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## **MODELING AND SIMULATION OF A HIP PROSTHESIS IMPLANTATION**

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### **ABSTRACT**

Hip replacement implants are currently increasing worldwide. However, the number of failing implants is also relatively high. In this paper, the design of a prosthesis hip joint is analyzed using finite element analysis. A FE model was developed in accordance with the ISO standard and was verified using ANSYS and SIEMENS NX packages. Furthermore, FE results were verified using published experimental results. The verified FE model was then used to simulate the joint under dynamic loading taking into account the effect of level of activity on scaling the loads up. Two material types: TI-6AL-4V and Cobalt–chromium alloys were used and the results showed that TI-6AL-4V alloy proved to have more durability under dynamic loading. In addition, the results showed that the level and type of life activates have an obvious and a pronounced effect on the service life of the prosthesis joint.

### **KEYWORDS**

Hip implant; hip prosthesis; finite element.

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## **INTRODUCTION**

Hip joint pain and/or fractures affect millions of men and women of all ages worldwide [1]. Arthritis is the main cause, it is estimated that between 1 to 6 per 1000 of men and 3 to 12 per 100 women are affected [2].

Hip replacement has become a safe and effective procedure to restore the normal movement of the hip joint. Thus, enabling the patient to continue his/her normal life activities.

The first hip replacement surgery was recorded in 1938 by Philip Wiles [3]. Poor design and under-developed materials resulted in many catastrophic failures of the joint. In 1961, John Charnley [4] revolutionized the procedure by: (1) arthroplasty, i.e. using low friction material to replace the joint, (2) using acrylic cement to fix the replacement joint in the femur bone and (3) using high-density polyethylene as a joint bearing material. These three improvements continue to form the basis of hip replacement surgery up-to-date. Currently, due to the increase in the number of elderly people in addition to increase in life span, hip replacement has become a commonly practiced surgery. It is estimated that in the US, 193,000 replacements are done annually and around one million worldwide [5]. However, despite the development in hip prosthesis design, material and production, there are in average between five and ten percent revision rates at ten years [6]. Moreover, it has been reported that only 60% of hip replacements survive for 15 years among patients who are 55 years or younger [7].

Finite element analysis has been used as a tool to develop more reliable hip prosthesis [8]. It enables designers to simulate the prosthesis hip under the variable loads under which the hip is acted due to the normal human activities.

## **THE HIP JOINT**

The hip joint is a ball and socket joint that connects the femur and the pelvis bones. The socket part (acetabulum) lies in the pelvis and the ball part is the head of the femur bone as shown in Fig.1 [9]. In healthy subjects, the bony contact surfaces are covered with a smooth and soft tissue (articular cartilage) that cushions the ends of the pelvis and femur bones and almost eliminate any friction between them. The ball and joint are stabilized by bands of tissue (ligaments) as shown in Fig. 2 [10]. It shows how the joint is strong and stable due to the reaction forces exerted by the healthy and strong ligaments.

The normal function of the hip joint is impeded by arthritis. Arthritis destroys the articular cartilage; hence, the two bones rub against each other causing pain and discomfort as shown in Fig. 3 [9]. Hip replacement surgery replaces the degenerated joint by an artificial implant as shown in Fig. 4 [9].

## **FORCE ANALYSIS**

In this section, the forces acting on the hip are derived. During slow walking, it can be fairly assumed that the weight of the whole body is instantaneously supported by

one leg. The forces acting on one leg are drawn as shown in Fig. 5. The weight of the body  $W$  is balanced by the reaction from the ground  $N$ , where  $W=N$ . in order for the person not to fall over, the two forces act on the same line.

Now, the forces on the right leg can be isolated as shown in Fig. 6, which shows the acting and reaction forces on an anatomical diagram as shown in Fig. 6 [11]. The weight of one-leg  $W_{leg}$  acts at a point near the knee, a reaction force  $R$  acts on the head of the femur by the acetabulum and the abductor muscles exert a pulling force  $M$  on the hip. The dimensions between the forces are the results of the extensive work by Inman in 1947 to measure the forces exerted by the abductor muscles [12].

The weight of one leg can be assumed as  $0.16W_b$  according to the data published by Miller and Nelson about the relative masses of body parts [13]. The forces in the x-y directions are analyzed as shown in Fig. 7. Then, the equilibrium equations are developed; for forces (Equations 1 and 2) and moment (Equations 3) about the z-axis at the point where force  $R$  acts.

$$\sum F_x = M * \cos(70^\circ) - R_x = 0 \quad (1)$$

$$\sum F_y = W_b - 0.16 * W_b - M * \sin(70^\circ) - R_y = 0 \quad (2)$$

$$\sum T_z = 10.8 * W_b - 3.2 * 0.16W_b - 7 * M\sin(70^\circ) = 0 \quad (3)$$

Solving the equations; it is found that  $M = 1.57W_b$ ,  $R_x = 0.54 * W_b$  and  $R_y = 2.31 * W_b$ . The magnitude of  $R = \sqrt{R_x^2 + R_y^2} = 2.37 * W_b$ . This means that the reaction force acting on one hip joint is roughly 2.4 times the total weight of the whole body. The abductor muscles play an important role in countering the loads on the pelvis.  $M = 1.57 * W_b$ . This shows the importance of having strong muscles to help in resisting the loads on the pelvis.

For an average person who weighs 70 kg, the force acting on the hip is ~1600 N. However, for a typical soldier carrying his mandatory gear with total weight as 100 kg, the reaction force on the hip is approximately 2300 N. This shows the extent of the stress under which the hip joint is undertaking. The force is even exacerbated by the activity type performed by a person. The maximum load that has been measured in-vivo is during stair ascending/descending is 7.2–7.4 times the body weight [14].

## EXPERIMENTAL TESTING

The ASTM F2996 - 13 standard [16] describes the experimental test method for a prosthesis hip joint. The test procedure is also described in full details including the test parameters and the requirements in the 7206-4:2010 ISO standard [17].

As shown in Fig. 8, the stem of the hip is fixed in a casing using special cement. The load is applied to the head of the hip joint at an angle. Strain gauges are glued

to the level at which the stem is cemented. The values of strain are then used to calculate principal stresses.

## **FINITE ELEMENT**

A CAD model of a prosthesis hip joint was obtained from the ASTM website [18]. The model represents the hip joint as surfaces; therefore, it required a processing to transform it into a solid part. This was done using ANSYS SpaceClaim software.

As shown in Fig. 9, the part was fixed from the bottom until a location 90mm distance from the top most part. A force of 2300 N was applied to the top surface of the hip joint. A material model was created according to the guidelines in [16] with following properties: Young's modulus is 113.7 GPa and Poisson's ratio is 0.3. These properties correspond to the Titanium Ti-6Al-4V alloy.

## **MODEL VERIFICATION**

For verification, the model was solved using two FE packages: ANSYS and Siemens NX Nastran. The mesh size was the same for both, 5 mm. In NX, the CTETRA element was used, as it is preferable for modeling parts with turns and sharp change in geometry. It was used with ten grid points for better accuracy. In ANSYS, the default SOLID187 element was used; it has similar capabilities to the CTETRA element. The two elements are shown in Fig. 10.

The results are shown in Fig. 11, which shows the maximum principal stress distribution. It is clear that the results are in very good agreement; the difference is seven MPa.

The results from the FE analysis are compared with ones from experimental testing which have been reported in the ASTM F2996 - 13 standard [16]. In the standard, the hip prosthesis joint was tested by five different commercial labs and two different universities. The average value of the maximum principal stress was 24.1 ksi (166.16 MPa) and the average standard deviation between all seven measured values was 0.8 ksi.

For clarification, the results are presented in Table 1. It is clear that both FE packages produced accurate results, where ANSYS produced better ones. The maximum error was 3%.

## **FATIGUE ANALYSIS**

### **Cyclic Loading**

In this section, the response of the hip replacement joint under fatigue loading is examined. Two typical types of materials are used; Ti-6Al-4V Titanium alloys and Cobalt-chromium alloy. Their mechanical properties are given in Table 2. The S-N curves for the two materials are given in Fig. 12.

**Table 1.** Maximum principal stress at the fixation level from FE simulations and experimental testing.

	Experimental	ANSYS	SIEMENS NX
Maximum principal stress, (MPa)	166.16	164.08	171.15
% Error	----	1.25	3

**Table 2.** Mechanical properties of the prosthesis hip joint.

Material	Young's modulus (GPa)	Poisson ratio	Yield strength (MPa)
Ti-6Al-4V	110	0.32	800
Cobalt-chromium alloy	220	0.30	720

ANSYS was used to simulate the prosthesis hip under fatigue loading using the verified FE model. Two new material types were created in ANSYS workbench using the mechanical properties in Table 2 and Fig. 12. FE analysis was conducted under the same loading conditions described in FE section above. It was assumed that the load was cyclic and varying around the nominal load value.

The minimum safety factor was calculated according to the Goodman fatigue theory. It was calculated at different fatigue scale factors from one to eight, where eight corresponds to the load during ascending or descending stairs. The results are shown in Fig. 13. It is clear that the two materials result in similar trend where the Ti-6Al-4V alloy has better durability under fatigue loading.

## REALISTIC LOADING

In this section, actual dynamic loading during walking is used as shown in Fig. 14 [8]. The forces acting on the surface of the head of the prosthesis in the three major directions are shown for duration of five seconds, in addition to the total resultant force. It is clear that the variation in total force value is relatively small. The total force varies around an average value of 3357 N with a standard deviation of 102 N.

The results of fatigue analysis are shown in Fig. 15. The values of minimum safety factor are very close for the two materials types. This can be explained by the fact that there is small variation in the amplitude of the acting load, which resulted in limited effect on the fatigue response.

The pronounced effect of the type of loading (amplitude and frequency) on the operational service life is obvious, especially when taking into consideration the results in Fig. 13. This shows that when replacing a hip joint for a patient, his/her levels of normal life activities must be taken into account when selecting the design of the prosthesis joint and its type of material. This is to ensure that minimum failure rates should occur during normal life activities and that no material wears which can cause tissue and blood poisoning from the absorption of worn out material particles.

## CONCLUSIONS

In this paper, the dynamic behavior of the prosthesis hip joint was thoroughly analyzed using FE analysis. All steps of the analysis were conducted in accordance with ISO and ASTM standards. The FE model was properly verified using ANSYS and SIEMSN NX packages in addition to experimental test results. The effect of dynamic loading was examined using fatigue analysis utilizing two alloys: TI-i-6AL-4V titanium and Cobalt–chromium alloys.

Based on the results, it can be concluded that in order to ensure the safe and reliable operation of the prosthesis joint, fatigue loading must be taken into consideration when selecting the most suitable joint. The results showed that fatigue loading is highly dependent on the type and level of life activities of the patient, which is highly variable from person to person.

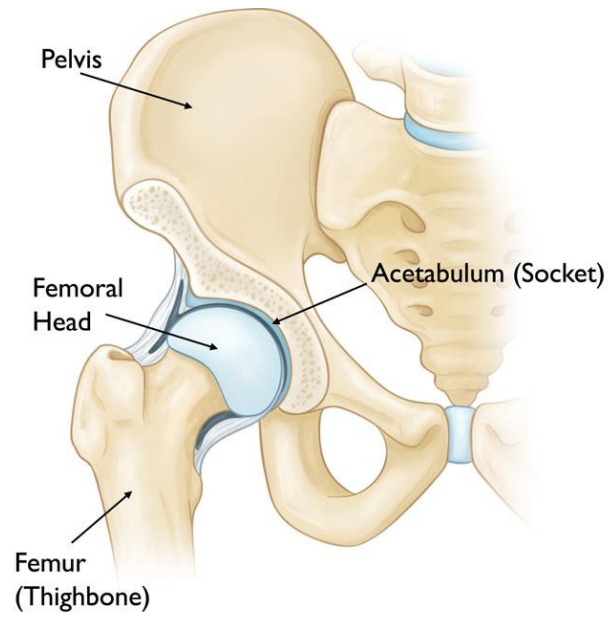
The type of material proved to have a significant role on the durability of the prosthesis joint under normal loading. It is there recommended to consider the material response under cyclic loading when selecting the type of material. This is essential since sudden material failure has a catastrophic effect on the patient as well as economical loss to extensive and re-replacement surgery costs. Moreover, if wear occurs and continues for a prolonged period, tissue and/or blood poisoning can occur causing more harm to patients.

Finally, this work showed that proper FE analysis and application will aide in the design improvement of the prosthesis hip joints. In addition to saving costs and time as conducting human lab experiments is highly expensive and time consuming with ethical complications.

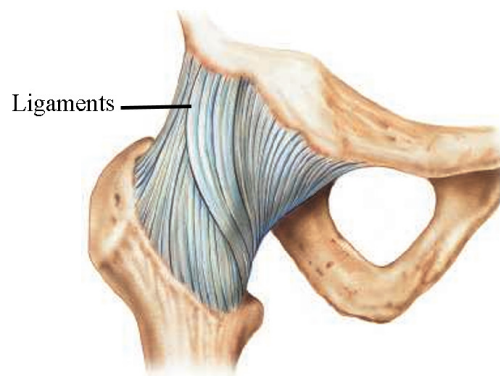
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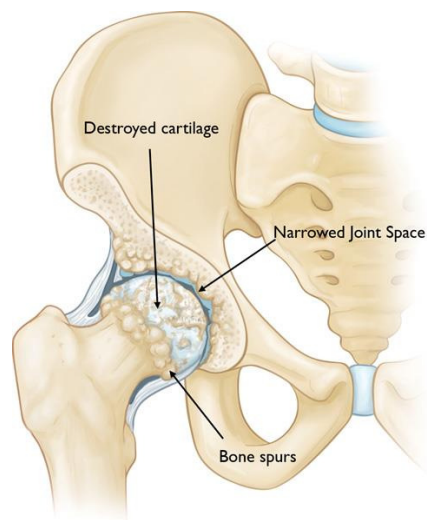
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**Fig. 1** Anatomy of a normal hip joint [9].

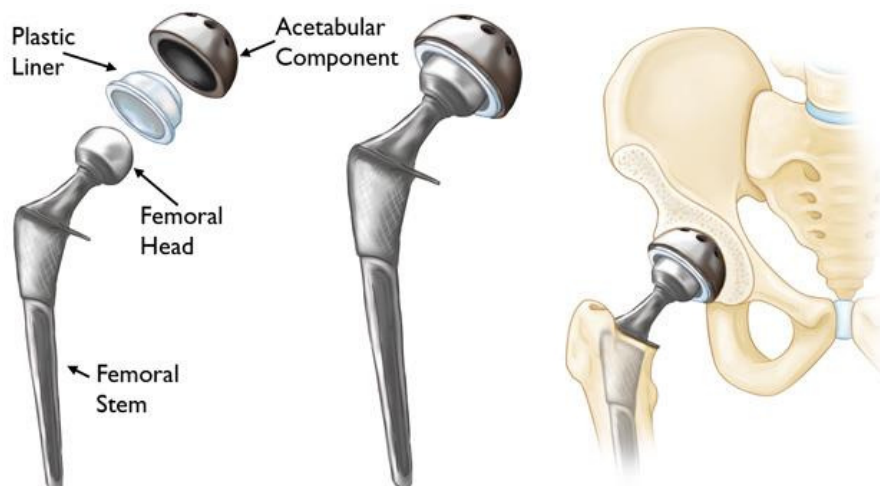


**Fig. 2** Anterior view of the right hip joint [10].



**Fig. 3** Wear in the joint surface [9].

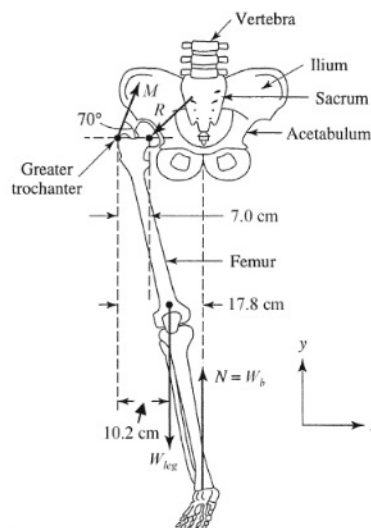




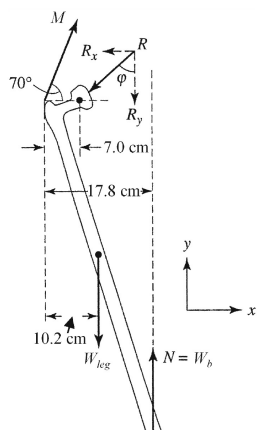
**Fig. 4.** (Left) The individual components of a total hip replacement. (Center) The components merged into an implant. (Right) The implant as it fits into the hip [9].



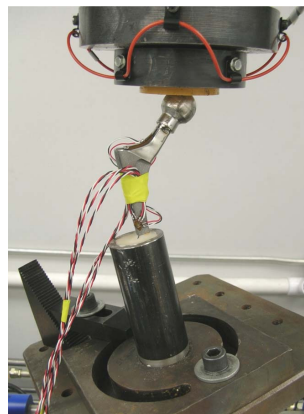
**Fig. 5.** A person standing on one foot.



**Fig. 6.** Anatomical free body diagram of a person standing on one leg [11].



**Fig. 7.** Force diagram of one leg [15].



**Fig. 8.** Experimental loading conditions [16].

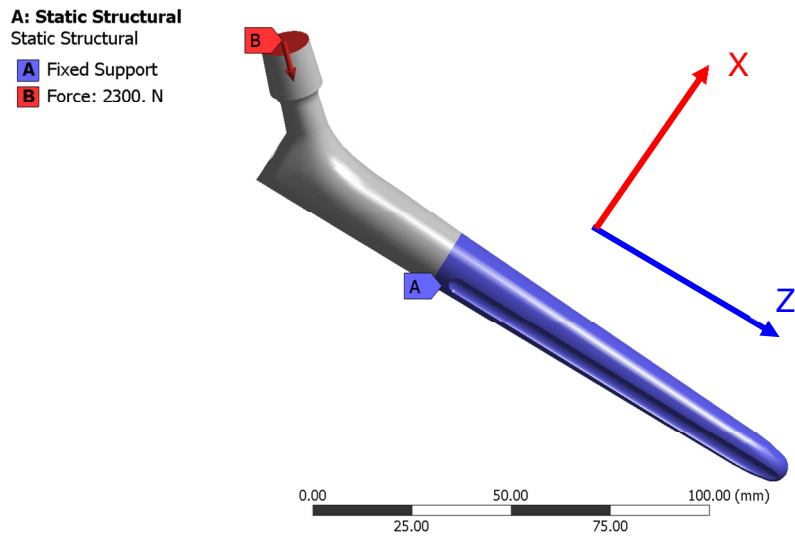


Fig. 9. FE model.

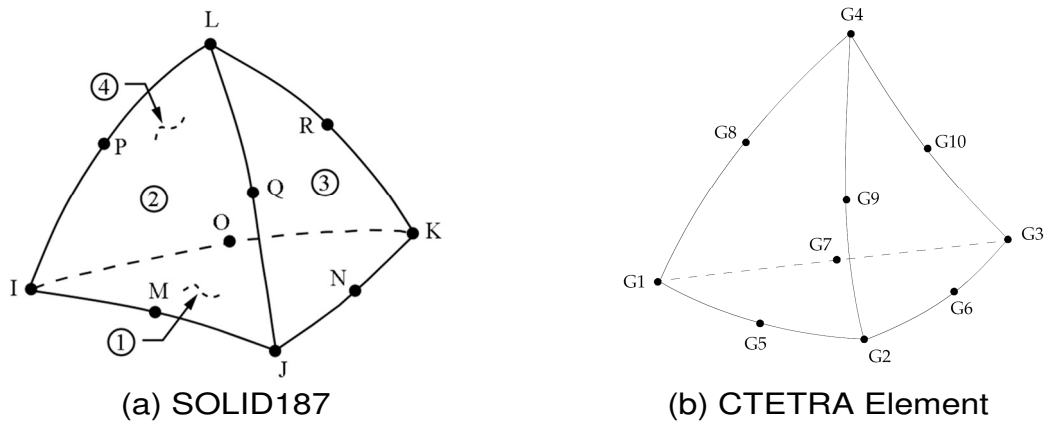


Fig. 10. ANSYS and NX elements.

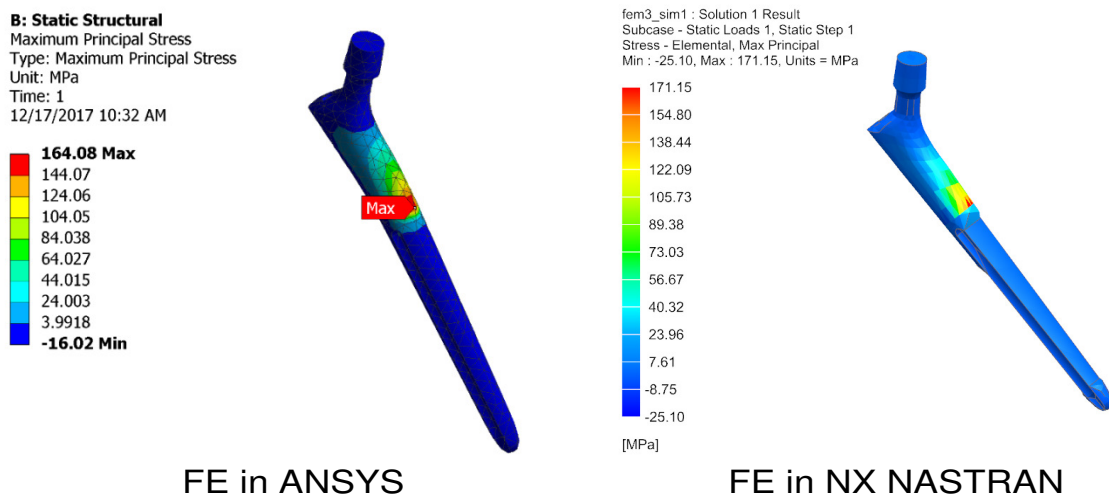
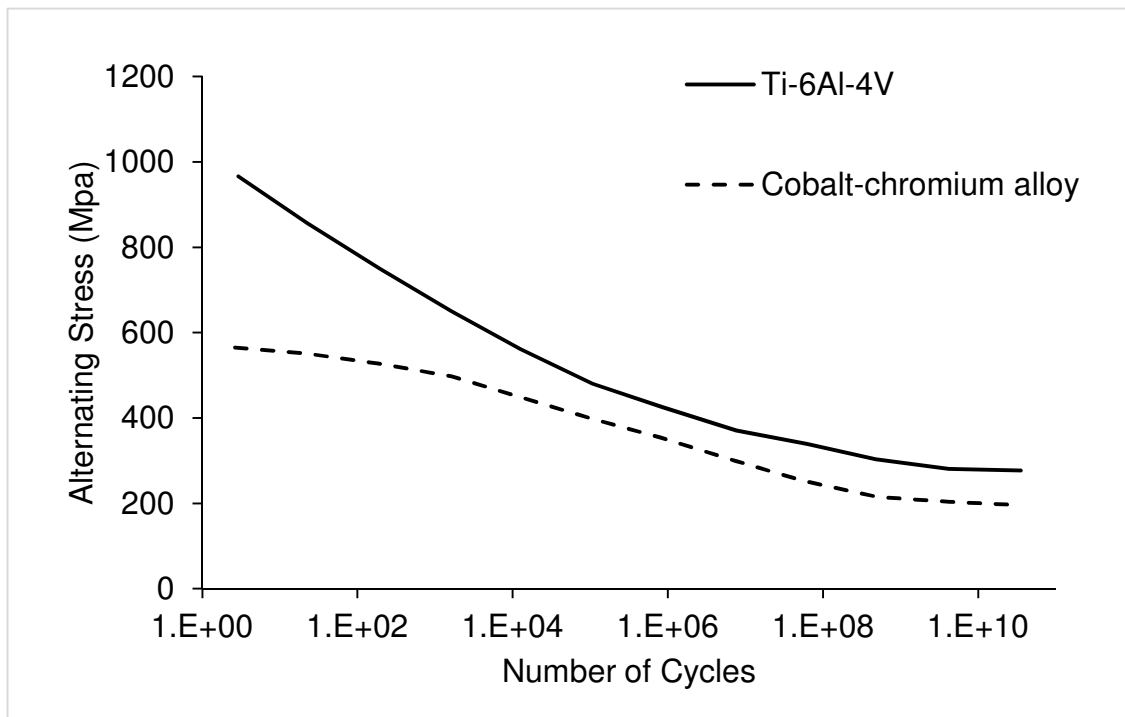
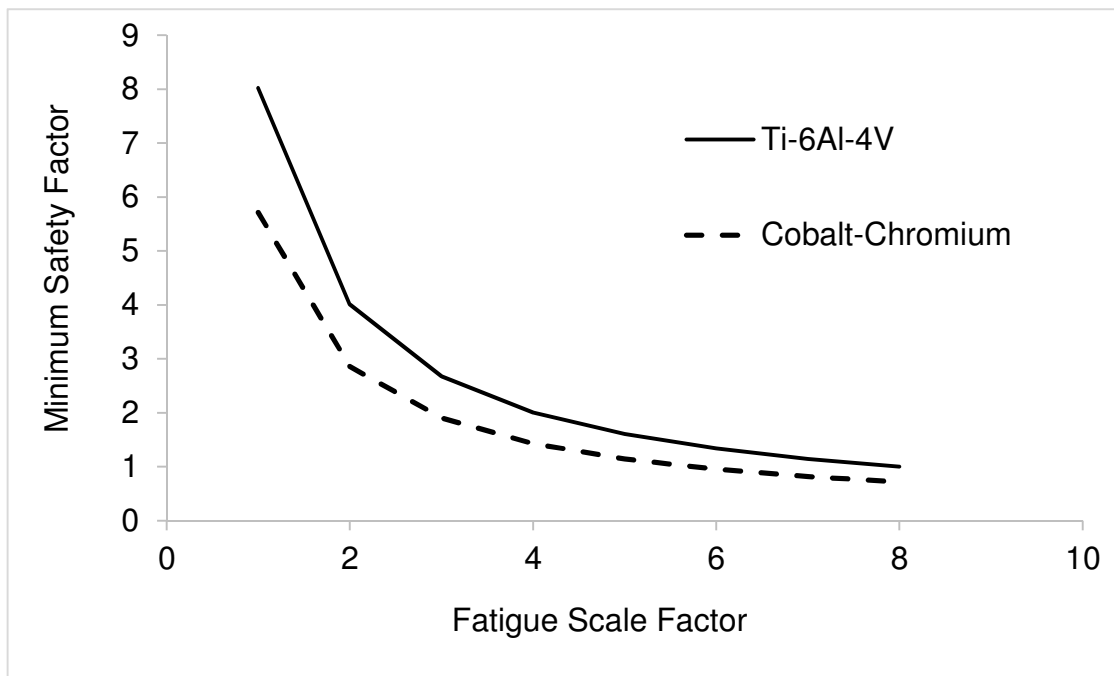


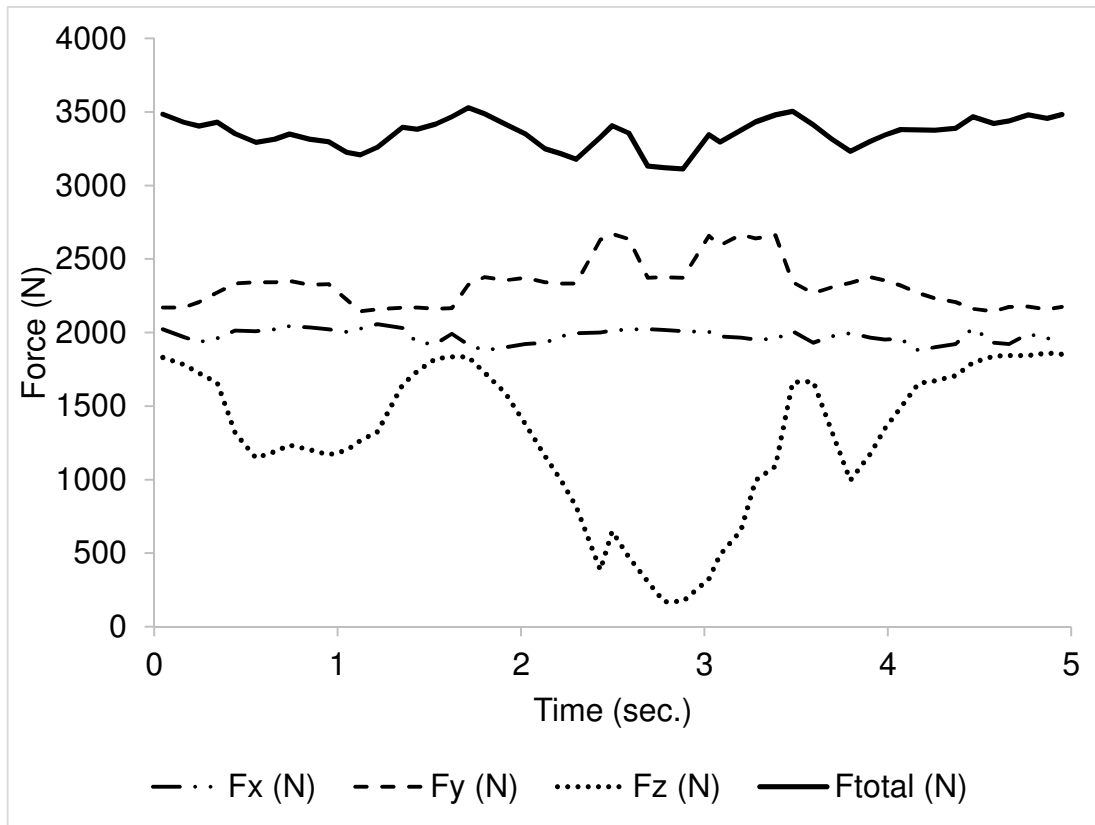
Fig. 11. Model verification.



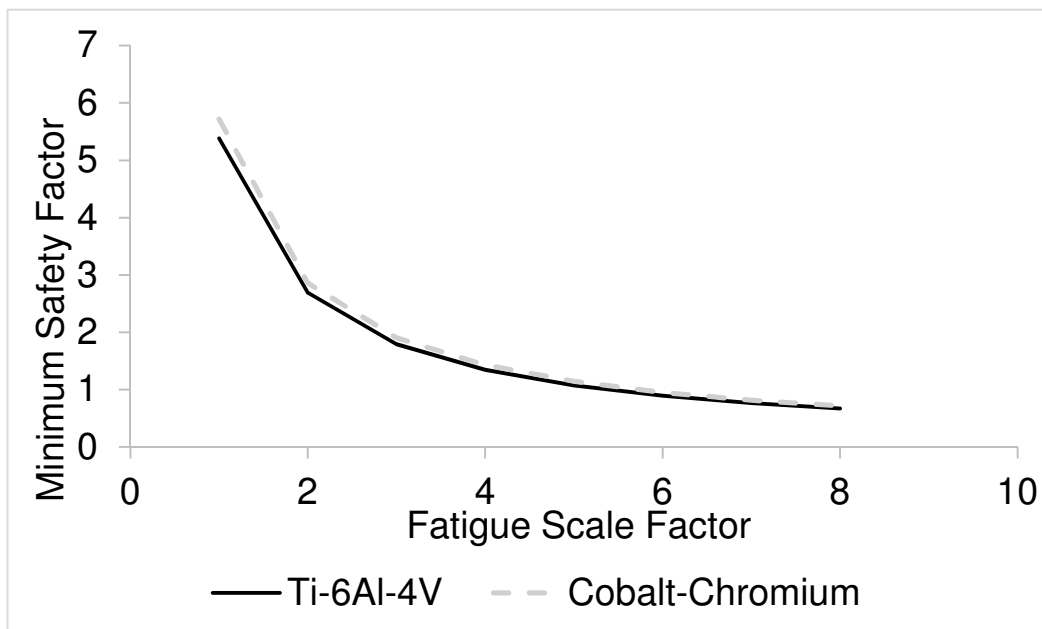
**Fig. 12.** S–N curve for the two materials.



**Fig. 13.** Variation of minimum safety factor with fatigue scale factor for the two material types for cyclic loading.



**Fig. 14.** Dynamic load on the prosthesis hip joint during walking [8].



**Fig. 15.** Variation of minimum safety factor with fatigue scale factor for the two material types for realistic loading.