Bioengineering Research

Biomechanical investigation of a hip simulator during physical activity

Efstratios L. Ntantis [1]*, Dimitra Pithara [1]

[1] College of Engineering and Technology, American University of the Middle East, Kuwait, KW

Correspondence to: Ntantis E. (E-mail: efstratios.ntantis@aum.edu.kw)

Abstract

Biomechanics is a combination of engineering with anatomy and physiology. Studies of biomechanics and ergonomics established to examine the effects of forces on human and animal bodies. Research on biomechanical behavior of the human body during physical activity, help to design special equipment for particular sports and exercises in order to avoid injuries. Spine considered as the most complex structure of the human musculoskeletal system providing support on human movement. This paper aims to explore the ability of a hip simulator to withstand similar forces acting on vertebral body and thus the possibility to replace the spine simulator. The design and stress analysis of the hip simulator performed with CREO Parametric software to show the ability of the design setup to withstand the forces acting on the hip and the loading acting on the vertebral body. For this research, measurements derived from the ORTHOLOAD website and rotation angles estimated in order to import the data to the hip simulator.

Received: 25 April 2021, **Accepted**: 26 May 2021
DOI: 10.22034/jbr.2021.283277.1038
Keywords: Biomechanical, Vertebral body, Spine Simulation, Hip Simulation, Von-Mises Criterion, CREO Parametric

1. Introduction

1.1. The spine

The spine referred as the most complex structure of the human musculoskeletal system. It provides the main support and stability to the body, it allows movement and bending with flexibility, and protects the spinal cord, which is the column of nerves connecting the brain with the rest of the body. Many research studies, most of them using performance tests or models (like finite element models and animal models) conducted to help to understand the biomechanics of healthy and diseased spine, fractures on vertebral body, and to test and optimize the surgical treatments of the spine [1]. Many fractures occur on the thoracic and lumbar region of spine because the position of the vertebral in the spine is low and must bear more weight than others [2]. A way to understand the structure and the function of the spine, the biomechanics divides the spinal column into three different parts; the anterior vertebral body, the anterior longitudinal ligament, and the anterior portion of the annulus fibrosis. When two adjacent columns or the middle column disrupted by congenital disease, trauma, or an operation, then the spine considered as unstable by a general rule [3].

Tissues & Biomaterials Research Group

1.2. Biomechanics

Biomechanics aids to a better understanding of forces and their effects on human and animal bodies. However, many different parameters induced in biomechanics of the spine that make the human spine hard to explain. For example, the biomechanical behavior of the vertebrae most likely affected by different physiological loading conditions, geometric and material changes, stiffness, and range of motion. The clinical studies of human lumbar spine have been significant for the better understanding of the spine biomechanics [4][5]. During designing facilities, tools, and equipment, the ergonomics is involved to identify unsafe conditions and poor body mechanics. Physical activities are important to help to maintain a healthy body. Exercises improve the strength and the mechanical properties of the bones by increasing their density and consequently their Young's modulus [6][7]. A wrong exercise though is possible to cause injuries. Therefore, studies on ergonomics and biomechanics of the human body during physical activities help to design assistive devices for particular sports and workouts. Before designing the special equipment and playing surfaces such as footwear, bandages, protective padding, artificial turf, sports training and aerobic flooring etc., the effects of all dead and dynamic loads have seriously considered during physical activities for protecting human body. In addition, the biomechanics considers the design of equipment for muscular exercises providing resistance. Similar to everyday activities, in sports and exercises, loads on bodies affected by many other grounds. For instance, the apparent weight of a human body is smaller under the water due to the upward buoyant force, therefore there is less resistance [6][7].

1.3. Mechanical properties of the bone

The mechanical properties of a bone vary from point to point because it is an anisotropic material [6]. In a fibrous organic matrix, the bone surrounds the cells that contain the bone tissues. Collagen takes 90 percent of the organic matrix and the rest 10 percent is amorphous ground substance. From these organic matrices, mineral salts are infusing providing the rigidity and strength characteristics [8]. Two bony tissues combined to construct the whole bone, the cortical bone and the trabecular (also known as cancellous) bone. The cortical bone has osteon, the main structural element, same as the trabeculae for the trabecular bone. Cancellous bone is more porous and makes it spongier, which assists to absorb energy. On the other hand, cortical bone is denser and has a higher strength [9]. Stiffness is important to the mechanical properties of a bone, presented in Table 4, because it makes the bone to resist compression and shear stresses. The higher the stiffness the more brittleness the body is, with less toughness [6].

2. Methodology

2.1. Movements data

This paper takes into consideration some physical activities of humans in a typical daily life, such as sitting down, standing up for squats movement, lifting arms, lifting pelvis, walking on a terrain, and walking on a treadmill. The data measurements of the forces and torques developing during the six (6) movements mentioned before, based on the Cartesian coordinate system (*see* Fig.1), found and downloaded from the ORTHOLOAD website. The ORTHOLOAD is a free publicly accessible online database with numerical load data on replacement orthopedic implants. The series of measurements taken from ORTHOLOAD e-library refer to the forces and torques acting on the vertebral bodies of five patients (aged 60-72), all with a lumbar vertebra fracture and an implant surgically added to their spine. The volunteer patients are presented in Figure 2 and their profiles are listed in Table 1 [7][10].

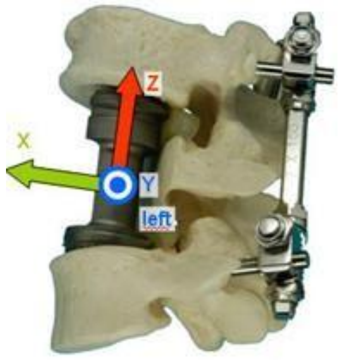

Figure 1. Coordinate system of the acting forces [11]

Figure 2. Patients with an implant surgically added to their spine forces [11]

Table 1: List of patients' basic information

patient	gender	Weight [Kg]	Age of Implantation [Years]	Indication
WP1	M	66	62	Fracture L1
WP2	M	72	71	Fracture L1
WP3	F	64	69	Fracture L1
WP4	M	60	64	Fracture L1
WP5	m	63	67	Fracture L3

Table 2: Formulas for the calculations of the rotation angles

	For Z axis	For Y axis	For X axis
Angles in radians	$\mathrm{Z} = \tan Z^{-1}(\frac{F_x}{F_z})$	$Y = \cos Y^{-1}(\frac{F_y}{F})$	$\mathrm{X} = \cos Y^{-1}(\frac{F_y}{F})$
Convert in degrees	$\mathrm{Z} \times \frac{180^0}{\pi}$	$\Upsilon \times \frac{180^0}{\pi}$	$\mathrm{X} \times \frac{180^0}{\pi}$

Table 3: Values of force during the activity of lifting arms

Fx [N]	*Fy [N]*	*Fz [N]*	*F [N]*	*Mx [Nm]*	*My [Nm]*	*Mz [Nm]*
-45.52	-23.25	-285.7	290.21	0.38	-0.45	-1.17

The data measurements of forces and torques from ORTHOLOAD, inserted as input parameters into the formulas shown in Table 2, to determine the rotation angles in degrees. For example, the values of the forces *Fx* and *Fz* during the movement of an arm lift taken from ORTHOLOAD are 45.52N and 285.98N respectively (*see* Table 3). The conversion into rotation angle Z using the formula taken from Table 2, solved below. The calculated rotation angle Z is 0.158 radians and after the conversion turned into 9.053 degrees. The list of the calculated rotating angles Z in radians for all under consideration movements (i.e. walking on a terrain, walking on a treadmill, lifting arms, lifting pelvis, sitting down, standing up for squat movements) plotted against the Gait Cycle (*see* Fig.3) which is the time period during the movement of activity.

$$Z = \tan Z^{-1}\left(\frac{\mathrm{F_x}}{\mathrm{F_z}}\right) = \tan Z^{-1}\left(\frac{-45.52}{-285.68}\right) = 0.1580108 \text{ radians}$$

$$Z = 0.1580108 \times \frac{180^0}{\pi} = 9.0533527^0$$

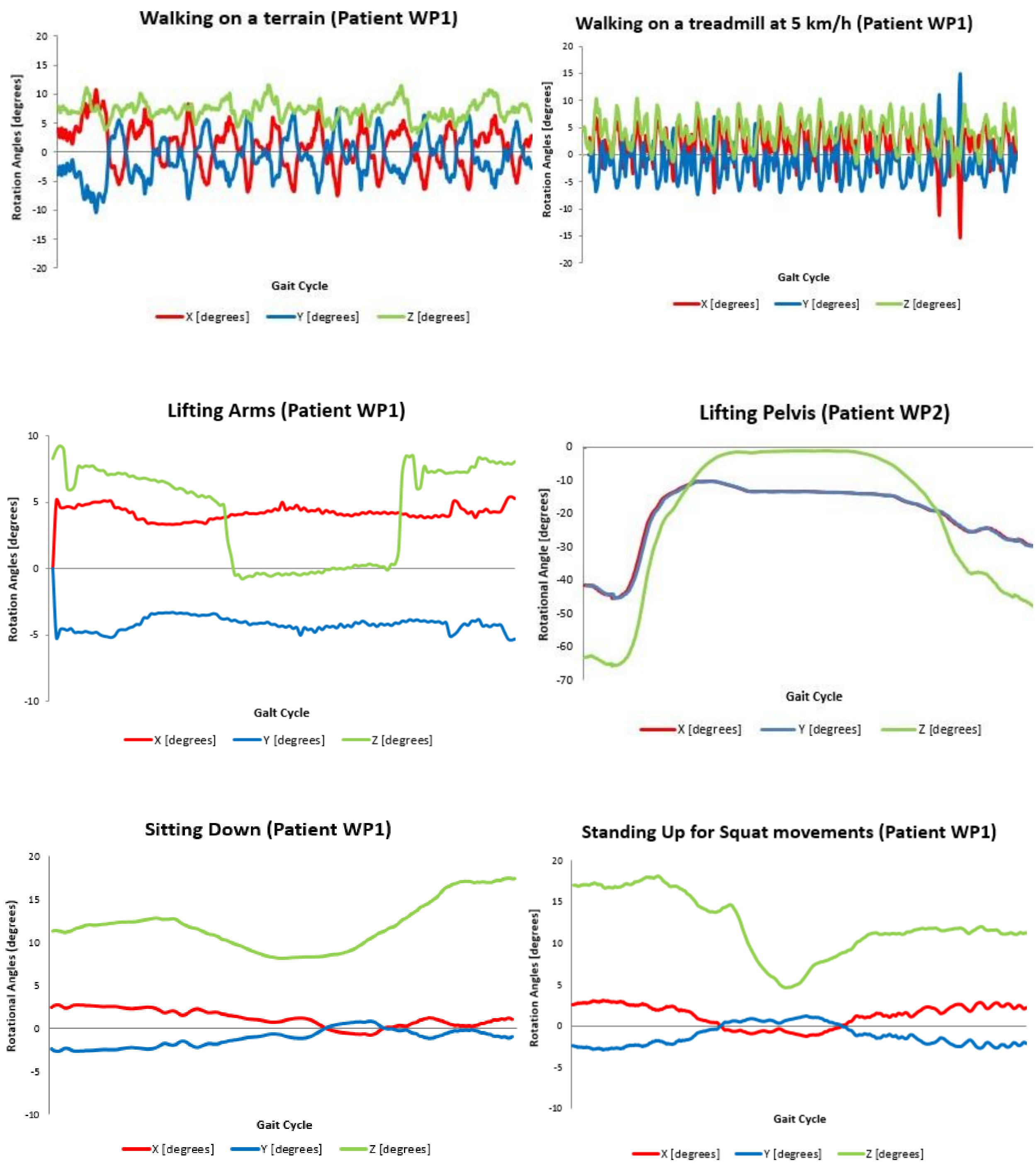

Figure 3. Calculated rotation angles (in degrees) vs Gait cycle (in secs) of the examined activities

2.2. Hip simulator

Due to the very large number of rotating angles Z from the solved equations and graphs, only 127 values kept to ensure the graphs would clearly plotted. These values imported as input data separately for each axis in the hip simulator and the simulation process let to run. The fact that the walk at the treadmill and the terrain are almost the same, their rotation angles at some points were ignored [11]. The same happened to the graphs of the squats exercises, where the sitting down and standing up movements are alike; hence, the rotation angles while the patient is sitting down on the chair also ignored. After the first diagram, new graphs plotted with smoother angles, which then inserted into the hip simulator system for a try. In addition to the first angle, which was initially zero, it changed to a value closer to the last one, so anytime the operation repeats the movement, it will not have a rough reaction [12].

2.3. Setup design

The vertebral body attached to the hip simulator must stay stable during the experiments thus it should be potted in a bone cement (*see* Fig.4) and connected to the hip simulator with some aluminium parts (*see* Fig.5). The aluminium used because it is light but strong material with low cost. The structure designed with CREO Parametrics software followed by structural analysis to confirm the ability of the setup to withstand the loads acting on it (*see* Fig.6). For the test, the materials with proper mechanical properties identified (*see* Table 4), and a downward distributed load from the top inserted, however different ways were also examined (*see* Fig.7). The design set the dimensions of the vertebral body, the hip simulator engine and the Loading Simulator Machine. The Von-Mises yield criterion performed in the structural analysis to estimate numerically the normal and the shear stresses acting on the body [13].

Table 4: Mechanical Properties of aluminium, cement, and bone

		Bone		
	Aluminium	Cortical	Trabecular	Cement
Poisson's Ratio	0.3	0.3	0.03	0.389
Young's Modulus (MPa)	68947.6	13760	151.7	1600

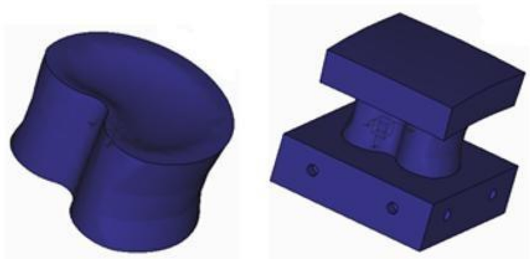

Figure 4. Vertebral body (left) and the potted bone in bone cement (right)

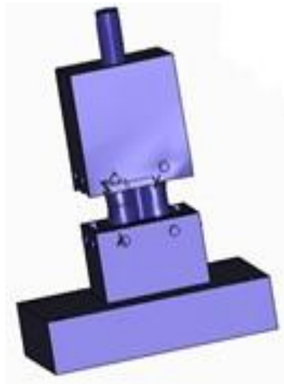

Figure 5. The aluminium assembly holds the vertebral body in the middle

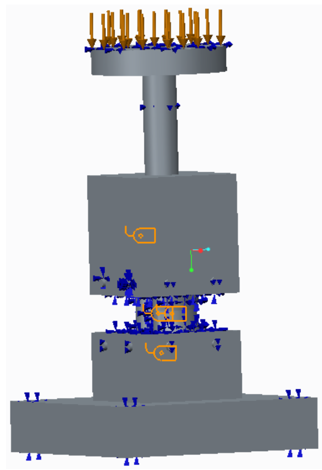

Figure 6. The assembly setup design with the constrains, materials and loading to be identified

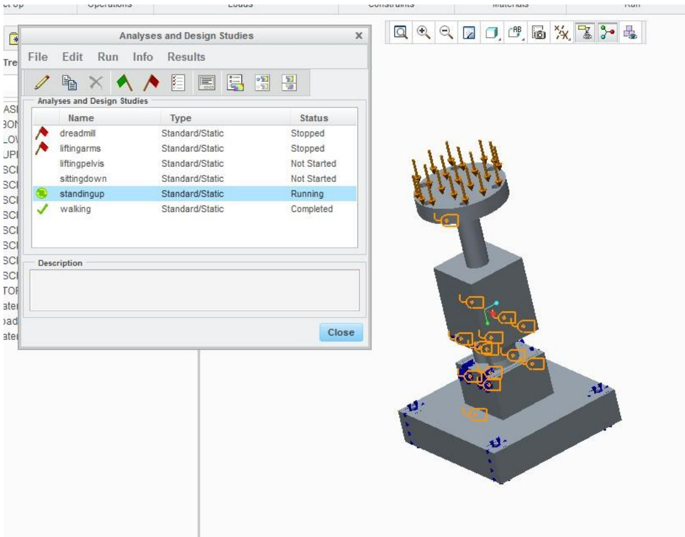

Figure 7. Information of each activity imported into CREO and waiting for the analysis results

3. Results

3.1. Results of hip simulator

During the rotation angles of lifting arms activity operated in the hip simulator in all three axes, the movement in the machine had a very strong vibration, with the movement on X-axis being the strongest one. While the graph got smoother, the X-axis movement did not change but the movement on Y- and Z- axis was a little bit softer on the second trial but afterwards it had no notable difference.

The rotation angles of sitting down activity were very small, because of the low intensity of the movement. Once the changes completed on Y-axis, the movement improved with small differences each time. Until the last trial, the Z-axis showed a significant change with a smoother movement. On the other hand, the movement on X-axis stayed strong and vibrant.

Similarly, during the standing up activity on the first trial, the X- Y- and Z- axis had a strong shuddering movement but after the graphs got smoother, the movement on Y- and Z- axis became less trembling. However, the movement on X-axis remained the same.

For the physical activity of walking, the hip simulator operated exceptionally fast with large movement. There was no difference shown in the experiments on walking on a treadmill followed.

The rotation angles on X-axis for the movement of walking on a treadmill (with a speed of 5 km/h) were too small, almost same with the zero line. The rotation angles on Z- and Y- axis were larger. Although the machine had a strong vibration during movement, it did not change after the modifications on the graphs.

The lifting pelvis movement in contrary with most of other activities, performed with large rotation angles except the angles on Z-axis which crossed the limit and so they were not accepted. On X-axis, even if the angles were in the limit of acceptance in the hip simulator the movement cut off during the experiment. Only the movement on Y-axis was small and strong. This activity was not tested again because the changes of the angles needed to be done were immense and most likely they could affect the demonstration of the movement.

As already observed, the phenomenon plays an important role in the wear of the metal prosthesis and it is the object of study in several research investigations. However, conventional simulators do not meet the needs of these research teams. The main reason is that in order to analyze the CREO process in pair of metals, a more demanding and specialized modeling software is required, where for instance a pair of prosthetic implants could undertake a dynamic and a kinematic analysis of the joint of a hip. In order to observe in details, the relationship of the CREO process with the applied load or with the kinematics adopted, there is a way to vary these characteristics. Under this context, the realization of the present research paper is justified.

3.2. Results of CREO Analysis

The CREO Simulation Analysis has showed that the setup design for the vertebral body can withstand the forces developed during the experiments in the hip and the load simulator. However, the results of the forces acting on the assembly design for the hip simulator did not show the impact of the forces all the way until the vertebral body. For that reason, a stress analysis completed at different parts of the assembly, in order to demonstrate a more detailed image and better understanding of the distribution of the applied load on the vertebral body. Images of the analysis illustrated below, selected randomly.

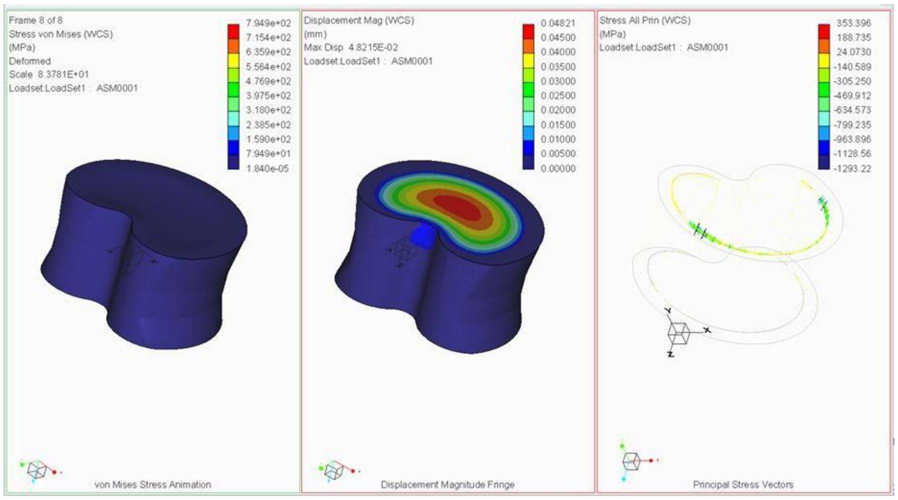

Figure 8. Analysis of bone during lifting arms

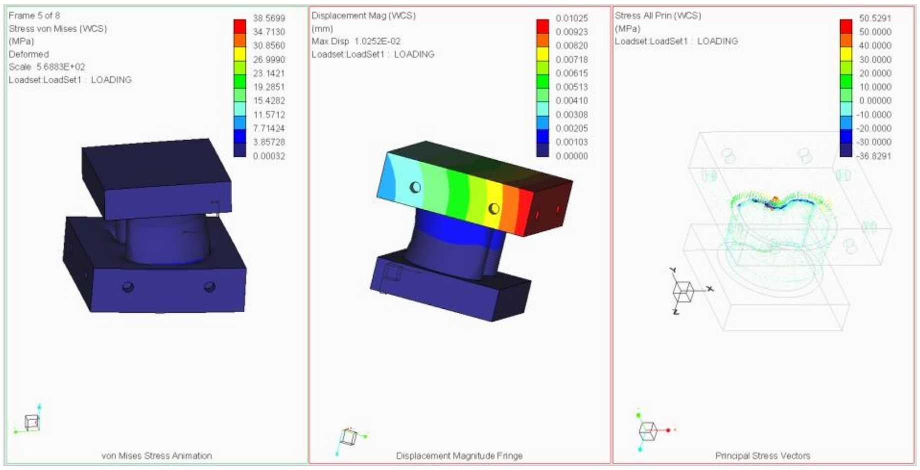

Figure 9. Analysis of the potted bone and the aluminium case during lifting arms

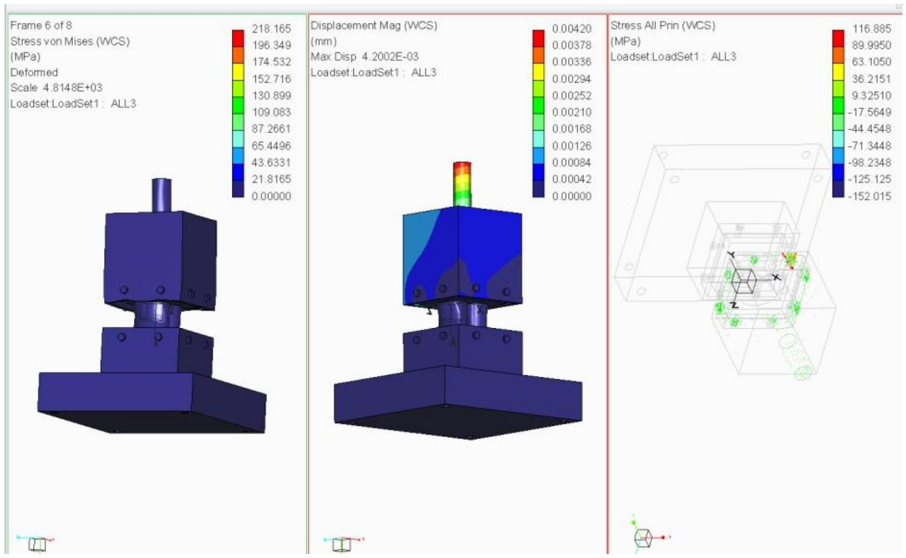

Figure 10. Analysis of the assembly during lifting arms

4. Discussion

The analysis shows that the setup for the hip simulator properly designed, although more tests should be conducted to show even better and more precisely the forces acting on the bone under different occasions. From the results of the Von- Mises stress analysis and the provided yield strength characteristics of the bone, was observed that no fractures occurred during the simulation of the selected activities. The results of the experiments indicated that the hip simulator could not replace the spine simulator for all the activities because it did not give a clear movement of the spine throughout a particular activity. A further improvement of the graph could make possible the replacement of the spine simulator by the hip simulator, but only if the rotation angles would be considered. Therefore, the rotation angles cannot used for any movements beyond those experimented. Regarding the simulation outcome, it could better validated with more detailed analysis, with the use of another simulation program instead of CREO software, which would be able to specify further the forces acting on the bone with or/and without the setup for the hip simulator.

5. Conclusion

The biomechanical investigation of a hip simulator during physical activity achieved in this paper. Studies of biomechanics and ergonomics established to examine the forces developed on the vertebral body. A design setup for the vertebral body created in CREO Parametric 3D Modeling software to investigate the ability to withstand the forces acting during the experiments in the hip simulator. The data of measurements for different physical activities of humans in a typical daily life received from ORTHOLOAD website. The stress analysis in CREO Parametric software and the experimental data from the hip simulator, demonstrated a proper use of the setup design for the hip simulator, although more tests shall conducted for higher precision and detailed forces acting on the bone under different occasions.

Ethical Approval

All procedures followed were in accordance with the ethical standards with the responsible committee on human experimentation and with the Helsinki declaration of 1975 (in its most recently. Informed consent obtained from all individual participants included in the study.

Conflict of Interest

The authors declare that they have no conflict of interest.

References

[1] Cristofolini L., Brandolini N., Danesi V., Mateusz M., Juszczyk P., Viceconti E. and Viceconti M. “Strain Distribution in the lumbar vertebrae under different loading configurations”. The Spine Journal, Vol.13 (10), 2013, pp.1281-1292.

[2] Eben D. “Lumbar Spine Anatomy and Pain”, (online), available from Spine-health digital libraries: <https://www.spine-health.com/conditions/spine-anatomy/lumbar-spine-anatomy-and-pain>, (accessed on 7 April 2021).

[3] Slone R.M., MacMilan M. and Mintgomery W.J. “Spinal fixation. Part 1. Principles, basic hardware and fixation techniques for the cervical spine”. Radiographics, Vol.13 (2), 1993, pp.341-356.

[4] Sinclair J., Brooks D. and Stainton P. “Biomechanical effects of a lightweight, sock-style minimalist footwear design during running: a musculoskeletal simulation and statistical parametric mapping approach”, Footwear Science, Vol.11 (2), 2019, pp.71-83.

[5] Park W.M., Wang S., Kim Y.H., Wood K.B., Sim J.A. and Li G. “Effect of the intra-abdominal pressure and the center of segmental body mass on the lumbar spine mechanics - a computational parametric study”, Journal of Biomechanical Engineering, Vol. 134(1), 2012, pp. 229-236.

[6] Williams M. and Lissner H.R. (1977). Biomechanics of human motion. Philadelphia. Saunders.

[7] Ashtiani M.N., Azghani M. and Shakouri S.K. “Biomechanical modeling of hip abduction exercise using optimized inverse dynamics”. Zahedan Journal of Research in Medical Sciences, Vol. 21(1) 2018, pp.9-16.

[8] Mow V.C. and Huiskes R. (2004). Basic orthopaedic biomechanics and mechano-biology. Lippincott Williams & Wilkins.

[9] Bartlett R. and Bussey M. (2011). Reducing injury risk and improving sports performance. Sports Biomechanics. Routledge.

[10] Farmakis I.K., Potsika V.T., Pakos E. and Fotiadis D.I. “Biomechanical evaluation of the impact of different weight loading conditions on the mechanical environment of the hip joint endoprosthesis”. World Congress on Medical Physics and Biomedical Engineering, 2018, pp.691-696, Singapore.

[11] Washabaugh E.P., Augenstein T.E. and Krishnan C. “Functional resistance training during walking: mode of application differentially affects gait biomechanics and muscle activation patterns”. Gait & Posture, Vol.75 (1), 2020. pp. 129-136.

[12] Sinclair J., Ingram J., Butters B., Brooks D., Stainton P. and Taylor P.J. “A three-experiment examination of iliotibial band strain characteristics during different conditions using musculoskeletal simulation”. Sport Sciences for Health, Vol.16 (4), 2020, pp.727-736.

[13] Cardoso M., McKinnon C., Viggiani D., Johnson M.J., Callaghan J.P. and Albert W.J. “Biomechanical investigation of prolonged driving in an ergonomically designed truck seat prototype”. Ergonomics, Vol.61 (3), 2018, pp.367-380.