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Research Paper

Strength and Stability of a Motorized Unilateral External Fixation Device for Fractures and Bone Loss under Compressive Load

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Abstract
Aligning broken bones using a unilateral external bone fixator has become popular today. One of the advantages of unilateral external bone fixators is their ease of installation and adjustment compared to circular and horseshoe models. Unlike circular and horseshoe-shaped models, the advantages of unilateral external bone fixators are ease of installation and adjustability. The present research aims to ensure the stability and strength of the unilateral motorized external bone fixation device, equipped with four motors designed and then built in SolidWorks software under a compressive load of 150N when the motors are in motion. The device was simulated and analyzed using the Finite Element Method (FEM) in ANSYS software. Then, the constructed device was tested in the laboratory by applying a compressive force of 150N. The results of the simulation and experimental method of compressive force application and examination of the reliability coefficient using the FEM in the simulated method indicated that: The device is designed with the necessary stability, rigidity, and reliability to stabilize fractured long bones with the ability to move broken long bones where part of the bone has been lost. The device has precise bone movement and four independent motorized units; based on the patient's needs and the specialist's instructions, bone formation is possible.

Keywords
Unilateral External Bone Fixator, Bone Fracture, Bone Loss, Orthopedics, FEM

1. Introduction

Unilateral external bone fixators are extensively utilized in orthopedics to fix broken bones and, in some cases, to displace broken bones that contain a lost part to make up the lost bone. Therefore, a motorized unilateral external bone fixator with the necessary strength and stability is suitable for treating bone loss and fractures.

In this regard, the device must be adjusted carefully to achieve the desired result because a mistake in it may lead to many injuries, including failure in treatment and the need for re-surgery and re-installation of the device, which in turn may cause various problems, including increased costs,

prolonged treatment time, infection, or loss of bone repair and recovery. By examining the history of evolution and external fixators, we can understand the importance of using unilateral external bone fixators in orthopedics. Here, the process of using and developing external fixators is discussed, and the investigations carried out on the degree of strength and stability and the effect of using these fixators in treating bone fractures are briefly reviewed.

The use of external fixators has primarily caused limb reconstruction techniques to advance. Fragomen et al. [1] stated that the external fixator, currently the latest resort for fixation and becoming the primary method of treating various bone and soft tissue pathologies, needs to be developed and optimized. According to the research by Paul et al. [2] and Hernigou [3], although external bone fixation is often considered a "novel" procedure in traumatology and orthopedics, it has been used by physicians and surgeons for centuries. Around 377 BC, Hippocrates made a natural early external fixator. The first models of these devices included wooden splints to treat fractures. The new and familiar concept of an external bone fixator was introduced by Jean-François Malgin, who made a device called a "metal point" in 1840. Then, in 1843, he introduced a two-fork machine called the "metal claw." Paul et al. and Hernigou [2, 3] stated that Dr. Clayton Parkhill was among the first to develop a true unilateral external bone fixator in 1897. According to Paul et al. [2], Lambert designed the monocortical fixator in 1911. The compressive mechanism of modern external fixators is derived from Lambert's design and ingenuity. In their study, Fleming et al. [4] stated that the Ilizarov external fixator is used to fix bone fractures. A study performed by Goodship and Kenwright showed that induced axial micromotion at the fracture site could accelerate the healing process. Broekhuizen [5] stated that the Wagner device (Mathys, Bettlach, Switzerland), which is made of stainless steel, was initially designed to lengthen long bones, especially the femur. Qiao et al. [6] developed robotic design and orientation techniques that can improve the device's accuracy. Wei et al. [7] provided digital measurement methods using Paley's deformity measurement method; they also proposed a deformity correction algorithm to calculate the elongations of the six rods. Corona et al. [8] worked on circular frames to improve the preoperative planning process for post-traumatic tibial deformities. Based on their research, Zhao et al. [9] stated that by combining the advantages of series and parallel mechanisms, the production of hybrid robots can meet specific clinical needs, including joint fractures and large multi-component fractures. Matsushita et al. [10] considered the hifixator device an external fixator that utilizes a new sliding mechanism. This mechanism maintained its capability in 72% of the motions performed at the fracture site, even when the pins were loose, and the torque was 4 Nm. In a study, Sangkaew [11] stated that the technique has modified the distraction osteogenesis with the help of the available external fixator of AO/ASIF, a safe, cost-effective, and versatile tool. In the research performed by Hussain et al. [12] on increasing the lower limb length by unilateral external fixator in the field of limb-length discrepancy using the Wagner method, it was concluded that according to the criteria of application, in 89% of bony results and 97% of functional outcomes, an excellent or good degree of success was achieved. Tang et al. [13] concluded that single-stage arthrodesis of the knee using a unilateral external fixator with cannulated screws is an effective way the treatment of end-stage tuberculosis of the knee. In the study of Basso et al. [14], it was concluded that 95.8% of patients were satisfied with using unilateral external fixators in treating humeral shaft fractures. A study by Yushan et al. [15] showed that treating significant tibial defects due to infection in trifocal bone transport using a unilateral rail system significantly improved postoperative function

and reduced the duration of regenerate consolidation and docking union. Sen et al. [16] examined a combined method for treating distal femoral bone defects following the removal of bone infection using an external fixator with a short supracondylar nail. They concluded that the combined strategy was influential in treating distal femoral segmental bone defects after debridement of osteomyelitis, which had a high union rate and acceptable complication rates. Strebe et al. [17] evaluated three strategies of double stacking, crosslinking, and diagonal pin from ultra-high molecular weight polyethylene bone models and available external fixator parts. The results showed that double stacking was the most efficient way to increase resistance against bending, especially anterior-posterior bending and axial pressure; however, it significantly increased the cost. Ang et al. [18] evaluated the axial and torsional stiffness of the externalized titanium locking compression plate (ET-LCP), the externalized stainless steel locking compression plate (ESS-LCP), and the unilateral external fixator (UEF) and introduced the use of LCP as an external fixator a proper alternative to the old UEF because its smaller structure is more acceptable for patients and puts no pressure on axial and torsional stiffness. Li et al. [19] used the FEM to examine the stress and deformation of the external bone fixator system under axial, torsional, and bending load and to compare the biomechanical properties of the two fixators. One of the two fixators had a pin deviation angle, and the other did not. When the pin deviation angle changed 0-20 degrees, the growth rate of stress or deformity was prolonged, but when it exceeded 20 degrees, the slope of the growth rate increased much more; in other words, the effect of pin deviation on the stability of external bone fixator system increased. In their study, Zainudin et al. [20] claimed that if parameters such as the biomechanical perspective are considered an external fixator, the bone will be successfully healed; pin diameter is one of these parameters. The FEM was used to simulate the standing phase. The results demonstrated that selecting a pin with a diameter of 6.5 mm leads to the least Von Mises stress on the joint surface of the pin and bone. Shi et al. [21] reported that the plate-type external fixator has higher stiffness and strength than the unilateral external fixator. The highest biomechanics belonged primarily to the classical plate-type external fixator, followed by the extended plate-type external fixator with a slight difference. The plate-type external fixator has higher stiffness and strength than the unilateral external fixator under axial compression, four-point bending, and torsion. Jean et al. [22] used the Hoffmann®3 device as a reference for comparison in the study. Six external fixators were examined in three modes: axial compression, mediolateral (ML) bending, and torsion to determine the structural strength. The results showed that the stiffness of UUEF (unilateral uni planar external fixator) and UBEF (unilateral biplanar external fixator) devices compared with the reference fixator may be helpful in fracture healing and protection. Lesniewska et al. [23] performed FEM on fracture healing using a fixation device. Relevant analyses were performed under axial and variable loaded boundary conditions. The results demonstrated that at the beginning of the fracture healing process, the stresses in the external fixator device are the highest and gradually decrease over time. A study by Donaldson et al. [24] concluded that local bone yielding at the pin-bone interface in the external fixation method using half-pin causes the fixator to loosen. The peri-implant yielded threefold bone volume increases from young to old patients. If three half-pins are used instead of two half-pins on each side of the fracture, the yielded bone volume will be reduced by 80% in all age groups. Using titanium half-pins minimizes the importance of yielded bone by about 60-65%. Roseiro et al. [25] developed a FEM simplified for external fixation of the tibia bone to determine the stiffness at the fracture center. The genetic

algorithm was also defined to minimize the displacement of the fracture center (objective function) by changing the position of the external fixator's mechanical parts and evaluating the load imposition types. Wang et al. [26] stated that based on the calculated results, if a solid screw is used, there is a lot of stress at the beginning of the fracture healing process, both on the screws and the femur. Still, when using a hollow screw, when an open screw is used, the stress is distributed more evenly, and in the middle of the healing process, the stress on the femur is significantly reduced. Li et al. [27] reported that stiffness is the main criterion used to evaluate the mechanical stability of external fixators. The external fixator's stiffness affects the fractured bone's local biomechanical environment. They developed a theoretical model by modifying Young's modulus of the callus using Castigliano's theory to evaluate the compression stiffness, torsional stiffness, and bending stiffness of the fixator-bone system during the healing process. The results showed the similarity of the three methods of stiffness assessment in the fixator-bone system. FEM shows that as the healing time lengthens, the transmission of the load between the fixator and bone changes. Moreover, FEM confirms the results of the theoretical analysis. Salunkhe et al. [28] designed a high-power external fixator, which weighed only 1.217kg and had a suitable mechanism for the dynamic treatment of unstable fractures. Maximum displacement was determined between fractured bone fragments. The maximum removal from applying a compressive force 2000N was only 0.0018 mm, within the acceptable range.

Stiffness of the external fixation system at axial pressure load and mechanical stability for the external fixation device in the case of Anterior-Posterior bending were analyzed, and results for displacements were obtained for selected critical places on the device and the place of fracture. Considering all data, it can be concluded that the external fixation device Orthofix has good mechanical stability for the AP bending load. Also, there is a possibility of improving the device using new advanced materials or redesigning it [29, 30]. A unilateral external fixator as a primary and definitive treatment is a viable, simple, and effective option for TDF with a high success rate, even in a resource-limited setting [31]. The unilateral fixation may provide desirable results in smaller fracture gaps, but its usage in more extensive gap fractures might be alarming [32]. The most common problem in the clinical application of external fixation is the failure at the pin-bone interface, which manifests as pin loosening that may lead to pin tract infection or loss of fracture reduction, which may be diminished by MDP (Micro-Motion Damping Pin) [33].

According to studies, designing a device to decrease errors is an efficient step in bone fracture treatments. Moreover, being motorized helps the device to function during the bone loss treatment and gives the patient a sense of comfort, which is very important. For this purpose, a motorized external unilateral fixator device was designed to be used in the treatment of fractures and bone loss. This device can effectively fix the bone in four areas. Most importantly, it can use the motors installed in each part (four separate units) to make the necessary displacements of parts of the bones to build bone and compensate for bone loss.

The purpose of this research is to simulate using FEM and experimental method to investigate the strength and stability of a motorized unilateral external bone fixation device; it is equipped with four motors capable of moving vertically (up and down) and is designed separately, which operates based on the patient's needs and the diagnosis of a specialist physician. The device was developed in SolidWorks software and then analyzed under compressive load using FEM; in ANSYS software, a suitable and reliable design was sought for use in medical orthopedic centers, and then, by

manufacturing and assembling the device parts, the condition of the pressure load bed device was examined in a standard laboratory environment. Therefore, with the confirmation of the stability, strength, and rigidity of the motorized unilateral external bone fixation device under compressive load, it is possible to mass produce and use it to solve the problems of patients with long bone fractures, Patients with limb defect or short leg, short patients who want to increase height, Patients who are missing part of a bone and a series of such diseases, which are considered bone diseases (orthopedics), are cured.

2. Material

The unilateral external bone fixator equipped with four motors has medical (orthopedic) use, and considering the need for long-term use of the device by the patient during treatment, it is necessary to choose a medically approved material with high thermal resistance, strength, corrosion resistance, and abrasion resistance. Therefore, we searched for medical devices and equipment materials, and stainless steel 316 was the most commonly used material. Stainless steel 316 has characteristics such as high machinability, ductility, weldability, and thermal resistance, and at the same time, it is non-magnetic. Therefore it was selected as the primary material, and the device stability and strength were analyzed by the FEM, considering stainless steel as the primary material. The chemical composition and mechanical and physical properties of stainless steel 316 were extracted from standard sources and shown in Tables 1, 2, and 3, respectively.

Table 1. The percentage chemical composition of stainless steel 316

Grade		C	Mn	Si	P	S	Cr	Mo	Ni	N
316	Min	-	-	-	-	-	16.0	2.00	10.0	-
	Max	0.08	2.0	0.75	0.045	0.030	18.0	3.00	14.0	0.10
316L	Min	-	-	-	-	-	16.0	2.00	10.0	-
	Max	0.03	2.0	0.75	0.045	0.030	18.0	3.00	14.0	0.10
316H	Min	0.04	-	-	-	-	16.0	2.00	10.0	-
	Max	0.10	2.0	0.75	0.045	0.030	18.0	3.00	14.0	-

Table 2. Mechanical properties of stainless steel 316

Grade	Tensile Strength (MPa) Min	Yield Strength 0.2% Proof (MPa) Min	Elongation (% in 50mm) Min	Hardness	
				Rockwell B (HR B) Max	Brinell (HB) Max
316	515	205	40	95	217
316L	485	170	40	95	217
316H	515	205	40	95	217

Table 3. Physical properties of stainless steel 316 under annealed conditions

Grade	Density (Kg/m^2)	Elastic Modulus (GPa)	Mean Coefficient of Thermal Expansion			Thermal Conductivity		Specific Heat	Electrical Resistivity (nΩ.m)
			0-100°C (µm/m/°C)	0-130°C (µm/m/°C)	0-538°C (µm/m/°C)	at 100°C (W/m.k)	at 500°C (W/m.k)	0-100°C (J/Kg.K)	
316 & 316L/H	8000	193	15.9	16.2	17.5	16.3	21.5	500	740

3. Method

3.1 For designing the device and analysis of the device

The unilateral external bone fixator was designed using Solid Works. The design was transferred to ANSYS software for simulation and analysis of the device using the FEM, The necessary simulations were performed for loading and applying compressive force, and Then, the device was made for testing, along with the clamps and the tube (the tube was used instead of the bone) with similar specifications to the tibia bone, which were fixed next to each other by the clamps, was subjected to compressive load according to actual conditions in a standard laboratory environment based on the designed test, Then, the results of the software simulation were compared with the results of the experimental method, examined and analyzed. In the following, the software simulation and the experimental process, and the necessary conditions are considered, and the boundary conditions used in the FEM and the experimental method are presented.

The device's Schanz holder is designed in a standard way to allow the use of standard threaded Schanz (threaded pins) commonly used in orthopedics.

The fixing device consists of four threaded pin-holding units (Schanz holder); each unit can move independently on the central axis through the motor. First, a 3D model of the bone fixation device was designed using SolidWorks software. Also, the CT scan image of a bone fractured into four pieces, part of which was lost, was prepared and converted using the Mimics software into a 3D model for SolidWorks software. Using the schanzes installed in the Schanz holder of the device, the segmented parts of the fractured bone were fixed next to each other. Fig. 1 shows the 3D view of the device with schanzes and the bone. (The process of attaching the device to the bone by schanzes and placement of the fractured bone segments was carried out according to the studies, some of which have been presented in the last part of this study). Table 4 shows the mechanical properties of stainless steel 316 and bone to achieve the specific force application process needed to perform the device simulations. The schanzes used in this study were 5×200 mm standard solid threaded schanzes (threaded pins)—the 3D file of SolidWorks with step. The extension was inserted into the ANSYS Workbench and meshed using the available elements (Figure 2). After studying the mesh convergence obtained by increasing the density of elements in sensitive places of the system, the number of components and nodes were set at 364770 and 639921, respectively, for simulation and analysis by the FEM.

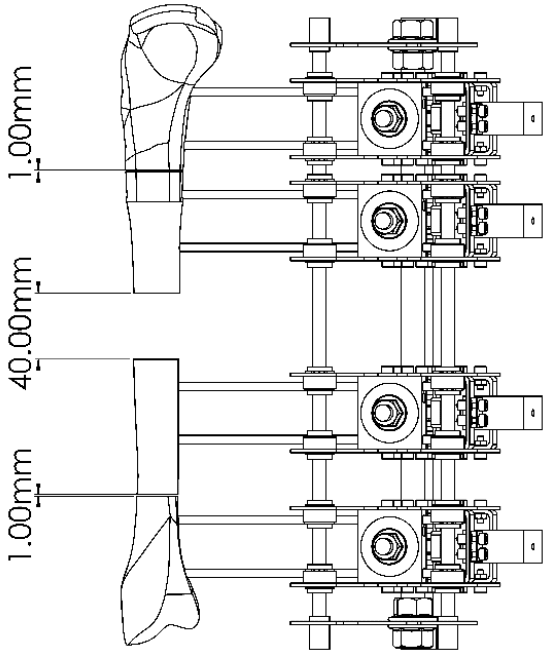

Figure 1. The motorized unilateral bone fixator was designed in SolidWorks software along with schanzes and how to attach threaded schanzes to the fractured bone containing lost parts

Table 4. Mechanical properties of stainless steel 316 used in the device [19, 25]

Materials	Young's modulus (GPa)	Poisson's ratio	Yield strength (MPa)
Stainless steel	193	0.31	205
Bone	17	0.3	300

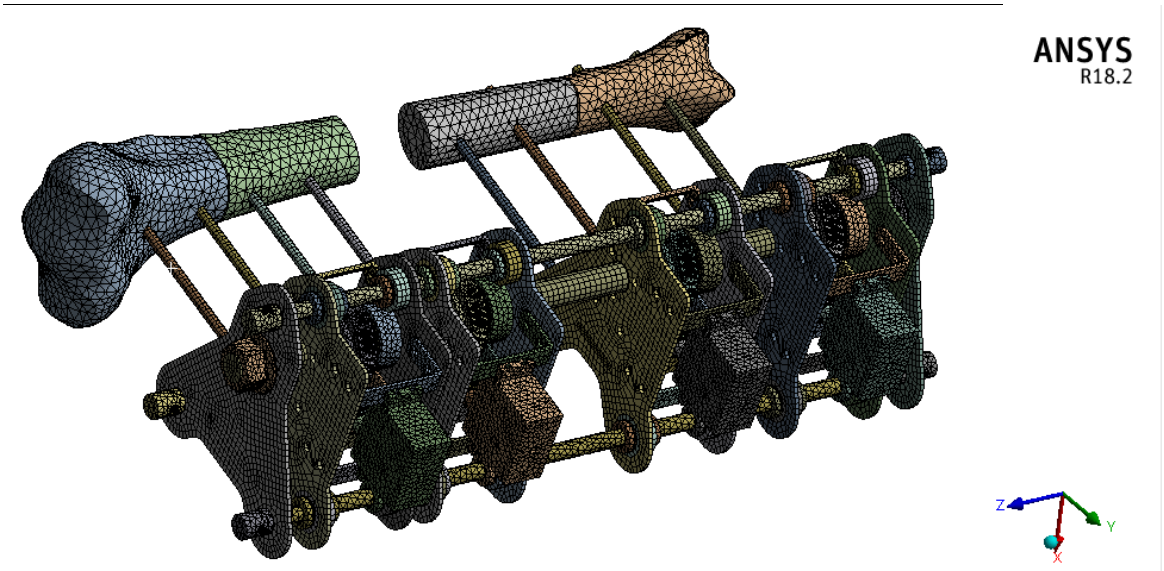

Figure 2. Final meshing for modeling a motorized unilateral bone fixator

3.2 Conditions of simulation and analysis using FEM

In this model, the bone density, the mean Young modulus, and Poisson's ratio were considered 1800 kg/m^3, 18 GPa, and 0.2, respectively. The contact surface of the schanzes with the bone was assumed to be fixed, explaining that none of the surfaces had a degree of freedom relative to each other. The

material of different device parts was stainless steel 316, which has a Young modulus and Poisson's ratio of 193 GPa and 0.3, respectively. Except for moving contact surfaces between the parts that have a degree of freedom from each other, the contact between different parts was bonded or fixed. In this simulation, it was assumed that the standard threaded pins (schanzes) inside the bone have no movement at all, the Schanz holders move on the central axis, and the guide rods move in the form of a cylindrical joint to be able to move along their axis if needed.

3.3 Conditions considered for experimental method

To be able to perform tests similar to actual conditions, a polypropylene pipe with an external diameter of 40 mm, inner diameter of 26.6 mm, Thickness of 6.7 mm, and a length of 400 mm was used instead of the tibia bone (the specifications of the tube used are similar to the tibia bone) which was cut into four pieces. The pipe pieces were fixed by the clamp using standard solid threaded clamps (threaded pins) made of stainless steel with a diameter of 5 mm and a length of 200 mm, then the device, along with the schanzes and pipe, was fixed in the pressure test device according to the test design. Then, by activating the motors, the movement of bone fragments began. At the same time, a compressive force equal to 150N (of course, during the experiment, a force greater than 150N was unintentionally applied) was applied in standard laboratory conditions (temperature 25 degrees Celsius and humidity 25%).

4. Results and discussion

4.1 In the first stage

To examine the device stability using the FEM, the force applied was simulated in the ANSYS software so that the force was applied in the axial direction of the bone while the other end of the bone was fixed. The boundary conditions considered for this simulation were that one end of the bone was assumed to be fixed. A force equivalent to 150N was applied (F) to the other end (Figure 3) [10, 17, 19, 21]. Simultaneously with the application of force, the two middle motors of the device moved 0.25 mm in 1 second, as shown in Figure 3. After simulation and force application, three displacement and deformation contours and Von Mises stress and safety factors were obtained as data for device analysis.

The maximum displacement in the bone and its schanzes (Figure 4) was about 1.3414 mm in the direction of applying force and the bone axis. Also, the maximum Von Mises stress was 246.43 MPa (Figure 5). The safety factor of the system was 15 (Figure 6), indicating that the maximum force allowed to apply to the system is approximately 150N. Still, within the range of schanzes, the confidence coefficient shows about 0.79227, indicating that the maximum force allowed to apply to the system is approximately 150N, and using more force will cause the schanzes to fail.

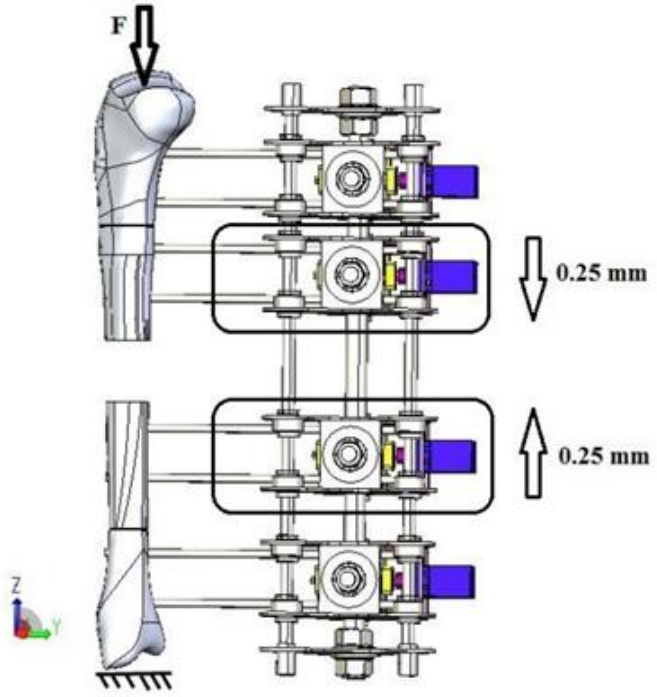

Figure 3. Boundary conditions for the simulation of compressive force in dynamic mode

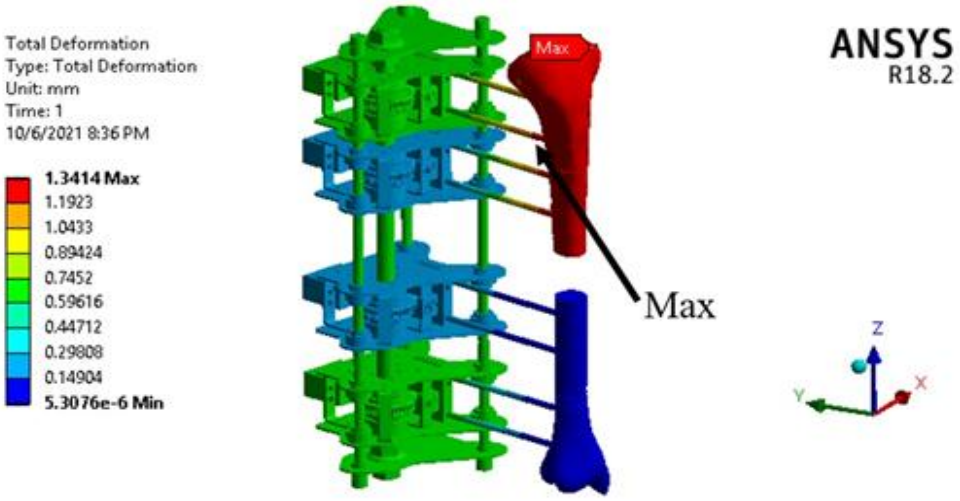

Figure 4. Displacement and deformation distribution contour of 150N compression force

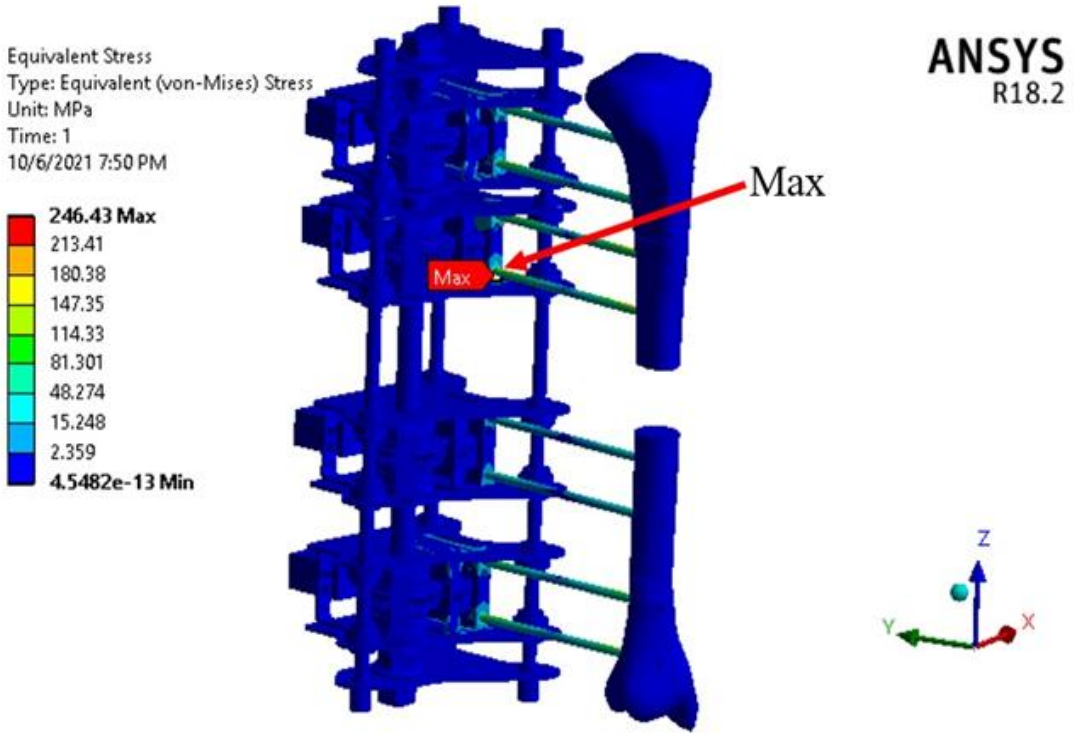

Figure 5. Von-Mises stress distribution contour applying a compressive force of 150N

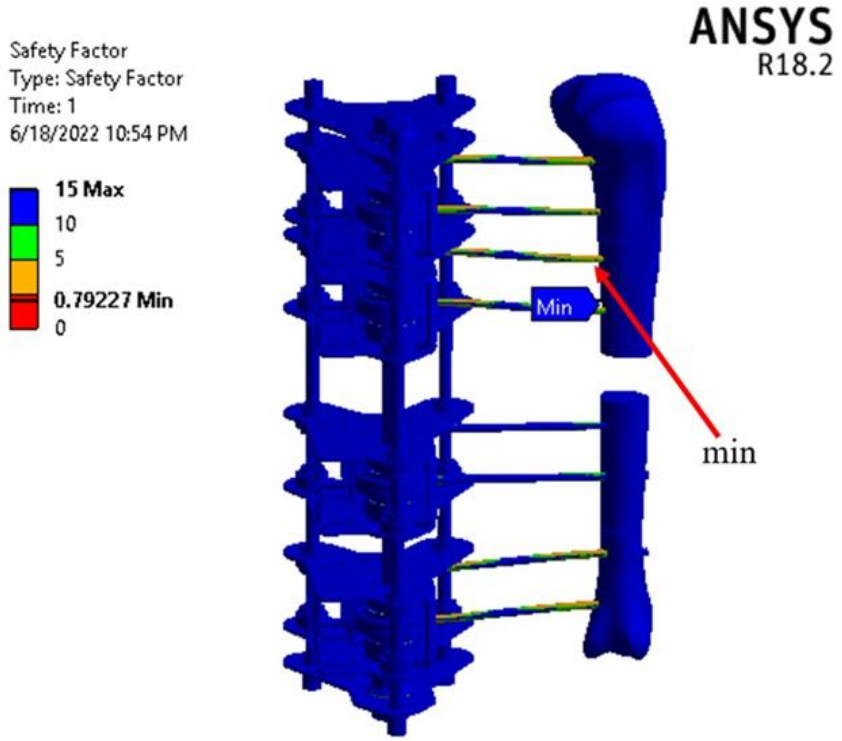

Figure 6. Distribution of the safety factor for applying a compressive force of 150N

In Figure 7, a force equal to 130 N has been applied [10, 17, 19, 21], and at the same time as the two middle motors of the device are used, they have moved by 0.25 mm in 1 second, as shown in Figure 3. After simulation and force application, three displacement and deformation contours and Von Mises stress and safety factors were obtained. The data required for device analysis was obtained.
The maximum displacement in the bone and its supporting schanzes (Figure 7) is approximately 1.1842 mm. This displacement is in the direction of force application and along the axis of the bone. Also, the maximum Von Mises stress (Figure 8) equals 207.3 MPa. Figure 9 shows the system safety factor. Which has a value equal to 15, and this means that the system will withstand a load of 130N and will not fail; but within the limits of the schanzes, the safety factor shows about 1.228; this means that by applying this amount of force, the schanzes will change shape. The location of the minimum safety factor across all system components indicates the application of a force of 130N along the axis of the fractured bone.

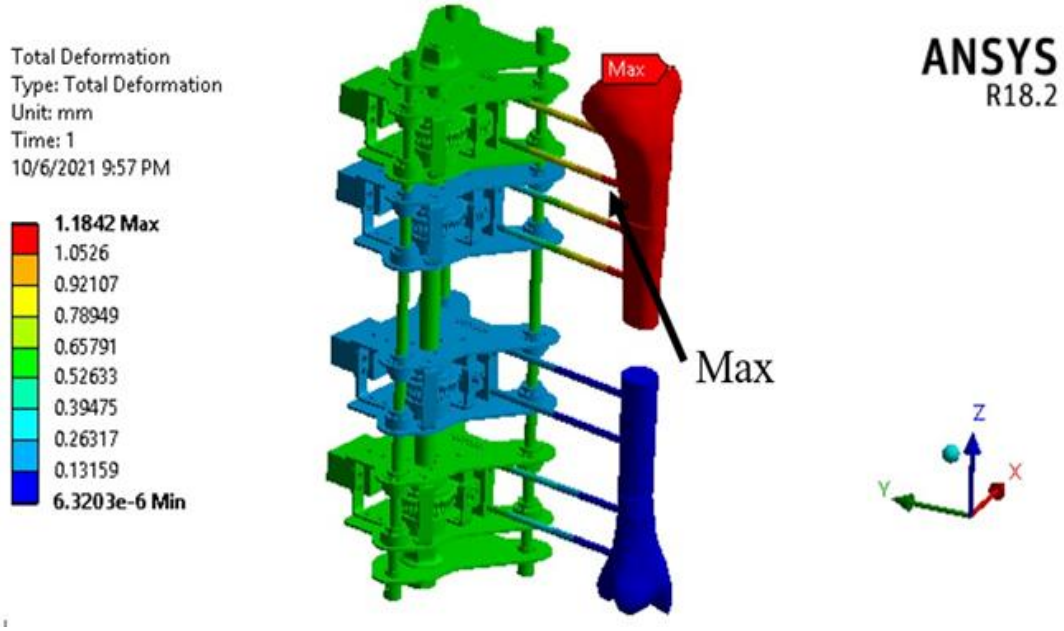

Figure 7. Displacement and deformation distribution contour of 130N compression force

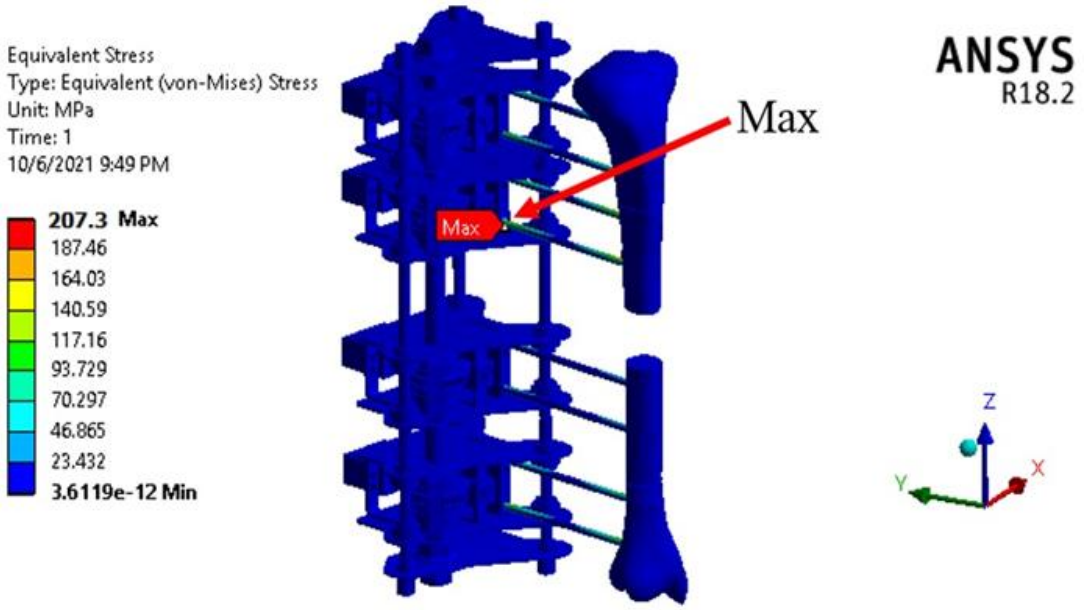

Figure 8. Von-Mises stress distribution contour applying a compressive force of 130N

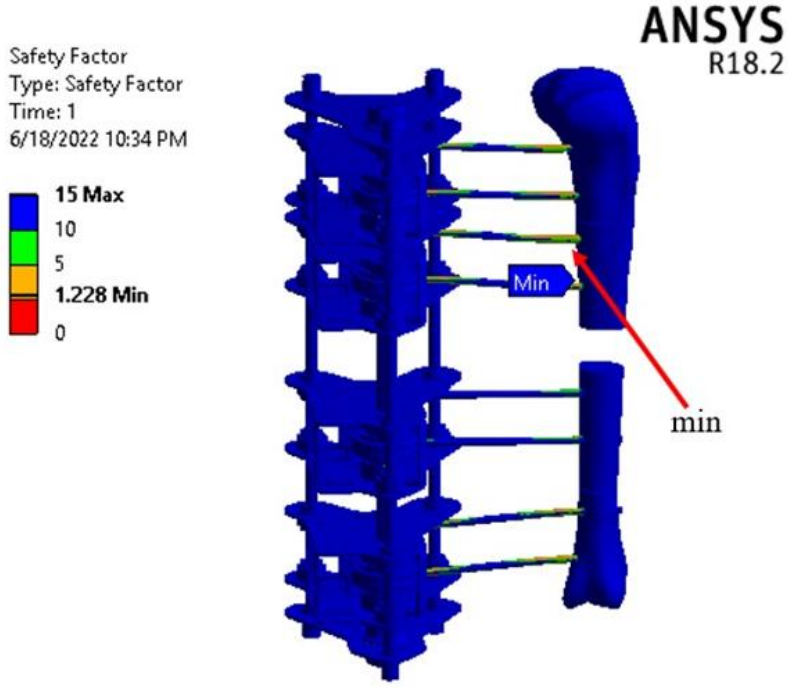

Figure 9. Distribution of the safety factor for applying a compressive force of 130N

According to Von Mises, stress obtained from the simulations performed and considering the yield stress of 205 MPa of stainless steel 316, the maximum allowable applied axial force was found to be 150N, and the safety factor was one. This condition means that any increase in force will change the shape of the chances and cause them to fail.

The data obtained from the deformation and Von Mises stress for the application of two forces are given in Table 5; a force of 150N is the force that leads to failure in the Shanzes piece.

Table 5. Data obtained from the application of force in the simulation of compressive force

Force (N)	Max def. (mm)		Stress (MPa)	
	Schanzes	Device	Schanzes	Device
130	1.1842	No change	207.3	46.865
150	1.3414	No change	246.43	48.274

Simulating the application of compressive force shows that the device has good stability and only the schanzes that are available and used as standard fail. Therefore, the maximum applied compressive force of 150N is approved. According to the Von Mises stress contour and the safety factor obtained in the structure of the bone fixation device without considering the schanzes, the device has significant stability against the applied forces. Using this force, the system's stability will not change. Therefore, the strength and rigidity of the device is desirable and acceptable.

4.2 In the second stage

The pressure test was performed in a laboratory environment to ensure the desired strength and rigidity of the device. The boundary conditions considered for this simulation include the following: It is according to Figure 3, which shows the method of applying the compressive force and the fixed location (bone) in the test. In this test, A compressive force of 150N (although a force more significant than the intended 150N was unintentionally applied during the test) and a displacement of 0.25 mm were applied under standard laboratory conditions (temperature 25 °C and humidity 25%) [10, 17, 19, 21]. After each test, the device is placed on a smooth stone to observe its state due to the application of force. The device was placed on a smooth stone, its condition was examined, and in all cases, the structure of the device was unchanged.

Figure 10: Compressive force applied on four pieces of bone fixed in dynamic mode

After applying the pressure force in dynamic mode on the four-part bone fixed by the standard schanzes attached to the schanze holder of the fixing device (to ensure correctness and accuracy, the test was repeated three times, and the results had very little difference), Based on the results obtained from this test, as shown in Table 6, it was observed that the bone displacement in the compression test with a force of 131.53N was 2.7 mm, By applying a force of 186.53N, it was 3.4 mm, and by applying a force of 229.99N, it was 3.98 mm. This displacement happened only in the schanzes, and the bone (bone replacement tube) and the device were unchanged.

Therefore, according to the results of the pressure test in the dynamic mode, the applied force was more than the considered force (150N), but at the same time, the results obtained are favorable. Therefore, it is concluded that the bone does not change and maintains its stability; the designed and manufactured motorized bone fixation device has desirable stability, strength, and rigidity.

The data obtained from the deformation due to the application of three compressive forces are given in Table 6.

Table 6. The data obtained from the test of compressive force applied on the four-piece bone in dynamic mode

Force (N)	Max def. (mm)	
	Shanzes	Device
131.53	2.7	No change
186.53	3.4	No change
229.99	9.98	No change

By comparing the results obtained from the FEM and the experimental method of applying a compressive force of 150N (in the experimental tests, the amount of force applied was slightly different according to the type of device used) on the quadrilateral bone in the dynamic state, which was determined by schanzes The standard is connected to the Schanz holders of the device and fixed, It can be seen according to Table 7, The results of the shape change obtained from the simulation using the FEM for the application of a compressive load of 150N with the results obtained from the experimental tests, disregarding the existing minor difference, considered the same.

Table 7. Comparison of simulation and experimental results of applying compressive force on the four-piece bone in dynamic mode

Check method	Force (N)	Max def. (mm)		Stress (MPa)	
		Shanzes	Device	Shanzes	Device
Simulation	130	1.1842	No change	207.3	46.865
Simulation	150	1.3414	No change	246.43	48.274
Simulation	131.53	2.7	No change	-------	-------
Experimental	186.53	3.4	No change	-------	-------
Experimental	229.99	3.98	No change	-------	-------

Therefore, with the closeness of the results (there is a slight difference, but it has no effect that has not been taken into account) as a result of the application of compressive forces in the simulation method and the experimental method, where the changes based on the observations and investigations

carried out have only been used within the range of standard schanzes and the structure of the device has not changed.

By reviewing and studying the research conducted and the results of this research, it is concluded that the designed device, unilateral motorized external bone fixator, which is used to fix broken bones and move the four areas of broken long bones, designed and built, has the necessary stability and safety and even if the applied force exceeds the force, which leads to deformation and fracture of the schanzes, the device still maintains its rigidity and stability. Therefore, the fixing structure in the present study corresponds to the structure presented by Alamdin et al. [34].

Therefore, the structure of the bone fixation device, regardless of the schanzes, has significant stability against the applied force. Using a compressive force of 150N, the system's stability will not change in any way. Therefore, the stability, strength, and rigidity of the device are desirable and acceptable, and taking into account the inability to bear this amount of force by the schanzes, there is no justification for applying more force to check the strength of the device's structure [35-36].

5. Conclusion

In the present study, simulation results were performed using ANSYS software using the FEM and experimental tests of the unilateral motorized bone external fixation device, which was designed using SolidWorks software, Simulated, and manufactured in the dynamic mode; it indicates that the designed and manufactured device has the necessary and desirable stability, strength, and rigidity against the application of compressive loads and taking into account the safety factor obtained in the simulated experiment using the FEM, it has been within the acceptable range, So the device has been approved in terms of reliability. Therefore, the designed and manufactured device is a safe, stable, and robust tool that can be used in orthopedics to stabilize fractured long bones. In addition, due to the four independent motorized units of the device, for moving broken long bones in a state where a part of the bone is missing, it can be used for ossification and compensation of the lost bone.

6. References

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