Stress analysis of implant-bone fixation at different fracture angle

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Abstract. Internal fixation is a mechanism purposed to maintain and protect the reduction of a fracture. Understanding of the fixation stability is necessary to determine parameters influence the mechanical stability and the risk of implant failure. A static structural analysis on a bone fracture fixation was developed to simulate and analyse the biomechanics of a diaphysis shaft fracture with a compression plate and conventional screws. This study aims to determine a critical area of the implant to be fractured based on different implant material and angle of fracture (i.e. 0°, 30° and 45°). Several factors were shown to influence stability to implant after surgical. The stainless steel, (S. S) and Titanium, (Ti) screws experienced the highest stress at 30° fracture angle. The fracture angle had a most significant effect on the conventional screw as compared to the compression plate. The stress was significantly higher in S.S material as compared to Ti material, with concentrated on the 4th screw for all range of fracture angle. It was also noted that the screws closest to the intense concentration stress areas on the compression plate experienced increasing amounts of stress. The highest was observed at the screw thread-head junction.

1. Introduction
Bone fractures occur throughout the population; causes can be the result of high load impact or minimal trauma injury. Statistical analysis has been carried out whereas the significantly, majority 81% sustained the fracture due to road traffic accidents. The rest is equally divided between low-energy trauma such as fall from height and assault (9.5%). The report stated the percentage of union (19.04%) achieved by 4 month, (66.66%) by 6 months and the rest complete union by 9 months [1]. However, the result may vary due to absolute the stability of bone fracture fixator and early range of motion to minimize the risk of reoperation.

Femur bone, subjected to a combination of axial, bending and torsion load that provides the variety type of fractures when the applied load exceeds its strength limit. Axial compression load can cause oblique fracture while the torsion delivers to the transverse fractures [2]. The different between
these are the fracture patterns, which can be classified by the angle of fractures. According to the Müller AO Classification of fractures—long bones, it was classified as category 32-D. Sub-categories complete transverse with an obliquity of 30° or less and complete oblique or spiral more than 30° are simple fractures and the complex fractures [3]. However, the treatment of bone fracture continues to be a challenging problem for a surgeon particularly involving the complication of surgical approaches [4]. It is necessary to provide the orientation of the bone fragments and promotes the tissue regeneration. Thus, the precise implant device is needed to support the fractured bone and restore compromised functional.

Figure 1: Load stress distribution in femur bone [5]

Besides that, mechanical loading applied to the bone structure allows the subsequent calculation of stress induced, for example, in femur bone structure as shown in Figure 1. The fracture pattern generated involving various soft tissues and the bone quality at the time of injury. This is further compounded by the fact that in these types of fractures, the articular surface may be involved which, if inappropriately treated, may lead to future complications such as osteoarthritis. The treatment of these fractures requires complete immobilization of patients who, ultimately result in prolonged recovery periods [6]. Thus, the study of proper internal fixation was necessary whereas it offers anatomical reduction of the bone fragment, stability and rigid of fixation for better formation of soft tissue and bone healing process.

Some researchers studied the stress analysis by using the finite element to determine the fracture mechanism of the long bone. They found that fracture angle 30° degree causes the highest stress on the screw and plate [7]. The bending stiffness and contact condition effects were considered [8]. It is expected to reduce contact with other living tissues may help diminish the foreign body reaction and thus increase curing efficiency. Similar to the investigation the plate failure reported takes place at the screw hole near the oblique 45° fracture site [2]. These studies reported that screw and plate have a significant effect on the implant failure. However, these studies were limited to the implant design geometry and materials used whereas should highlight the optimal parameters granting the implantation success.

Progressive losing of bone fixation screws by stress shielding and causing adaptive bone remodeling due to bone loss around the screw [9]. Bone adaptation caused losing implant due to the stress-shielding pattern which dependent on material and geometrical characteristics of the screw [10]. While the primary stability depends on the immediate mechanical engagement between the implant and the bone seat, the secondary stability requires the establishment of a resistant bone-implant adhesion. These research efforts pursue a clear goal: improve the implants’ efficiency in standing conditions, i.e. as bone fracture fixator subjected to stress due to the mechanical loading. These are the
conditions in which the bone-implant ‘assembly’ must grant stability and reliability. However, the high-stress concentration due to load can cause the loosening of the screw, and results implant failure and infection occurs on the fracture site. These study aims to determine a critical area of the implant to be fractured based on different implant material and angle of fracture (i.e. 0°, 30° and 45°).

2. Finite Element Modeling
The three-dimensional geometrical model of the bone fracture fixation and femur bone structure was created using Solidwork software 2013. The geometry of femur bone took the shape created from CT database through image segmentation with IGES format. All assembly processes of the screw into femur bone model were performed in Ansys 18v software by using Boolean operation in design modular as shown in Figure 2. The bone fracture fixation model comprised compression plate and four internal conventional cortex screws (full threaded) anchored along the diaphysis shaft as shown Figure 3. The fixation or implant was placed in the middle of the fracture line. Three type of the fracture line developed (i.e. 0°, 30° and 45°) and the screws orientation also used based on the fracture angle.

Table 1 describes the details of bone fracture fixation for conventional cortex screw and compression plate based on Synthes product. In this study, two type screw lengths (i.e. 36 mm and 40 mm) and two types of material (i.e. Stainless Steel (no product : 204.8xx) and Titanium (no product : 404.8xx)) were used where “xx” refer as screw length. These types of bone fracture fixation chose based upon the fracture type suggested based on AO Foundation.

Table 1: Dimensional parameters for bone fracture fixation

|                        | Conventional Cortex Screw | 8-holes compression plate |
|------------------------|----------------------------|---------------------------|
| Diameter of thread (mm)| 3.5                        | Width (mm)                |
|                        |                            | 11.0                      |
| Thread pitch (mm)      | 1.25                       | Thickness (mm)            |
|                        |                            | 3.3                       |
| Diameter of Core (mm)  | 2.4                        | Distance between center holes (mm) |
|                        |                            | 13.0                      |
| Diameter of head (mm)  | 6.0                        |                           |

Table 2 presents the information regarding the material properties of bone; cortical bones were assumed to be orthotropic, homogeneous, and linearly elastics as were the others’ materials used in this study.
### Table 2: Mechanical properties of bone structure and implant materials in FE analysis

| Material          | Young’s modulus (GPa) | Poisson’s ratio (\(\nu\)) | Shear modulus (GPa) |
|-------------------|-----------------------|----------------------------|---------------------|
| Cortical Bone     | \(E_3 = 20.0\)        | \(\nu_{12} = 0.376\)      | \(G_{12} = 4.53\)   |
| \((\text{Longitudinal transverse})\) | \(E_1 = 12.0\)        | \(\nu_{23} = 0.235\)      | \(G_{23} = 4.53\)   |
|                   | \(E_2 = 12.0\)        | \(\nu_{23} = 0.376\)      | \(G_{13} = 4.53\)   |
| Stainless steel (S.S) | \(E_{S.S} = 200\)  | \(\nu_{S.S} = 0.3\)       | -                   |
| Titanium (Ti)     | \(E_{Ti} = 113.8\)    | \(\nu_{Ti} = 0.34\)       | -                   |

Femur bone model was categorized into five parts, which separated by using virtual topology to combine the vertical edge. It can be done by inserting a virtual topology branch. Thus, the model was separated into four parts, which are a femoral head, greater trochanter, diaphysis shaft, and condyle as shown in Figure 4. The surface that separates for each part has been considered as a bonded region in order to connect the part together with the elements of the upper part overlap with the elements of the lower part. The fracture section represents the fracture line whereas divided the diaphysis shaft into two section (i.e. upper and lower part). The static load based on standing conditions was applied by tension load 400N (body weight) and compression load 300N (abdomen muscle) as shown in Figure 4, to produce significant influence instability.

An optimized convergence meshing has been developed using ANSYS Workbench an appropriate setting and values have been executed in order to use smaller elements on proximities and curvatures for the model. The numbers of elements used for the bone fracture fixation model are 1058958, 1078497 and 1075567 for fracture angle of 0°, 30° and 45° respectively.

![Figure 4: Boundary condition of bone fracture](image-url)
3. Result

The primary stability of implant is a reference of the mechanical loading attained between screw surface and bone structural designed. This tension and compression loads is analyzed in terms of the implant bending, mainly in situations where a non-Osseo-integration device response is planned for supportive purposes. In this work, the minimal static loads based on standing conditions was applied by compression load 400N (body weight) and pulling load 300N (abduent muscle), at the different of fracture angle (0°, 30° and 45°). The magnitude of the applied forces was in accordance with the orthopedic demand for clinical needs.

All the different fracture angle and configurations of fixation were subjected to the same load, each one evaluated separately regarding the critical stress of implant. The stress results, even in the bone, screws and compression plate are presented in Table 3.

| Materials     | Fracture type | Maximum von-Mises stress (MPa) |
|---------------|---------------|-------------------------------|
|               | Bone stress   | Conventional screws | Compression plate |
| Stainless steel (S.S) | Transverse (θ=0°) | 152.10 | 215.03 | 274.21 |
|                | Transverse (θ=30°) | 175.12 | 335.59 | 234.26 |
|                | Oblique (θ=45°) | 185.96 | 332.06 | 232.61 |
| Titanium (Ti) | Transverse (θ=0°) | 154.19 | 163.46 | 195.34 |
|                | Transverse (θ=30°) | 177.25 | 222.5  | 154.64 |
|                | Oblique (θ=45°) | 182.25 | 221.47 | 153.73 |

Table 3 shows the maximum von-Mises stress for bone, screws and compression plate at the different fracture angle, and implant materials. The bone stress showed inversely proportional to the stress of compression plate. Dissimilar to the conventional screws it was challenging to correlate with bone stress due to unevenly force transfer to each screw. Nevertheless, the early assumption could be made which the higher screw stress will tend to optimize the stability of the fixator. This assumption meets the statement from some researchers that verified the higher the screw tension will reduce the displacement of bone segment [11]. Although, the stress of bone shows an increment as increasing the fracture angle for both implanted materials. The bone stress value of Ti implanted shows slightly higher than S. S implant. It was compared to the effect of increasing the fracture angle to the implant stress (screws and compression plate), its shows independent of each other. Thus, a comparison was made to analyze the affected parts that influence by the fracture angle and implant materials. The details of implant stress could describe in a data as shown in Figure 5.
The maximum equivalent von-Mises stress implants were shown in Figure 5. The Ti material for both the compression plate and screws materials experienced a lower amount of stress as compared to the S.S at fracture angle range (0°, 30° and 45°). This difference in stress was also observed on the screws of all range of fracture angle. The both materials of screws that anchored through diaphysis shaft on the femur bone experienced higher stress at 30° fracture angle as compared to the 0° and 45°, with mean differences of 335.59MPa and 222.5MPa for S.S and Ti material respectively. Dissimilar with the stress of compression plate whereas, it indicates decrement as increasing the fracture angles for both materials.

![Stress Contour of S.S Compression Plate](image)

**Figure 6**: Stress contour of S.S compression plate (a) at 0°, (b) at 30°, and (c) 45° fracture angle

![Stress Contour of Ti Compression Plate](image)

**Figure 7**: Stress contour of Ti compression plate (a) at 0°, (b) at 30°, and (c) 45° fracture angle

In the stress simulations, green color demonstrates less stress, and the red color is related to higher stress values. Figure 6 and Figure 7 shows the stress contour of compression plate for S. S and
Ti material. The stress distribution on the implant also was concentrated on the compression plate supporting anterior section of the femur whereas, the segment around the empty screw holes 4th on the section of the plate had the highest stress concentration as shown in Figure 6 (a) and Figure 7 (a). Comparing to stress distribution at 30° and 45° for both materials, maximum stress concentrated at 1st screw hole as shown in Figure 7 (b) and (c) and Figure 7 (b) and (c). It was due to the inclination of bending of the upper diaphysis shaft. However, a reduction of stress for the compression plate was shown by increasing fracture angle.

Figure 8 and Figure 9 shows the stress contour of conventional screws for S.S and Ti material. It was also noted that in both materials, the screws closest to high up concentration stress areas on the compression plate experienced a higher amount of stress. Both materials observed that the highest stress concentration at the screw neck and the upper thread for each fracture angle, as shown in Figure 8 (a), (b) and (c) and Figure 9 (a), (b) and (c). However, the stresses’ distribution appears to decrease along thread as it approaches the tip of the screws. Thus, screws shows had a major influence of fracture for all range of fracture angle compared to the compression plate. Hence, further studies using biomechanical testing on the screws models will need to establish the details of information in order to provide better accuracy of the screw failure.

The present study also established that conventional screw provides better stability for the different type of fracture regarding diaphysis shaft. Figure 10 indicates the results of the interfaces point of these two sections (bone-implant) undergo high stresses as a result of pull-out strength on the 4th screw created by the deformation of bone fragments. The present of screw thread, allow the implant to hold the bone fragmented and maintain the stability of bone-implant interaction [12]. However, the conventional screw implanted in each bone fragment will provide a critical section to fracture that will affect the surrounding tissue.
Figure 10: Stress concentration on the screw thread

Figure 11: Screws orientation at different fracture angle.

Figure 12 (a): Screws stress of S.S and Ti materials at 0° fracture angle

Figure 12 (b): Screws stress of S.S and Ti materials at 30° fracture angle

Figure 12 (c): Screws stress of S.S and Ti materials at 45° fracture angle
Figure 11 (a) shows the orientation of four screws along the compression plate through the diaphysis shaft. The stress concentration was determined on the individual screws (i.e. screw 1, screw 2, screw 3, and screw 4) at the different fracture angle. Figure 11 (b) shows the screws implant straight through the diaphysis shaft compare to screws orientation in Figure 11 (c) and Figure 11 (d) as the screw 3 placed till 30° based on surgical approach.

Based on screw orientation, the maximum principal stress of both implanted material was determined after load applied. The stress was significantly higher in S. S material as compared to the Ti material, with concentrated on the 4th screw for all range of fracture angle as shown in Figure 12 (a), (b) and (c). It was due on bending of the upper diaphysis, whereas the intensities of bending transferred. Thus, the pull-out force generates upon the bending on the 4th screw, and cause the higher stress concentration especially on 30° fracture angle. This would lead the likelihood of implant failure to occur. A reduction in stress concentration points on 2nd and 3rd screws for all fracture angles as shown in Figure 12 (a), (b) and (c).

In addition, the present study analyzes the stresses of a static load, which if cyclical loading were applied would result in the increased likelihood of implant loosening. The reason for determined stress concentration in the screws remains unclear since the various types of thread design, and the material used for implants. It can be expected that the critical region of implant would fail for the internal implant fixation of the diaphysis shaft.

Several limitations were observed in this study. Static loading was used in the analysis for this model whereas, in the real-life situation, cyclic loading is usually applied. This would have resulted in different outcomes being observed. Instead, findings in this study may be demonstrative to this type of implant fracture pattern. Moreover, fracture configuration is also shown to be one of the criteria to determine the screw orientation to be used for fracture fixation.

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5. Conclusion
The stainless steel (S.S) and Titanium (Ti) screws experienced the highest stress at 30° fracture angle. The fracture angle had the most significant effect on the conventional screws as compared to the compression plate. The stress was significantly concentrated on the 4th screw for all range of fracture angle. The highest was observed at the screw thread-head junction.

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