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Vibration Analysis of Prosthesis for the through knee Amputation

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Abstract

Prosthesis is an artificial extension that replaces a missing body part, lost by injury or missing from birth, or supplements, and a defective body part. Through-knee amputation should be considered as an alternative to trans femoral amputation. This work involves an experimental part to measure the Ground Reaction Force (GRF), and pressure distribution for both old and new prostheses. The interface pressure (IP) between leg and socket was measured by using F-Socket sensor. The natural frequency was measured using impact hammer test, while the vibration data (acceleration and frequency) are measured at the last stage for the patient at different positions during the gait cycle. The results show that the GRF for the designed prosthesis Gait cycle time in the left leg is equal to (1.50sec) and right leg is equal to(1.58sec). The maximum interface pressure (IP) is recorded in the Tensor fascia lata regions in the thigh part with (408KPa). Results show that the maximum vibration data was found at the thigh region with (2.09 $\frac{m}{s^2}$ and 2.66 Hz) for acceleration and frequency respectively. At thigh region for the old prosthesis and the maximum Vibration data was found at the thigh with (4.44 $\frac{m}{s^2}$ and 3.12 Hz) for acceleration and frequency respectively at thigh region for the designed prosthesis for the same person. From the hammer test, the natural frequency was found that for the first mode of the old prosthesis 33.2 Hz and for the designed prosthesis was 46.8 Hz.

Keywords: Socket, Gait cycle time, Interface Pressure (IP), Natural Frequency.

Introduction

The most common cause of amputation in war-torn countries of the world such as Iraq, Cambodia, Iran, and Afghanistan, 80 to 85percent of amputees are land mine survivors. These mines are responsible for 26000 amputations per year and have produced 300,000 amputees worldwide [1]. In the world, one million people have been killed or maimed by landmines since 1975 and there are approximately 26 thousand new victims every year with approximately 90% of all amputees being lower limb amputees [2].

In 1946, a major advancement was made in the attachment of lower limbs. A suction sock for the above-knee prosthesis was created at University of California (UC) at Berkeley. In 1975, Ysidro M. Martinez' invention of a below-the-knee prosthesis avoided some of the problems associated with conventional artificial limbs.

Martinez, an amputee himself, took a theoretical approach in his design [3]. The main research problem in the field of developing prostheses is how to make the best use of modern technology and still keep the production price accessible to the people in need [4].

Through-the-knee amputation (TKA) is an excellent lower extremity treatment for the ischemic extremity when revascularization is not feasible and a prosthesis is not practical. Over the past 8 years 185 major amputations have been performed at their hospital of which 63 were of the TKA type. as shown in figure1[5]. The through knee is the most distal amputation in which normal knee function is completely lost by the amputation level. However the physical differences may be of considerable importance in the prosthesis procedure. The end of the stump is composed of tissue normally adapted to weight bearing in the kneeling amputated patient. In biomechanical terms the problems are very similar to those of the above knee position [6].

Figure 1: Through-Knee prosthesis [5].

1. Gait cycle

Gait analysis is important when developing a power and typically broken down into two phases; the stance phase (60%) and the swing phase (40%).The gait cycle for the right side begins with heel strike of the right foot. At this point, both feet are on the ground. This is known as the initial double support phase

This sub-phase of the gait cycle is also known as weight acceptance as the body weight is shifted to one leg [7].The gait cycle can be described in terms of percentage, rather than time, thus allowing normalization of the data for multiple subjects. The initial foot strike occurs at 0% and occurs again at 100 % (0-100%). The stance phase of normal person usually lasts about 62% of the cycle, the swing phase about 38% [8].

2. The Effects of Vibration on the Human Body

Vibrations influence the human body in many different ways. The response to a vibration exposure primarily depends on the frequency, amplitude, and duration of exposure. Other factors may include the direction of vibration input, location and mass of different body segments, level of fatigue and the presence of external support. The human response to vibration can be both mechanical and psychological.

Mechanical damage to human tissue can occur, which are caused by resonance within various organ systems. Psychological stress reactions also occur from vibrations; however, they are not necessarily frequently related. From an exposure point of view, the low frequency range of vibration is the most interesting. Exposure to vertical vibrations in the 5-10 Hz range generally causes resonance in the thoracic-abdominal system, at 20-30 Hz in the head-neck-shoulder system, and at 60-90 Hz in the eyeball. When vibrations are attenuated in the body, their energy is absorbed by the tissue and organs. The muscles are important in this respect. Vibration are leads to both voluntary and involuntary contractions of muscles, and can cause local muscle fatigue particularly when the vibration is at the resonant-frequency level [9].

3. Experimental Programme.

3.1 Materials of socket of through knee prosthesis

In this work the materials needed in the lamination of the through knee socket

- Perlon stockinet white (ottobock health care 623T3).
- Fiber carbon stockinet (ottobock health).
- Lamination resin 80:20 polyurethane (proter hand icap technology).
- Hardening powder (ottobock health care 617P37).
- Polyvinyalcohol (PVA) bag (ottobock health care 99B71).
- Materials for Jepson mold.

3.2 Manufacturing for a prosthesis socket

The positive molds mounted the laminating stand, complete the connection with the vacuum forming system through the pressure tubes, pull the inner PVA bag in the positive mold, and open the pressure valves to a value of approximately 30 mm Hg at room temperature. The perlon stockinet is put and fiber carbon stockinet according to the laminating layup(4Perlon +2fiber carbon +4perlon) layers and pull the outer PVA and keeping the smaller end positioned over the valve area by using a cotton string to tie off the PVA bag. Mix The lamination with the hardener as shown in figure 2 , then the resulting matrix mixture will be inside and outside PVA bag, and distribute the matrix homogeneously over all area of lamination stockinet. Maintain constant vacuum until the composite materials becomes cold and then lift the resulting lamination as shown in figure 3.

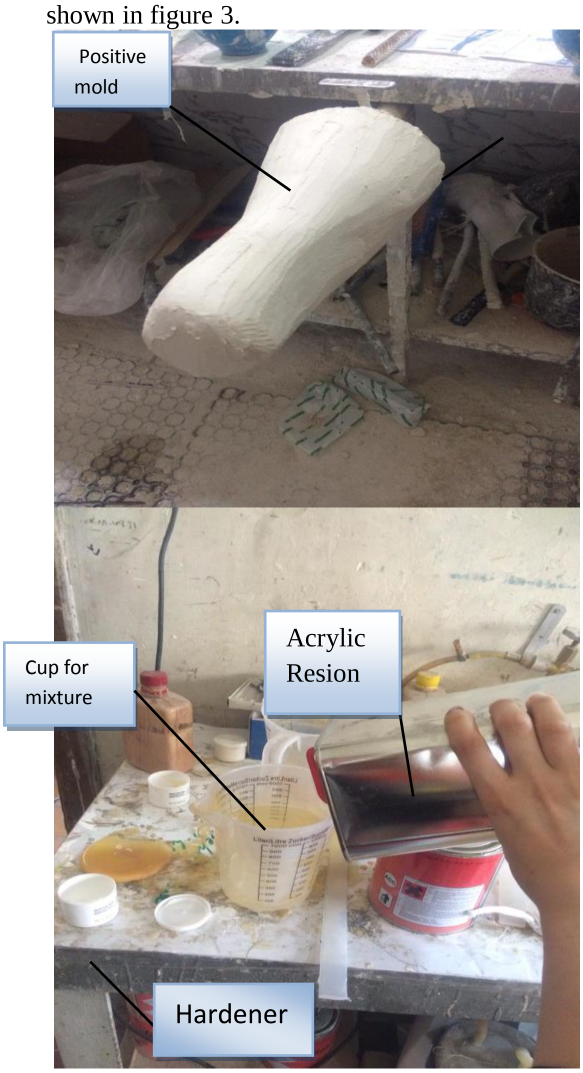

Figure 2: Positive mold and mixed the lamination with the hardener used in through knee socket.

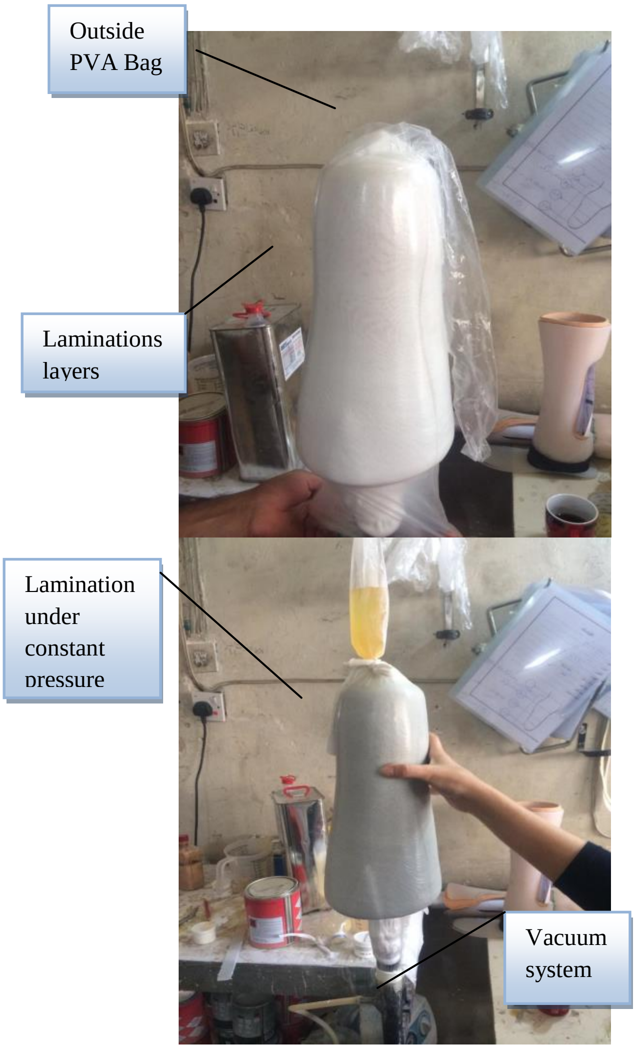

Figure 3: Positive mold before and after lamination.

3.3 Experimental Procedure

In order to study the mechanical properties of the manufactured socket, tensile and fatigue tests were carried out. This is for measuring the socket strength and its reliability before running the gait cycle analysis. Figures 4 and 5 illustrate the experimental types of specimens for examining the manufactured socket mechanical properties. The tensile specimens were machined at the (Iptysam center for prosthetic and orthotic workshop). Three samples for the designed material were machined according to ASTM D638 [10] with 50 mm original length and 13 mm the width while thickness varied with the type of the layup. Figure 4 shows the shape of tensile specimens. The tensile test was done by using the tensile test machine at University of AL- Nahrain that is used to evaluate modulus of elasticity, yield stress and ultimate stress for design the socket prosthesis. The test was carried out at a speed of 5.000 mm /min. The specimens after test are shown in fig. 5.

Figure 4: The general shape of tensile specimen of designed material.

Figure 5: Tensile test specimens of designed material after test.

They were manufactured in 10 mm width and 100 mm length. The fatigue performance of a material is determined by testing a number of similar test specimens at different levels of maximum stress. The fatigue test of material specimens was carried out at (University of AL-Nahrain) .The group includes eight lamination specimens. The specimens before and after test are shown in figs 6 and 7 respectively.

Figure 6: Fatigue specimens before testing

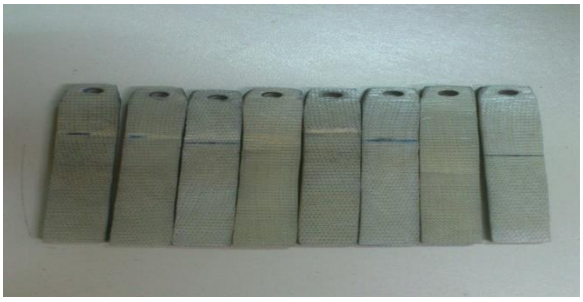

Figure 7: Tested Fatigue specimens.

4. Results and Discussion.

4. 1 Gait Cycle Parameters:

This test was done in AL-Nahrain University at the prostheses and outhouses engineering on person stump through knee prosthesis in left leg with weight of 67 kg. The results of this test are compared with prosthesis' test done for the same person.

Comparison of results obtained from old and designed prostheses subject case study can be shown in figures 8 to 13 for both cases.The differences can be detected in step length for the left leg equal to 43.9 cm and the right 20.9 cm, while for the designed prosthesis case was 49.5cm and 46.1cm .The person with old prosthesis has Gait cycle time in the left leg they equals to (1.59 sec) and in the right leg equals to (1.70sec),while the designed prosthesis Gait cycle Time in the left leg is equal to (1.50sec)and right leg is equal to (1.58sec).From results it was noted that in the parameters shown in tables 1 to 6 for the old prosthesis and designed prosthesis respectively, that the behavior of the gait cycle of old prosthesis subject was different from designed prosthesis subject. This difference is due to the effect of weight of the old prosthesis on walking person stump through knee prosthesis. The weight of the socket of old prosthesis is equal to 0.7kg while the weight of the socket of designed prosthesis is equal to 0.5kg. The difference belongs to the manufacturing procedure for both cases. There is an apparent difference, in the first part (initial contact) with gradual increase in the pressure value until it reaches to maximum in old prosthesis. The old prosthesis used the heel gradually to base on ground, while designed prosthesis has sudden increase in pressure value due to distribution of pressure under the subject's heel of his shoe. By comparing figures 8 to 13 it was found that the value of the pressure in the left leg is more than that in the right leg because of the high impact of defected left lower limb with floor when he tries to keep the balance of his body during walking.

Table 1: Step table for old prosthesis**.**

Step table	Left leg	Right leg	Difference
Step time (sec)	0.77	0.70	-0.07
Step length (m)	0.439	0.209	0.23
Step velocity(m\sec)	0.569	0.298	-0.271
Steplength\leglength	1.91	0.91	-1.00
Step width (m)	0.182	0.194	0.012

Table 2 : Step table for design prosthesis.

Step table	Left leg	Right leg	Difference
Step time (sec)	0.70	0.74	0.04
Step length(m)	0.495	0.461	-0.0341
Step velocity(m\sec)	0.702	0.626	-0.076
Steplength\leglength	0.02	0.02	-0.00
Step width(m)	0.149	0.155	0.006

Table 3 : Gait cycle table for old prosthesis.

Gait cycle table(sec)	Left leg	Right leg	Difference
Gait cycle Time	1.59	1.70	0.11
Stance Time(sec)	0.92	01.09	0.17
Swing Time(sec)	0.55	0.60	0.05
Single supportTime	0.46	0.56	0.1
Initial Double time	0.30	0.23	-0.07
Terminal Double	0.23	0.30	0.07
Total Double time	0.53	0.53	0.00
Heel Contact Time	0.80	0.69	-0.11
Foot Flat Time	0.55	0.67	0.12
Mid Stance Time	0.53	0.46	-0.07
Propulsion Time	0.16	0.38	0.22
Active Propulsion	0.02	0.20	0.18
Passive Propulsion	0.32	0.18	-0.14

Table 4 : Gait cycle table for the design prosthesis.

Gait cycle table(sec)	Left leg	Right leg	Difference
Gait cycle Time	1.50	1.58	0.08
Stance Time	0.90	0.92	0.02
Swing Time	0.58	0.49	-0.09
Single support Time	0.47	0.52	0.05
Initial Double time	0.24	0.22	-0.02
Terminal Double	0.22	0.24	0.02
Total Double time	0.46	0.46	0.00
Heel Contact Time	0.65	0.58	-0.07
Foot Flat Time	0.16	0.26	0.10
Mid Stance Time	0.19	0.56	0.37
Propulsion Time	0.37	0.42	0.05
Active Propulsion	0.13	0.34	0.21
Passive Propulsion	0.24	0.09	-0.15

Table 5: Symmetry table for the old prosthesis.

Symmetry table	
Step Time (%)	0.91
Step Length (%)	47.7
Step Velocity (%)	52.4
Step Length\Leg Length (%)	47.7
Step Width (%)	93.9
Gait Cycle Time (%)	93.3
Single Support time (%)	83.2
Initial Double Support Time (%)	77.7
Terminal Double Support Time (%)	7.7
Total Double Support Time (%)	100

Table 6: Symmetry table for the design prosthesis.

Symmetry table	
Step Time (%)	95.7
Step Length (%)	93.2
Step Velocity (%)	89.2
Step Length\Leg Length (%)	93.2
Step Width (%)	96.6
Gait Cycle Time (%)	95.0
Single Support time (%)	89.6
Initial Double Support Time (%)	90.0
Terminal Double Support Time (%)	90.0
Total Double Support Time (%)	100

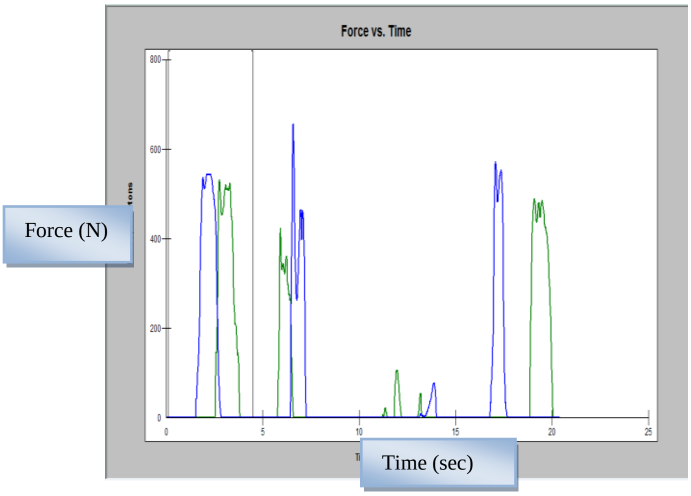

Figure.8. :The GRF for the old prosthesis

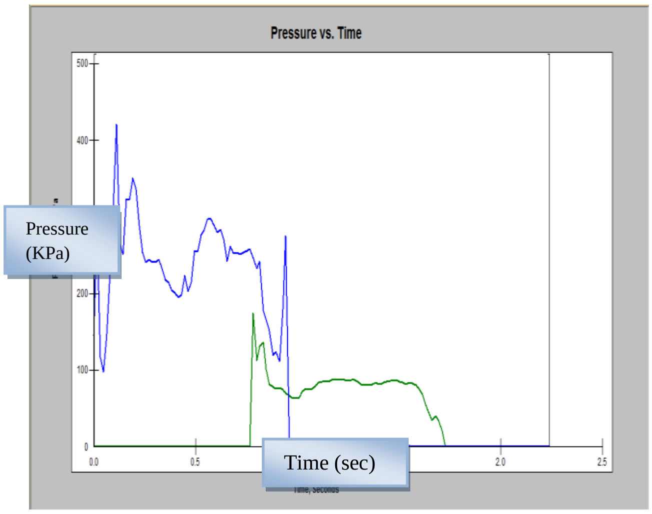

Figure.11. The pressure distribution for designed prosthesis.

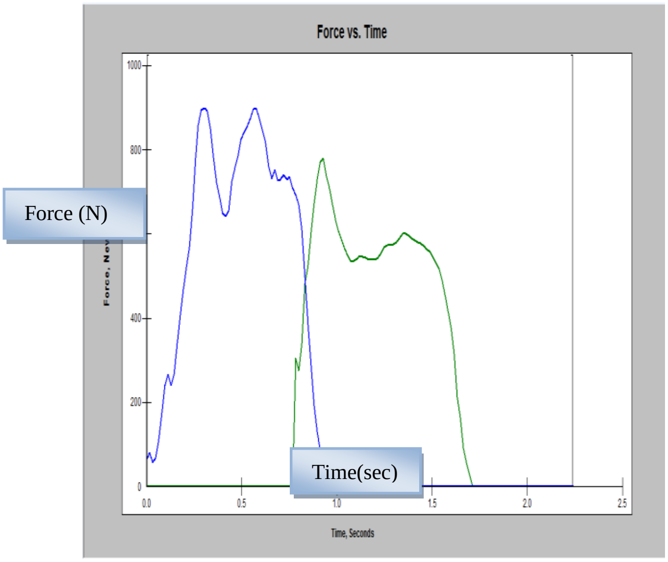

Figure 9 : The GRF for the design prosthesis.

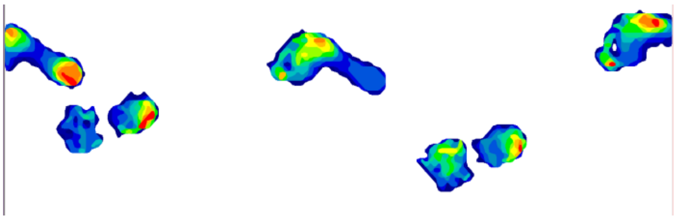

Figure 12: The Foot Print for old prosthesis.

Figure 13 : The Foot Print for design prosthesis.

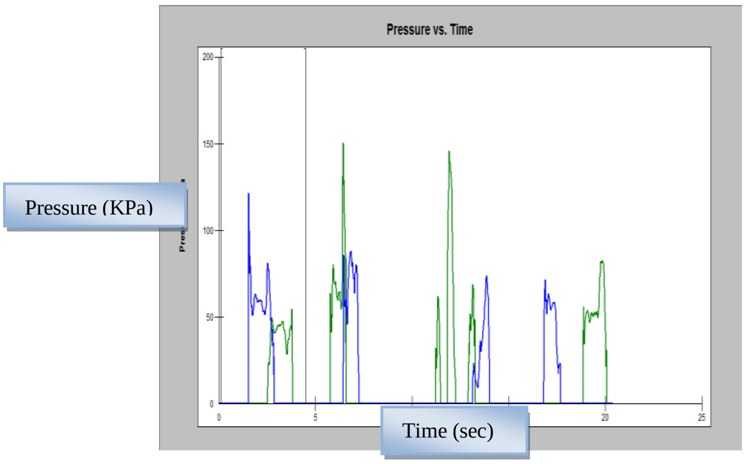

Figure 10 : The pressure distribution for old prosthesis

4.2 Interface Pressure Parameters:

The experimental part of case study design prosthesis. The maximum IP is recorded in the Tensor fascia lata muscle regions in the thigh part with (408kPa) for the design prosthesis of the same person. The reason of the behavior is that Tensor fascia lata muscles are more active from other for design prosthesis .Nevertheless flexion knee and shorted left leg lead to alignment of leg to the internal side of body during gait cycle. This making most of body weight based on the Tensor fascia lata muscle and Sartorius muscle. The results obtained for the designed prostheses are, shown in Figs 14and15.

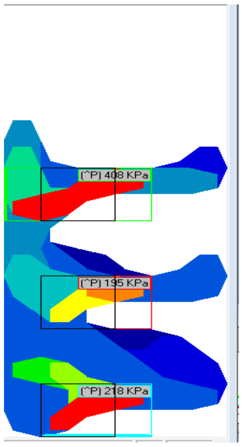

Figure 14: The regions in the thigh part pressure distribution for designed prostheses obtained using F-Socket sensor.

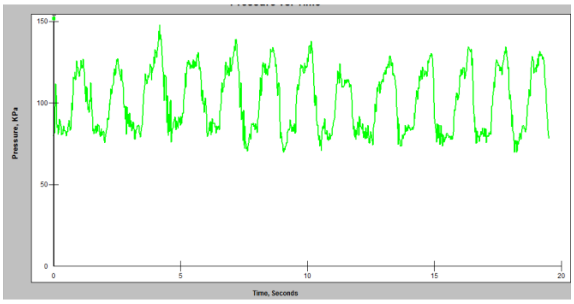

Figure 15: The interface pressure for design prostheses.

4.3 Acceleration and Frequency Measurement:

The values of acceleration are taken from RMS acceleration .figs16 to 20 and tables 7 to 8 show frequency and acceleration for both two cases studies. When comparing the results it has been found that the value of acceleration increases from thigh region both for two cases studies. Also it is found that the values of acceleration in design prosthesis are more than those of old prosthesis due to the person stump wearing design prosthesis with weight of 3.5kg while the weight 4.6kg that his makes have more effective muscles because of his defected left leg which is flexion knee, while it is noted that the value of acceleration is less in old prosthesis because of that effect of the prosthesis weight and make weak muscles. It is found that there are increase in the value of acceleration and frequency in thigh. This means that there is no muscle working as a damper to reduce acceleration and frequency at this region.

Table 7 : Vibration data for old prosthesis.

Vibration data for the old prosthesis		
Point	RMS Acceleration amplitude $(^{m}/_{s^2})$	Frequency(Hz)
ankle	1.82	1.71
thigh	2.09	2.66
knee	2.00	1.90

Table 8 : Vibration data for design prosthesis.

Vibration data for new prosthesis		
Point	RMS Acceleration amplitude $(^{m}/_{s^2})$	Frequency(Hz)
Ankle	2.20	2.08
thigh	4.44	3.12
knee	2.96	2.48

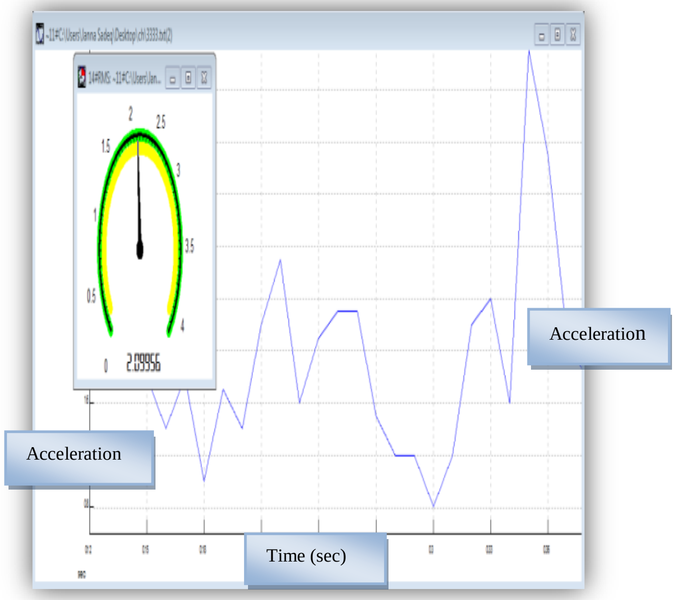

Figure 16: Absolute acceleration at thigh for old prostheses.

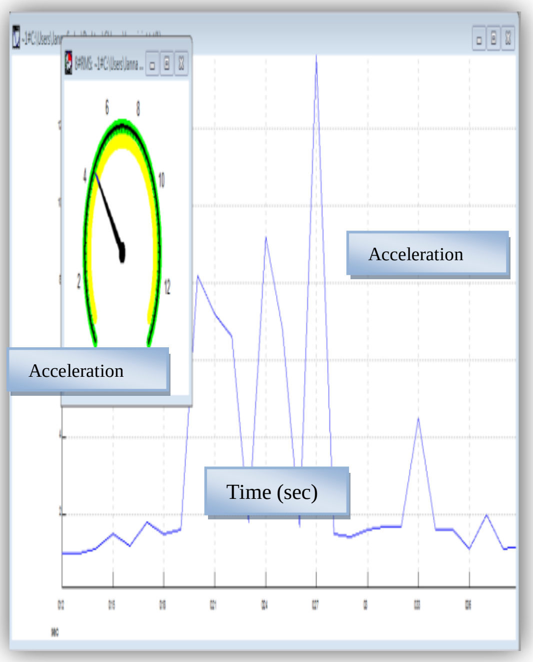

Figure 17: Absolute acceleration at thigh for design prostheses

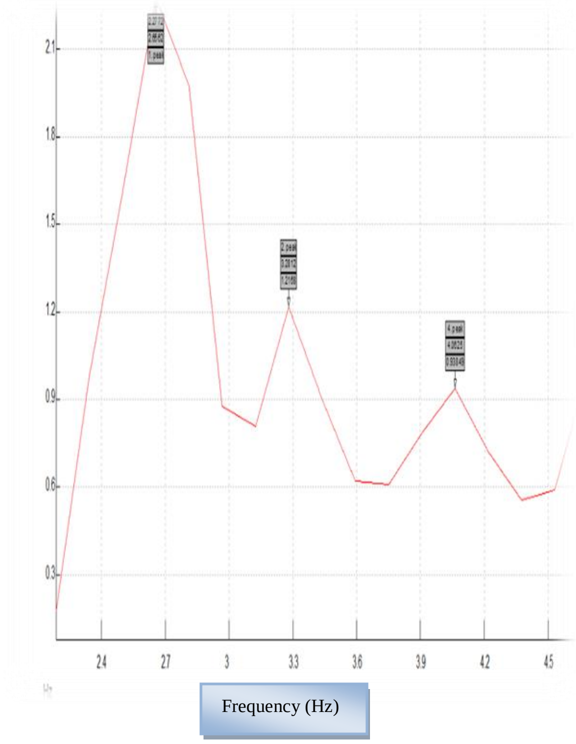

Figure 18: Absolute acceleration with frequency at the thigh for old prostheses

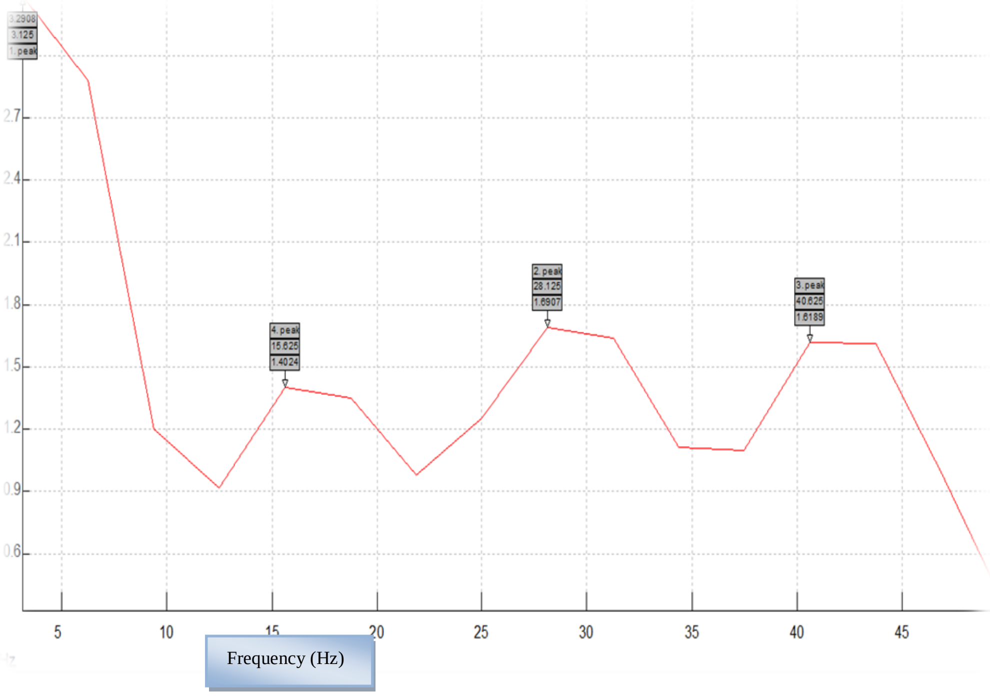

Figure 19: Absolute acceleration with frequency at the thigh for design prostheses.

Figure 20 : Sensor of accelerometer for man with Through knee

4.4 Natural frequency:

Natural frequencies refer to the frequencies that the structure tends to naturally vibrate and the mode shapes refer to what shape the structure would tend to vibrates at each frequency benefits of modal analysis that is to allow the design to avoid the resonant vibrations or to vibrate at a specified frequency. From the hammer test the natural frequency was found for the first mode of old prosthesis is equal 33.2 Hz,while the first mode of design prosthesis is equal 46.8 Hz as shown in Figures 21to25 .This result means that the patient during wearing prosthesis must avoid or stay away from source of vibration with frequencies approaching to natural frequencies values so that resonant vibrations does not take place. The boundary conditions were set according to the real case study, and the measurements were taken while the patient was standing as shown in Fig. 25.

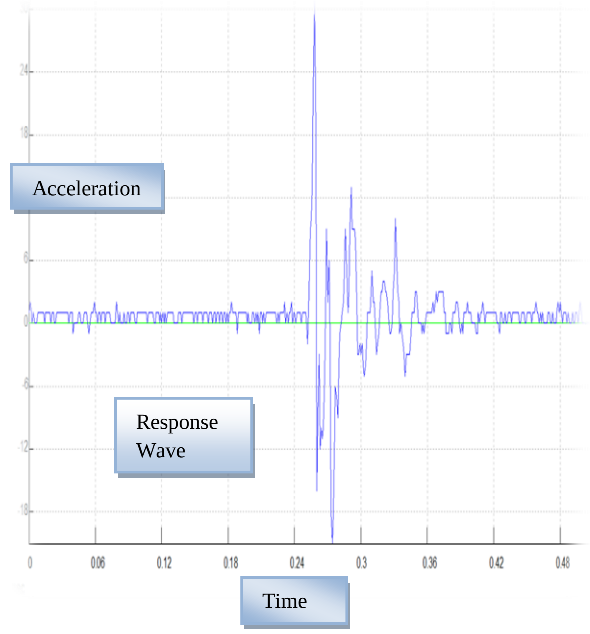

Figure 21 : Vibration responses for the old prostheses.

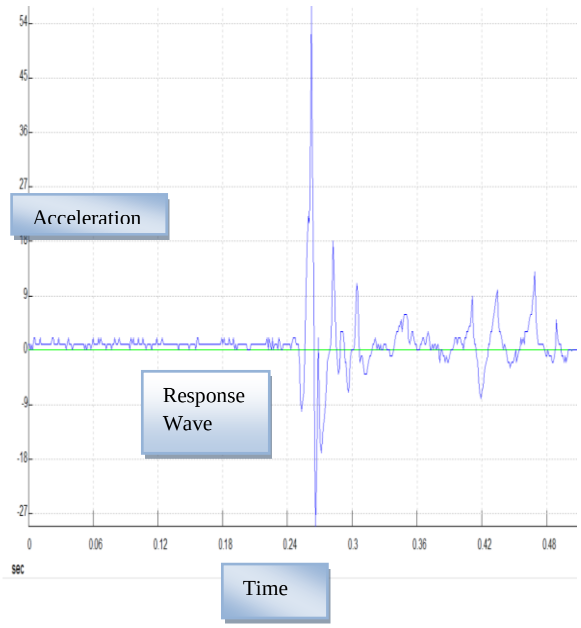

Figure 22: Vibration responses for the designed prostheses.

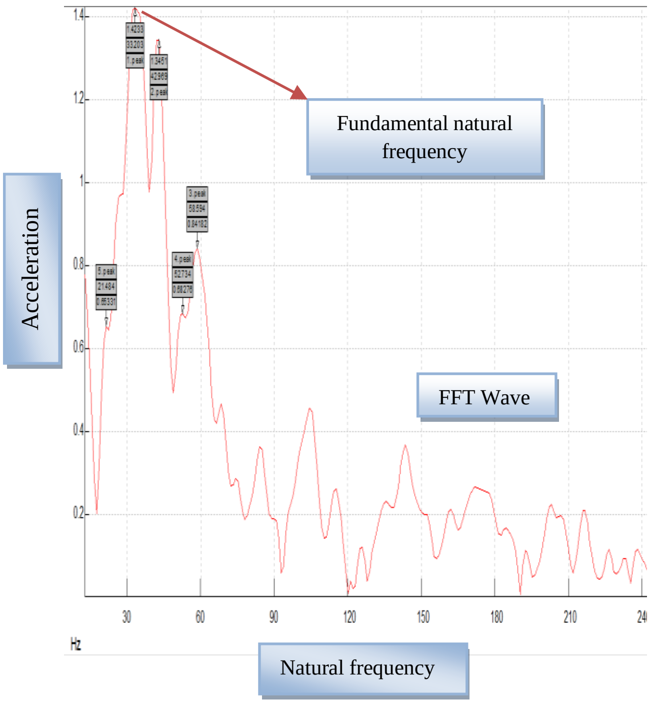

Figure 23: Sig-View Programs for FFT Analysis Function for old prostheses.

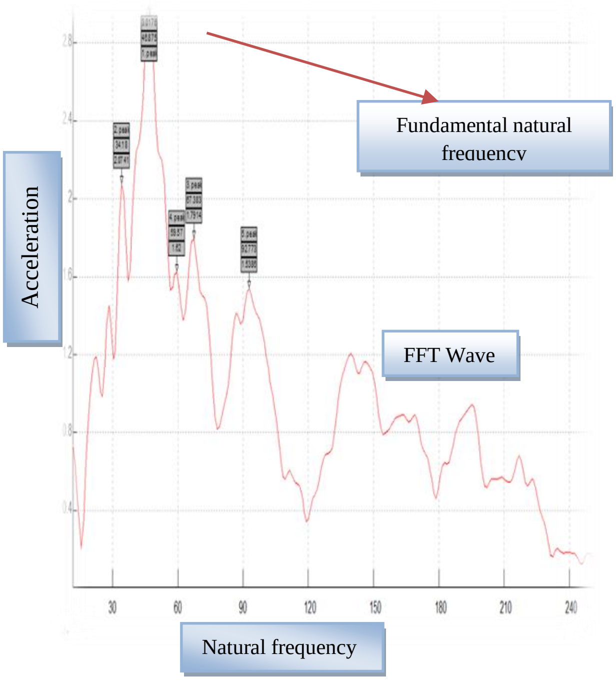

Figure 24: Sig-View Programs for FFT Analysis Function for design prostheses.

Figure 25: Hammer Test for the old prostheses and the design prostheses.

4.5 Tensile Results & Fatigue Results.

All 3 specimens were tested using a Tinius Olsen instrument for tensile testing. The sample of the lamination present 10 layers (4perlon 2 carbon fiber 4 perlon).The mechanical properties for all laminations can be listed in table 9. Fatigue failure occurs when the specimen fractures under alternative loading. The relationship between the stress and the number of cycle from fatigue testing is illustrated in fig.26.

Table 9: The mechanical properties of material socket.

Young's Modulus (GPa)	Yield Stress (MPa)	Ultimate Stress (MPa)
1.109	33	36

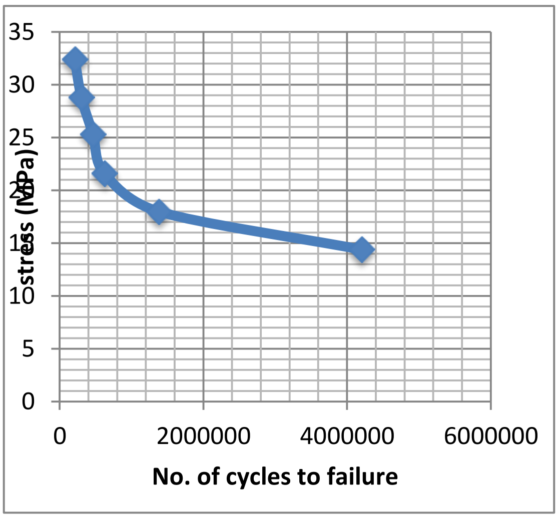

Figure 26: S-N fatigue tests curve for The lamination specimens.

Conclusions:

The following conclusions are drawn from the results obtained in this work**:**

1. **T**he designed prosthesis Gait cycle Time in the left leg is equal to (1.50sec)and right leg is equal to (1.58sec).
2. The max. Value of IP in design prosthesis is recorded at Tensor fascia lata muscle with values of 408 KPa.
3. The first mode of old prosthesis is starting from the value of 33.2 Hz,while the designed prosthesis first mode is starting from the value of 46.8Hz.
4. Maximum value of frequency to recorded in thigh for both case studies with the value 2.66 Hz in old prosthesis and 3.12Hz in design prosthesis. Both case studies have recorded maximum value of acceleration at thigh but design prosthesis has larger value of about $4.44^{m}/_{s^2}$ than old prosthesis $2.09^{m}/_{s^2}$.

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تحليل الاهتزاز لطرف مبتور خلال الركبه

محسن جبر جويج **جنه صادق جعفر**

كليه الهندسة - جامعة النهرين

الخلاصة:

الطرف هو امتداد اصطناعي يحل محل الجزء المفقود من الجسم نتيجة الحوادث او الفقدان الولادي او لعيوب لأسباب أخرى. ويمكن تأمل البتر خلال الركبه وكأنه بتر الـ Transfemoral. تضمن الجانب العملي في هذا البحث قياسات قوة رد الفعل الارضية GRFوتوزيع الضغط لكلا الطرفين المقترح منقبل الباحث والمستخدم إضافة إلى قياس الضغط المتولد بين الساق والوقب باستخدام جهاز F-Socket sensor. إضافة إلى ذلك فقد تم قياس التردد الطبيعي باستخدام Hammer Test وتسجيل بيانات (التعجيل- التردد) في مواقع مختلفه اثناء دورة المشي. من نتائج ال GRF للطرف المصمم وجد ان زمن دورة المشي في الساق اليسرى 1.50 sec والساق اليمنى 1.58sec والضغط المتولد في منطقة الفخذ هو408 KPa . بينت النتائج ان القيم القصوى للتعجيل - التردد في منطقة الفخذ هي(2.09 m/s2 and 2.66 Hz) في حالة استخدام الطرف الاصطناعي القديم و (4.44 m/s2 and 3.12 Hz) في حالة استخدام الطرف الاصطناعي المقترح من قبل الباحث، من خلال اختبارات الاهتزاز باستخدام الـ Hammer كان التردد الطبيعي للنسق الاساس 33.2 Hz و46.8 Hz لكل من الطرفين الاصطناعيين القديم والمقترح استخدامه.