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# How a pilot hole size affects osteosynthesis at the screw–bone interface under immediate loading

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ABSTRACT

An inappropriate pilot hole size (PHS) is one of several factors that affects the stiffness of the screw–bone fixation. The present study uses finite element models to investigate the effect of varying the PHS on the biomechanical environment of the screw–bone interface of the fractured bone, after the screw insertion and under the immediate body weight pressure (BWP). Four PHS from 71% up to 85% of the screw external diameter (SED) were considered for analysis. A non linear material behaviour of the bone with ductile damage properties was used in the study. To validate the numerical models, an experimental pull-out test was carried out using a synthetic bone. The results of the insertion process demonstrated that the relatively smaller holes (71% and 75.5% of SED) increased the insertion torque value within the recommended level, caused more bone radial extension deformation and maximized the contact area between the bone threads and the screw, in comparison to the PHS higher than 80% of SED. Under the immediate BWP after osteosynthesis, the stress level exceeds the elastic limit and becomes high enough to initiate the ductile damage of the bone. Also, enlarging PHS from 71% to 75.5% of SED increased the bone microdisplacement at the screw–bone interface from 75 up to 100 $\mu$m, and that reduced the stiffness of the fixation.



## 1. Introduction

Plates and screws are widely used as an internal fixation for the various bone fractures [1]. The success of this conventional internal fixation depends usually on the implant primary stability that occurs immediately after the implant insertion [2]. Among several factors, the primary stability is affected by the micromovement of a screw as a result of the low stiffness of screw–bone interface [3]. Other researchers have reported that the holding power of the screw depends on the changes induced in the bone by the insertion trauma as well as the bone reaction around the implant [4]. It is also known that the anchorage of the screw in the bone depends necessarily on the bone material, the implant design, the size and the preparation technique of the pilot hole [5–8]. The mechanical relations between the implant and the bone are established by the anchorage of the screw to the wall of the pilot hole. The tight fit between the screw and the bone is in relation with the size of the pilot hole and could improve the quality of the implant fixation [4,5].

In experimental tests, most of the studies have investigated the influence of varying the pilot hole sizes (PHS) on torque measurements and pull-out analysis of the osteosynthesis screws [3–8]. Using in vitro tests, Heidemann et al. [5] stated in accordance with Gantous and Philips [6], that the PHS may be increased up to the critical size of 85% of the screw external diameter (SED) without affecting the holding power of the screw. On the other hand, using in vitro test, Boyle et al. [7] suggested that a maximum difference between the SED and the drill size is recommended for an enhanced pull-out strength. In fact, Kuhn et al. [8] stated that the difference between the screw diameter and the drill size affected the plastic deformation areas and the maximum radial extension of the pilot hole. In a few instances, Macleod et al. [9] have found when comparing different contact interfaces of screw–bone numerical models, that the undersized pilot hole simulation has a considerable impact on the local stress–strain environment at the screw–bone interface after the insertion process and under the body weight pressure (BWP). An appropriate study of the effect PHS on the stiffness of the screw–bone interface requires knowledge of the load and the stress transfer between the implant and the surrounding bone (the cortical and cancellous bone), as well as the bone deformation. Experimental studies are restricted to the insertion torque and the pull-out strength to investigate the effect of varying the PHS. By using FEA, the stress distribution and the interaction between the implant and the surrounding bone have

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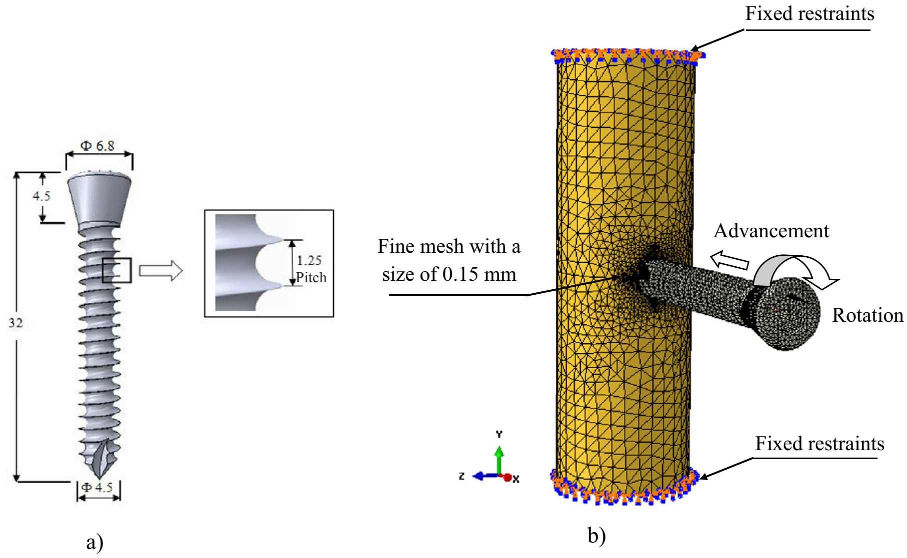


**Fig. 1.** (a) Geometries and dimensions (in millimeters) of the self-tapping screw; (b) boundary conditions with the restraint conditions and applied loads.

to be considered in order to understand and quantify the screw–bone fixation performance of fractured long bones. The goal of this numerical study is to evaluate the mechanical response of the self-tapping screw–bone interface of the fractured bone, for different PHS after the insertion process and under the immediate BWP. The stress distribution, the biomechanical load transfer to the bone and the micromovement at the screw–bone interface which are considered as important factors to the success of the internal fixation [10], were evaluated to enable targeted improvements towards a robust fixation.

## 2. Materials and methods

### *2.1. Finite element modelling*

Three dimensional (3D) models were carried out using Abaqus/Explicit 6.10 (Simulia, Providence, RI, USA), Finite Element Analysis Software [11]. The diaphysis of the human tibia was modelled as a simplified cylinder bone with an outer diameter of 16 mm [1]. It was composed of an outer cortical layer with a thickness of 5 mm and an inner trabecular core [9]. A standard 4.5 mm cortical screw with a core diameter of 3 mm was modelled in conformity with the ASTM F543-02 standard specifications for the metallic medical screws. It was a bicortical locking self-tapping screw with three cutting flutes (Fig. 1(a)).The locked head of the screw was simplified by omitting the threads. The FEA was carried out in two parts. Firstly, a comparative study is conducted to determine how the PHS (from 71% up to 85% of SED) will affect the residual bone stress, the insertion torque and the bone deformation at the screw–bone interface, during the insertion process of a self-tapping screw in the bone. The intention of this part was: (i) to isolate the influence of altering the PHS on the response of the screw–bone interface just after the insertion process, (ii) to reduce the high computational cost. Therefore, only one screw was inserted into a simplified bone model. Secondly, two selected PHS were compared using a fractured bone models fixed with screws and plate.

In the first part of the study, a predrilled pilot hole was located in the middle of the bone model and the screw was initially placed above the bone with its centerline collinear with that of the hole (Fig. 1(b)). The upper and bottom surfaces of the bone were fixed, then the screw was driven with a combined movement of an incremental pitch displacement per one revolution with an insertion depth of 8 pitches into the pilot hole (Fig. 1(b)). The model of the bone was partitioned to make possible the use of different mesh densities with a finer mesh closer to the screw–bone interface. The bone was meshed with 236,907 tetrahedral elements (C3D4), and a fine mesh was created at the pilot hole interface (Fig. 1(b)). The mesh sensitivity study was performed and the convergence criterion was set to be less than 2% difference of the highest von-Mises stress within the bone between the element sizes. For the second part, a bone with transverse fracture (gap of 1 mm) in the middle was considered and 4 pilot holes were created (Fig. 2(a)). A simply-shaped plate (with four holes) was used to fix the two parts of the fractured bone (Fig. 2(a)). Four screws were inserted with a depth of 22 pitches to reach the plate in a first modelling step, then the BWP was applied uniformly on the upper surface of the bone in a second step (Fig. 2(b)). According to Wehner et al. [12], the highest values of the internal loads along the tibial axis varied up to 1.5 BW for a healthy person. Haak et al. [13] stated that immediate weight-bearing can be allowed following the locking plate osteosynthesis of the tibial fracture. Therefore, a compressive load of 75 kg body weight person while standing up was progressively applied on the top of the fractured bone models, taking into consideration the residual stress and the bone deformation after the insertion process. Since the screws and plate are not the primary concern of this study, they were considered as rigid bodies. They were modelled by software Solid-Works 2010 (SolidWorks Corp. Dassault Systems, USA) then, the models were transferred in Abaqus. The linear triangular elements R3D3 were used to mesh the plates and the screws.

### *2.2. Materials properties and contact modelling*

Bones have anisotropic material properties but for the simplicity of the model, they were assumed to be isotropic.The elastic–plastic behaviours of the cortical and cancellous bones were considered for modelling [14–15]. Fig. 3 shows the stress–strain relationship that was reconstructed and used in this study. Bones exhibit a quasi-brittle behaviour at the low strain rate [16,17]. During the insertion process of the self-tapping screw, the bone is expected to exceed a specific yield point. Then, the ductility and toughness of the bone play a significant role in preventing the brittle failure as a result of the localized strain. The post-

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