (1993))
20	0.008683068	0.006116017	0.017019181	0.012244154
40	0.027496382	0.00669566	0.01745557	0.012787479
60	0.034732272	0.007328822	0.01789196	0.013361864
80	0.047033285	0.008190015	0.018328349	0.014109234
100	0.046309696	0.008550296	0.018764738	0.01442462
120	0.056439942	0.009309912	0.019201127	0.015054437

For sections (I), (II) and (III) the following mean of thicknesses and diameters were obtained, respectively: h(I) = 0.00298 m, h(II) = 0.00264 m, h(III) = 0.00372 m, D(I) = 0.01224 m, D(II) = 0.019992794 m, D(III) = 0.009132196 m. A disparity was observed between the values obtained by the selected analytical methods and the experimental ones. The three selected methods come from an analysis of the Laplace differential equations, whose approximations have a range of errors greater than 4% for D / h <10 (Hibbeler, 2011). Another important factor was the use of a caliper for the measurement of deformation. While this instrument is relatively accurate, it was not ideal for this experiment, since the pressure applied to the walls of the fistula during the measurement can alter the results due to the elasticity of the material and vessel geometry.

Another important fact to be taken into account in analysis of the results is the question of the properties of the material. It was observed that the initial experimental values approximated the theoretical values for some models this may be due to the fact that the Young's modulus was approximated to Coser, 2009. We may believe that the strain-strain curve of the silicone to which the fistula are constituted may not resemble of the silicone that the data were used.

4. CONCLUSION

Based on the results a disparity was seen between the actual and analytical values, possibly due to the problems previously described (equipment with an inadequate range of precision, error due to the intrinsic approximations in the equations, non-uniform geometry of the vessel section, varying diameter, among others). These divergences can be corrected with the use of more precise equipment and the manufacture of an AVF with homogeneous thickness. This equipment is currently being designed.

Another important fact to emphasize is the applicability of this study to prototypes that will be used in vivo. While the theoretical and experimental values may not match each other, the work is developing a methodology for analysis in non - rigid vessels, which is not restricted to use in vessels and biomedical devices, but can be applied in the analysis of non-rigid tubes (PVC, Rubber and high elastic polymers for example) for transporting fluids.

Projects involving computational simulations were and are being developed based on this AVF and this work can serve as validation for these simulations. Two experimental flow benches are being developed, which are to be linked to the fistula in order to measure deformations. Through this, it will be possible to obtain data that is closer to reality, as well as use it for several other aspects such as analysis involving pressure, recirculation, wall tension among others.

5. REFERENCES

Akoh, J.A., 2009. “Prosthetic arteriovenous grafts for hemodialysis”. *The journal of vascular access*, Vol. 10, p. 137-147.

Amorim, P., Sousa, G., Vieira, J., Sousa, L., Ribeiro, K., Sobrinho, G., Vieira, T., Meireles, N., Fortes, A., Neves, F., Albino, P., 2013. “Complications of vascular access for hemodialysis - Limits, imagination and commitment”. *Revista portuguesa de cirurgia cardio-toracica e vascular : orgao oficial da Sociedade Portuguesa de Cirurgia Cardio-Toracica e Vascular*, Vol. 20, p. 211-219.

Briones, L., Diaz Moreno, A., Sierre, S., Lopez, L., Lipsich, J., Adragna, M., 2010. “Permanent vascular access survival in children on long-term chronic hemodialysis”. *Pediatric nephrology*, Vol. 25, No. 9, p. 1731-1738.

Chien, S., 2007. “Mechanotransduction and endothelial cells homeostasis: the wisdom of the cell”. *American Journal of Physiology. Heart and Circulatory Physiology*, Vol. 292, p. H1209-H1224.

Coser, E., 2009 “Caracterização da Borracha de silicone utilizada em Isoladores para Rede Elétrica”. *Dissertation (Dissertation in Engineering)* – UFRS. Porto Alegre, p. 73.

Dember, L, M:, 2011. “Fistulas First-But Can They Last?”. *Clinical Journal of the American Society of Nephrology*, Vol. 6, p. 463-464.

Faury, G., 2001. “Function-structure relationship of elastic arteries in evolution: From microfibrils to elastin and elastic fibres”. *Pathologie-Biologie*, Vol. 49, p. 310–325.

Fung, Y.C., 1993. *Biomechanics: Mechanical Properties of living Tissues*. Springer, San Diego, 2nd edition.

Hibbeler R. C., 2011. *Mechanics of Materials*. Prentice Hall, New York, 8 nd edition.

Humphrey, J. D., 2001. *Cardiovascular Solid Mechanics: Cells, Tissues, and Organs*. Springer, Texas, 1nd edition.

Johnson, B.D., Mather, K.J., Wallace, J.P., 2011. “Mechanotransduction of shear in the endothelium: basic studies and clinical implications”. Vascular Medicine, Vol. 16, p. 365-377.

Jufri, N.F., Mohamedali, A., Avolio, A., Baker, M.S., 2015. “Mechanical stretch: physiological and pathological implications for human vascular endothelial cells”. *Vascular cell*, Vol. 7, p. 1-12

Kaunas, R., Nguyen, P., Usami, S., Chien, S., 2005. “Cooperative effects of Rho and mechanical stretch on stress fiber organization”. *Proceedings of the National Academyof Sciences of the United States of America*, Vol. 102, p. 15895-15900.

Krzanowski, M., Janda, K.., Chowaniec, E., Sulowicz, W., 2011. “Hemodialysis vascular access infection and mortality in maintenance hemodialysis patients”. *Przeglad lekarski*, Vol. 68, p. 1157-1161.

Noubiap, J.J., Naidoo, J., Kengne, A.P., 2015. “Diabetic nephropathy in Africa: a systematic review”. *World J Diabetes*, Vol. 6, p. 759-773.

Pippias, M., Stel, V.S., Abad Diez, J.M., Afentakis, N., HerreroCalvo, J.A., Arias, M., et al., 2015. “Renal replacement therapy in Europe: a summary of the 2012 ERA-EDTA Registry Annual Report”. Clin Kidney J, Vol. 8, p. 248-261.

Paulson, W.D., 2014. “Does Vascular Elasticity Affect Arteriovenous Fistula Maturation?”. *The Open Urology & Nephrology Journal*, Vol. 7, p. 26-32.

Sivanesan, S., How, T.V., Black, R.A., Bakran, A., 1999. “Flow patterns in the radiocephalic arteriovenous fistula: an in vitro study”. *Journal of Biomechanics*, Vol. 32, p. 915-925.

Sorace, A.G., Robbin, M.L., Abts, C., Lockhart, M.E., Allon, M., Hoyt, K., 2011. “Arterial elasticityas a predictor for arteriovenous fistula maturation: preliminary results”. *In Proceeding of the IEEE International Ultrasonics Symposium Proceedings 2011*, Orlando, FL, USA. Paper no. 6293571.

Sorace, A.G., Robbin, M.L., Umphrey, H., Abts, C., Berry, J.L., Lockhart, M.E., Allon, M., Hoyt, K., 2012. “Ultrasound Measurement of Brachial Artery Elasticity Before Hemodialysis Access Placement: A Pilot Study”. *Journal of Ultrassound in Medicine: Official Journal of the American Institute of Ultrasound in Medicine*, Vol. 31, p. 1581-1588.

6. RESPONSIBILITY NOTICE

The authors are the only responsible for the printed material included in this paper.