

The Effects of Bone Screw Configurations on the Interfragmentary Movement in a Long Bone Fixed by a Limited Contact Locking Compression Plate

Jalil Nourisa¹, Amin Baseri², Leszek Sudak³, Gholamreza Rouhi^{1*}

¹Amirkabir University of Technology, Faculty of Biomedical Engineering, Tehran, Iran

²Department of Biomedical Engineering, Science and Research Branch, Islamic Azad University, Tehran, Iran

³Department of Mechanical and Manufacturing Engineering, University of Calgary, AB, Canada

Email: *grouhi@aut.ac.ir, grouhi@uottawa.ca

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Abstract

The locking compression plates (LCP) are efficient tools in open reduction and internal fixation (ORIF), especially in osteoporotic bones. Two important factors of screw density and screw position can affect the functionality of the bone plate. Several studies have assessed the influence of the screw configurations on the bone-plate stiffness, but the effects of screw positions on the interfragmentary strain, ϵ_{IF} of LCP construct have not been investigated yet. In this study, finite element method was used to investigate the influence of screws number and position on the interfragmentary strain of LCP-femur system for a mid-shaft fracture. Results of this study showed that by insertion of screws closer to the fracture site, ϵ_{IF} decreases by 2nd degree polynomial function versus screw position, but by adding the screws from the ends of the plate, or by moving and placing the screws towards the fracture site, the reduction of ϵ_{IF} will be linear. Results of this study were compared and are in agreement with some studies in the literature, even though their scope was mostly stability of the bone-implant system, whereas our scope was focused on the interfragmentary strain.

Keywords

Long Bone Fracture, Locking Compression Plate, Interfragmentary Strain, Screw Density, Screw Position

*Corresponding author.

1. Introduction

Femur is the biggest and strongest bone, and one of the main components for load bearing in our body [1]-[3]. Mid-shaft fractures of the femur are very common and one of the most prevalent fractures, especially in car accidents [2]. Comminuted and segmental fractures of mid-shaft femur can be treated by internal fixation systems, such as orthopedic plates specially when there is extremely narrow medullary canal [4]-[7]. Locking compression plates (LCP) by taking the advantage of the threaded holes in the plate, is not pressed to the bone, which this can greatly reduce the biological damage [8]. Because of the threaded holes, the bone-plate structure turns into an integrated structure that increases in the resistance of the screws to the pullout force [3]. Thus, direct contact between the bone and plate is not needed, and the gap between the plate and bone results in a better blood supply to the fracture site, and there are also no need to plate shaping, based on the geometry of the broken bone, during surgery [3] [8] [9].

Fracture healing process is affected by the stability of the bone-plate structure. Based on Perren's study [3], callus generated at the fracture site has different healing rates according to the interfragmentary strain (ϵ_{IF}). The strain range of 2% - 10% guarantees the production of cartilage and woven bone tissue, and smaller strains, below 2%, causes direct bone healing and lamellar bone tissue [3]. Furthermore, strains above 10% result in fibrous tissue production. The stability of fracture site and the healing time are strongly influenced by the surgical techniques, such as the number and positions of screws [3] [8] [10]-[13]. There are many studies, which investigated the influence of screw position in the conventional plates on the stability of the bone-implant system experimentally and clinically [14]-[18]. There are some other studies that investigated the effects of screw configurations on the stability of the locking plates [10] [19]-[21].

Stoffel *et al.* [10] investigated the importance of the position and density of the screws on the axial stiffness and torsional rigidity of the bone-LCP construct. They demonstrated that by increasing the number of screws, the rigidity of the structure increases, and by increasing the distance of screw from the fracture site, the bone-plate construct's stiffness decreases. MacLeod *et al.* [22] added the effect of bone quality on the response of construct to the screws positions and reported similar results with Stoffel and co-workers [10]. In an earlier study, Lee *et al.* [21] employed an optimization method to determine the best combination of number and position of screws to optimize the stability of the bone-plate system. All these studies [10] [21] [22] investigated the effects of the screw configurations on the mechanical properties of the bone-implant construct, but did not focus on the effects of screw configurations on the bone fracture healing process. The main scope of the present study was to determine the relations between the screw configurations and the interfragmentary strain (ϵ_{IF}), which is an indicator of the tissue phenotype during the healing process, in a femur-LCP construct.

2. Materials and Methods

2.1. Construction of the CAD Model

To create a CAD model, CT data of a 65 years old male's left femur were used by considering the intramedullary canal, and cross-sectional images at every 1 mm slice along the longitudinal axis of the femur. The point clouds obtained by Mimics (V.10.01) were imported to Catia (V5.R19) for making the solid structure. For the sake of simplification, only the proximal fragment of femur was created by Mimics and for modeling the diaphysis, the cross section of the mid-shaft was extruded 150 mm (**Figure 1(a)**). To simulate a 3 mm transverse fracture gap in the mid-shaft of the femur (AO classification: 32-A3 [23]), the distal fragment of the femur was placed 3 mm away from the proximal part.

An LCP (type = broad; 12 holes; length = 239 mm; width = 17.5 mm), along with bone screws (screw core diameter = 5 mm; outer diameter = 5.5 mm) was used to fix the fracture. In order to reduce the complexity of the finite element analysis, the threads of the screws were all neglected, and they were modeled as a cylinder with a truncated cone head (**Figure 1(b)**, right). In accordance with the AO procedure [23], the plate is positioned in the lateral side of the femur. In general, the locked constructs lose their stability when the distance between the bone and plate exceeds a limit [24]. Ahmad *et al.* [24] suggested that the distance between the LCP and bone surface should not exceed 2 mm. Based on Ahmad and co-workers' results [24], a band analysis was performed in Catia to make sure that the plate is located in a way, where the distance between bone surface and plate does not exceed 2 mm.

Nine different screw configurations were developed, which divided into 3 groups with 4 different configurations

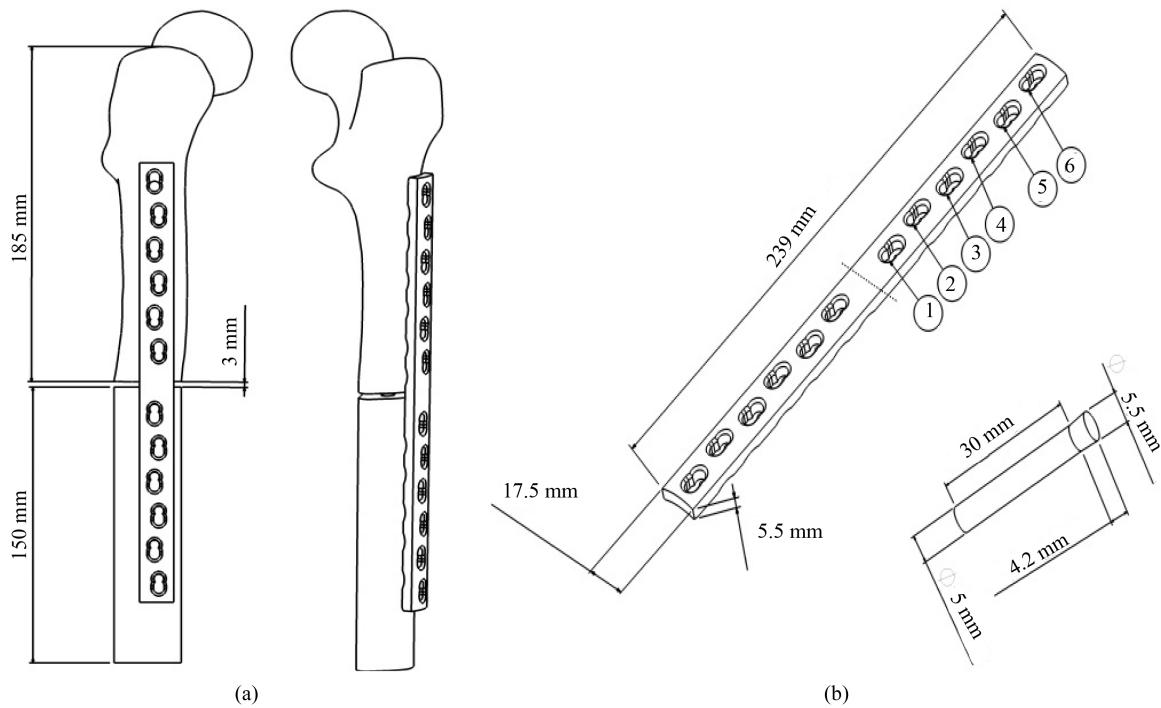


Figure 1. CAD model created in Catia, (a) Assembly of bone-plate. The outer diameter of femoral shaft was around 30 mm, and the inner diameter was 20 mm; (b) Specifications of LCP and screw.

in each group (**Figures 2-4**). In groups 1 and 2 screws were added from the fracture gap side and lateral side, respectively (**Figure 2** and **Figure 3**). In group 3, with constant number of the screws, the second screw stepped towards the plate ends (**Figure 4**). It is highly recommended that the number of screws in each main fragment of the bone should not be less than three [8] [10] [11] [19]. It was reported that the closest screw to the fracture site plays the most important role in the stiffness of the bone-plate structure [10]. Thus, in this study, all models contain the 1st screw inserted closest to the fracture site, and also not less than 3 screws were used in each fragment side.

By defining $\Delta L1$ and $\Delta L2$ as the interfragmentary movement in the medial and lateral side of the femur, respectively (**Figure 5**), the interfragmentary strain, ε_{IF} is determined as the average of $\Delta L1/L$ and $\Delta L2/L$ (**Equation (1)**).

$$\varepsilon_{IF} = \frac{\Delta L1 + \Delta L2}{2L} \quad (1)$$

2.2. Finite Element Analysis

Bone-plate complex was constructed in Catia and then was imported to Abaqus (V 6.11-1, Simulia, Dassault Systems). The plate and screws were assumed to be made by stainless steel, with a Young's modulus of 191 GPa, and Poisson's ratio of 0.3 [25]. In accordance with Wang *et al.* [26], the proximal fragment of the femur was divided into 3 regions with various material properties (**Figure 6**). Material properties of these 3 regions are listed in **Table 1**. It was assumed that bone is homogeneous and linear elastic, and also bone marrow and callus were disregarded in our FE model.

The contact areas of the screw-plate and screw-bone were assumed to be tied together. To simulate the different load regimes for different weights, loads of 600, 700, 800 and 900 N were applied to the femur head vertically, in X direction (see **Figure 7**), and the end of the distal part was fixed with no degrees of freedom (**Figure 7**). In order to reduce the complexity, muscle forces were neglected in our model.

The types of elements and the average number of elements for each part are listed in **Table 2**. The ten node tetrahedral element (C3D10) was used for all different parts of the bone-plate construct. The effect of mesh

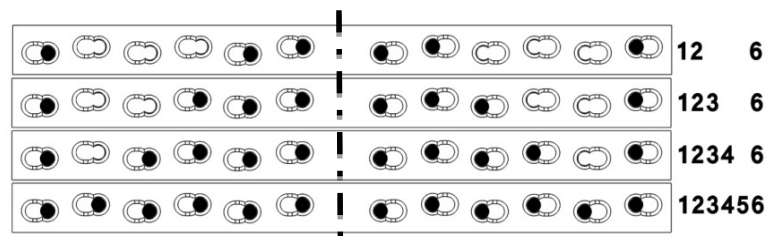


Figure 2. Screw configurations in group 1.

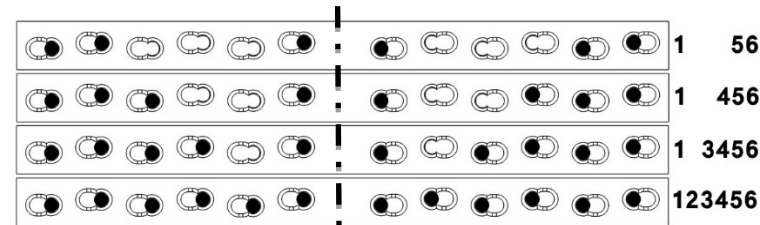


Figure 3. Screw configurations in group 2.

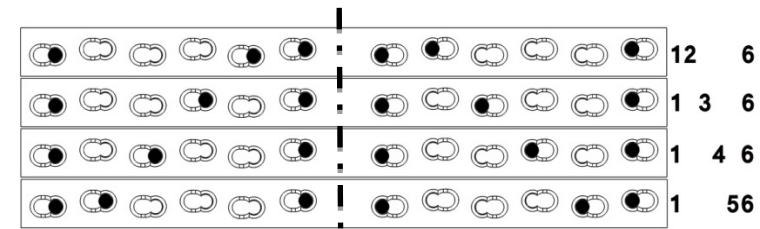


Figure 4. Screw configurations in group 3.

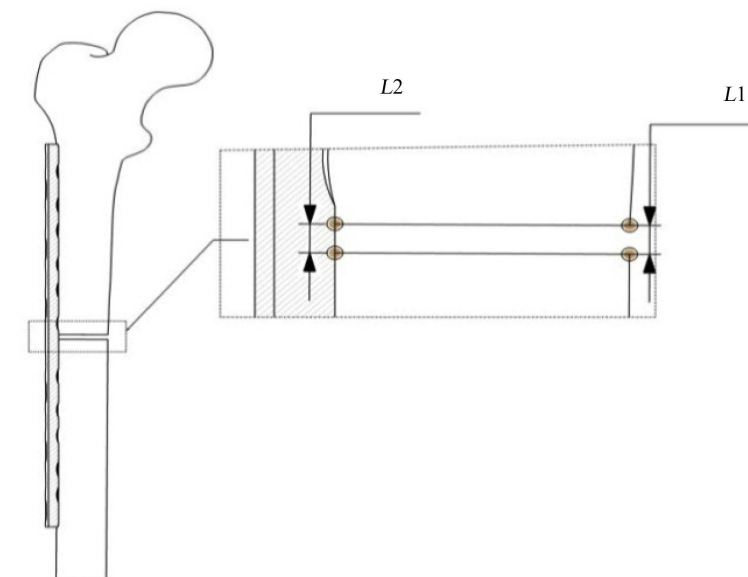


Figure 5. The initial fracture gap lengths in the medial ($L1$) and lateral ($L2$) sides of the femur ($L1 = L2 = L = 3$ mm).

refinement was explored by increasing the local seeds in the high stressed area especially around the screw holes. The number of elements was increased until the third decimal of interfragmentary movement (mm) reached a

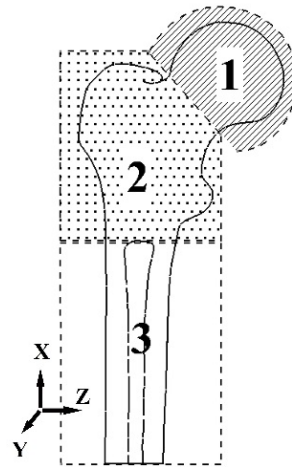


Figure 6. The proximal femur was divided into 3 different regions; high density spongy bone (region 1), low density spongy bone (region 2), and cortical bone (region 3) [26].

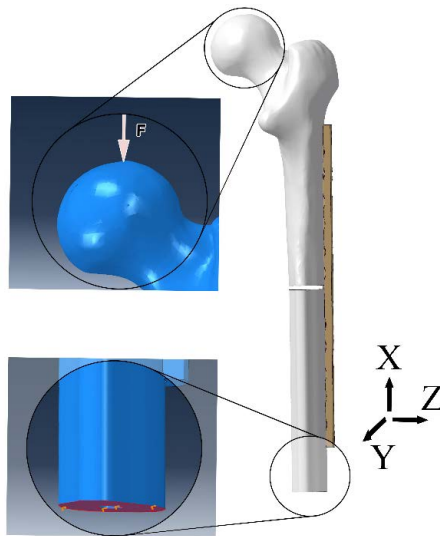


Figure 7. Loading and boundary condition: External load was applied on the femoral head and the distal end was fully constrained.

Table 1. Material properties of different regions of femur (see Figure 6).

Region	Young's modulus (GPa)	Poisson's ratio
1 [26]	$E = 1.3$	0.3
2 [26]	$E = 0.32$	0.3
3 [25]	$E_x = 18.4$	$\nu_{xy} = 0.12$
	$E_y = 7.00$	$\nu_{yz} = 0.12$
	$E_z = 8.50$	$\nu_{xz} = 0.14$

constant value, and the average seed size was 1mm in the refined regions.










3. Results

Results of all simulations can be seen in Table 3. For all different configurations, ϵ_{IF} is greater than 0.1 for the

Table 2. Number and type of elements used in the FE model for femur proximal fragment, femur distal fragment, LCP and screw.

Part	Element type	Number of elements
Femur proximal fragment	C3D10	207,000
Femur distal fragment	C3D10	119,000
LCP	C3D10	33,000
Screw	C3D10	14,000

Table 3. Interfragmentary strains (ϵ_{IF}) for nine different configurations of screws placement (see **Figures 2-4**).

Screw configuration		The interfragmentary strain (ϵ_{IF})			
		F = 600 N	F = 700 N	F = 800 N	F = 900 N
	126	0.058	0.079	0.102	0.125
	1236	0.056	0.076	0.097	0.118
	12346	0.054	0.073	0.093	0.114
	123456	0.054	0.073	0.093	0.114
	13456	0.061	0.080	0.106	0.130
	1456	0.069	0.088	0.118	0.144
	156	0.073	0.095	0.125	0.153
	146	0.069	0.091	0.118	0.144
	136	0.062	0.083	0.107	0.131

external load of 900 N, and lower than 0.1 for the external loads of 600 N and 700 N (see **Figure 8**). Based on Perren's theory [3], results of this study show that for an external load between 700 N to 800 N, the screw positions are the most critical for changing the tissue phenotype (see **Figure 8**).

The relations between ϵ_{IF} and the screws configurations for different external loads in groups 2 and 3 are linear (**Figure 10** and **Figure 11**), and in group 1 are polynomial (**Figure 9**) (R-square > 0.95). **Figure 8** shows, there is a linear relation between the load and the interfragmentary strain for all different configurations. Furthermore, slopes of ϵ_{IF} versus load are steeper for the less stable bone-implant constructs, e.g. 156 configuration compared to 123456, which the latter is more stable.

4. Discussion

In this study, an investigation was made on the impact of screw position on the interfragmentary strain in a LCP-femur construct. In group 1 (see **Figure 2**), insertion of more than 4 screws showed a little and marginal effect on the interfragmentary strain (see **Table 3** and **Figure 9**), in agreement with Stoffel *et al.* [10] and Lee *et al.* [21]. In group 2 (see **Figure 3**), because the pattern of screw insertion has been changed, increase in the number of screw more than 4 has same effect on the interfragmentary strain as from 3 to 4 (see **Table 3** and **Figure 10**). The latter result was not reported by other investigators, since these specific patterns of screws (**Figure 3**) were not investigated by others, and suggests that when insertion of the screw number 2 (see **Figures 2-4**) is not possible, to increase the stability of the system, insertion of more screws is needed. In group 3 (see **Figure 4**), by stepping the second screw to the middle of plate, the interfragmentary strain will decrease, in agreement with Stoffel and co-workers' results [10], and the reduction pattern of ϵ_{IF} follows a linear function with respect to the screw position (see **Figure 11**).

LCP with a small number of locking screws can reduce both bone and soft tissue damage, as well as decreases bone-implant stiffness that can ultimately lead to a reduction in the stress shielding effect [27]. By comparing results of configurations 126 with 1456, and 13456 and 136 with 1456 (see **Table 3**), it can be concluded that proximity of the screws to the fracture site has a greater impact on the interfragmentary strain (ϵ_{IF}) than the number of screws itself. This result is in agreement with the finding of Lee *et al.* [21], who showed that LCP

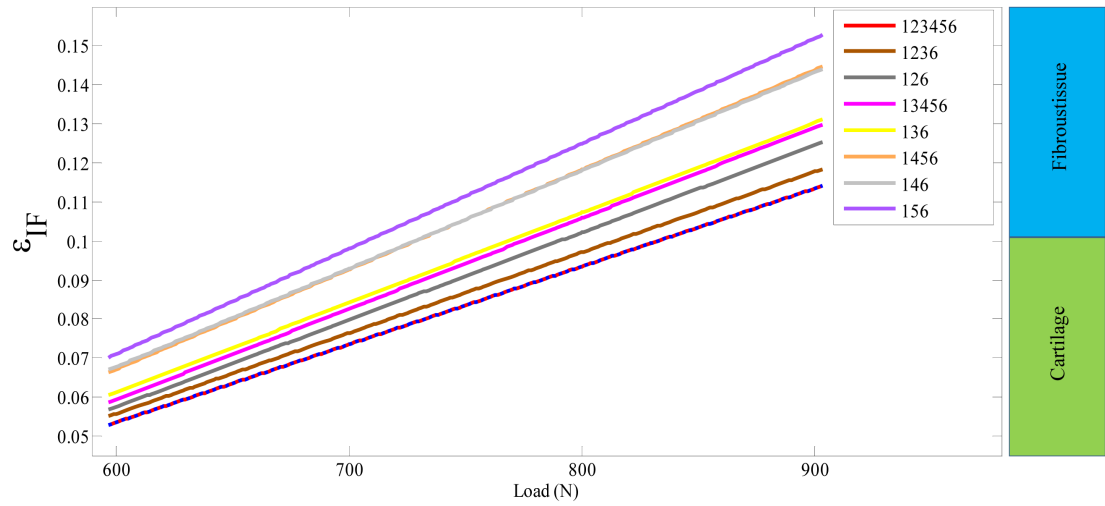


Figure 8. The interfragmentary strains (ϵ_{IF}) versus external load for different configurations of screw placement.

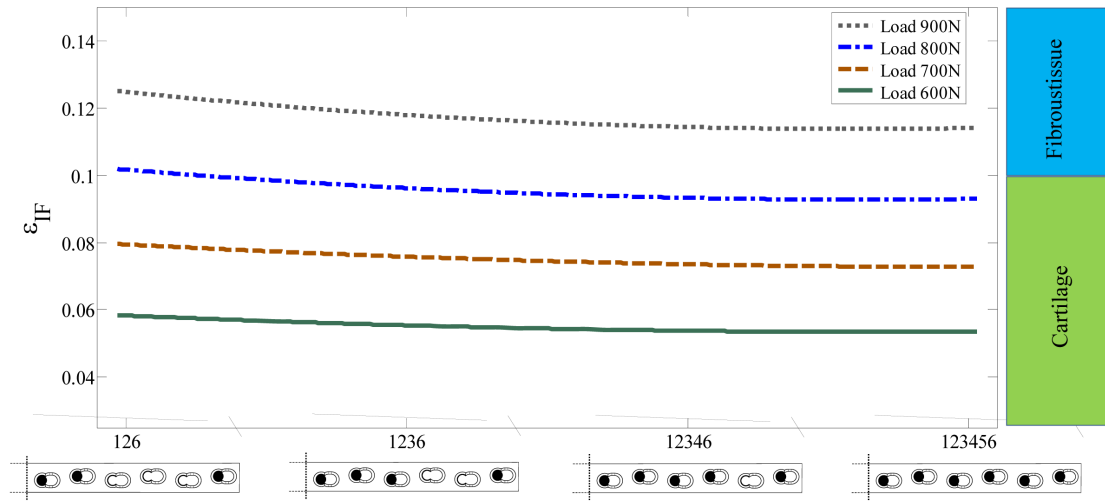


Figure 9. The interfragmentary strain (ϵ_{IF}) for different screw configurations of group 1.

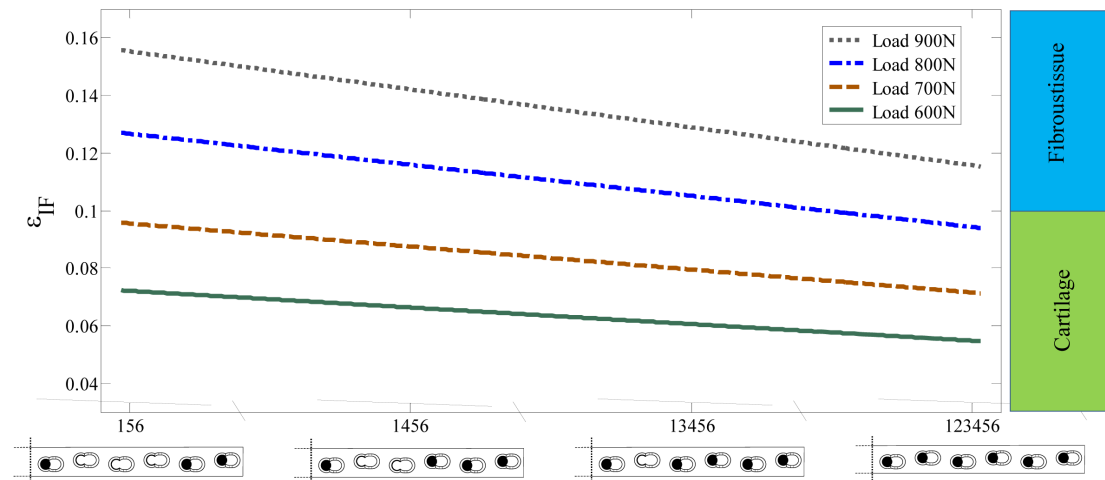


Figure 10. The interfragmentary strain (ϵ_{IF}) for different screw configurations of group 2.

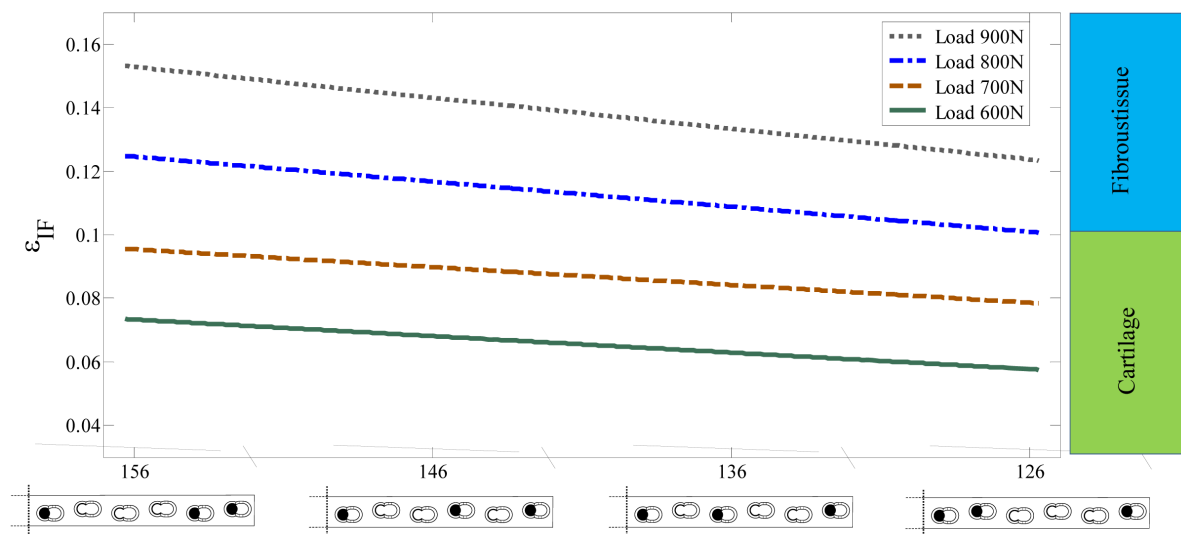


Figure 11. The interfragmentary strain (ϵ_{IF}) for different screw configurations of group 3.

with six locking screws results in the necessary fixation stability, and the optimized screw configuration is 126, three locking screws on either side of the bone fragment with two locking screws right beside the fracture line. This finding can be used by the surgeons to choose the appropriate selection of screws in order to achieve the same healing rate, but by using less screws, and consequently less damage to the soft and hard tissues, and less surgery costs. More investigations, especially well-defined *in-vivo* tests on animal models are needed to reinforce our findings.

The relationships between the interfragmentary strain (ϵ_{IF}) and the screw configurations were investigated for four different constant values of external load, *i.e.* 600 N, 700 N, 800 N or 900 N, which represent load bearing after surgery for 4 different weights (60 - 90 kg). Based on Perren's theory [3], it was shown for the majority of loads, change in the screw configurations cannot greatly affect the tissue phenotype (see Figure 8). This fact is partly due to the lack of exact relationships between the interfragmentary strain and tissue phenotype, for instance, for the strains in the range of 2% - 10%, and 10% - 30% in Perren's theory [3]. In order to more accurately predict the impact of screws configuration on the outcome of internal fixation, more quantitative mechano-biological models, such as [28] [29], should be employed in the future studies. Furthermore, results of this study show that the screw positions are the most influential for an external load between 700 N to 800 N (see Figures 8-11). Needless to emphasize that by changing either of the geometry and/or material properties of bone and/or implant, the critical load with the greatest impact on the tissue phenotype will be different.

Figure 8 shows the influence of the load magnitude on the interfragmentary strain (ϵ_{IF}). It indicates that by an increase of 200N in the load magnitude, the tissue phenotype can completely change. It is interesting to note that for vertical external loads, altering the screws positions, from the most stable configuration (configuration 123456) to the most unstable configuration (configuration 156), cannot alter the tissue type. This result implies that change of load regime has greater impact on the interfragmentary strain in comparison with changing the screws positions (compare Figure 8 with Figures 9-11).

The effect of gap size on interfragmentary strain (ϵ_{IF}) was not investigated in this study. However, in the bone-LCP construct, because there is no contact between plate and bone (biological fixation) [3] [8], the gap size should not have considerable influence on the stiffness of the bone-implant system and the interfragmentary movement (as reported by [30] [31]). The interfragmentary strain (ϵ_{IF}) that is defined by the interfragmentary displacement over the initial gap length (see Equation (1) and Figure 5), has an inverse linear relationship with the initial gap size. Based on these explanation, it can be concluded that results obtained in this study can be extended to the models with a gap greater than 3 mm by applying an appropriate coefficient. It should be noted that because of the possibility of bone contact for gap size of smaller than 3 mm, the above mentioned point might not be true.

This study has some limitations. No experimental tests were conducted in this research, and the validation of our results was done by comparing with others' studies reported in the literature. Verification of the developed

numerical model can increase the impact of the results. Moreover, as it is difficult to apply anatomical boundary condition to the condyle of femur, in most studies, boundary conditions are simplified through constraining the distal end of the femur rigidity [21] [32]-[36]. Geometrical complexity of condyle was also disregarded in this study, and the distal fragment of femur was modeled by a cylinder (see **Figure 1**). These simplifications reduce the computational costs and can help address convergence issues (see **Figure 7**) [26] [33]. To simulate the physiological condition during normal gait, the effective muscle forces, *i.e.* abductors and vastus lateralis [32] [34] [37] [38] should also be considered, which was another limitation of this work. Furthermore, in the present investigation, only vertical component of the hip reaction force was modeled, but according to [32]-[34] [37]-[39], consideration of two other components of hip reaction force, *i.e.* loads in the z and y directions (see **Figure 6**), can provide us with more accurate results. In this study only axial interfragmentary movement was investigated, but it should be concerned that by considering muscle forces and other components of hip reaction forces, shear interfragmentary movement plays a more significant role in the production of the fibrous tissue [40] and consequently will result in delay in union or non-union [41] [42], that should be investigated in the future studies.

Modeling the interaction between the bone and screw is a challenging task, which was tackled differently in various studies [36] [43]. In this study, we used tied contact in which the outer nodes of the screw were bonded to the surface of the bone holes, and subsequently there is no relative displacement between these two surfaces. This approach implies that there was full osseointegration between bone and screw, which is not a realistic assumption especially in the early phase of bone healing process. To simulate bone-screw interaction more realistically, the quality of interconnection of the bone-screw surfaces should increase gradually from more poorest osseointegration, by considering no friction between the surfaces of the bone-screw, to low friction that implies imperfect osseointegration [43] [44] and ultimately "Tie" contact that represents full osseointegration [36] [43]-[45]. Besides, in order to have a more realistic simulation of the bone-screw construct, geometry of the screw must be modeled more realistically by considering the screw threads [43] [45], and then stress shielding as a crucial factor in determining failure of screw-bone could be investigated [46].

5. Conclusion

In this study, the effects of screw number and position on the interfragmentary strain, and the relations between the screw configurations and the interfragmentary strain (ϵ_{IF}) in a femur-LCP construct were investigated. It was shown that inserting two innermost screws is the most influential factor in interfragmentary strain, while the effects of other screws, as well as screw numbers are less important. Thus, an optimal configuration of the screws for minimizing the healing time and also the cost of surgery might be a configuration with a small number of screws, which are close to the fracture site, considering appropriate magnitude of external load, as a crucial contributor.

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