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FINITE ELEMENT BASED MODEL FOR THE ASSESSMENT OF A PROSTHETIC FOOT STIFFNESS

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ABSTRACT

Prosthetic foot stiffness behavior is not fixed. Its variation depends not only on the foot shape and material, but also on the attitude of loads on the foot during walking. Many studies have evaluated foot stiffness by different ways. The purpose of this paper is to establish a new method for stiffness assessment depends on using finite element (FE) model for a Niagara foot at different load conditions that simulate what occurs in the gait. This technique is based on developing mathematical models of the force-displacement data, then the stiffness is determined mathematically corresponding to a set of four definite loads (250 N, 500 N, 750 N and 1000 N). Results showed that the developed technique was capable of determining the stiffness at any load. The modified models showed lower displacement and higher stiffness behavior compared with the Niagara Foot which provide less dynamic performance for the users with greater stability. The suggested technique simulated a methodology for a prosthetic foot designers to interactively vary the foot geometry or material and to track their effects on the stiffness properties and/or make comparisons. This technique can easily be standardized for evaluating the stiffness of any other type of prosthetic feet.

KEYWORDS: Biomechanics, Finite Element Analysis, Prosthetic foot, Prosthesis, Stiffness.

1. INTRODUCTION

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An important aspect of a prosthetic foot characterization that strongly influences its mechanical behavior and function is the structural stiffness [1]. Such characterization should substitute the loss of muscles and tendons of an intact biological foot [2]. The structural stiffness is not a fixed value since it is based not only the shape design, the type of material but also on the manner of loading upon the foot which changes during the walking process [3, 4]. Evaluation of the appropriateness of a prosthetic foot for an amputee requires study of both, its functionality and its mechanical behavior [5]. The commonly used method for characterizing the functional behavior of a prosthetic foot is the gait analysis. Using such a method, models the foot as a rigid body [6, 7]. Gait analysis of amputees is usually performed in order to study the kinematics and kinetics of the walking process [8-10]. The mechanical behavior of prosthetic feet, being evaluated by measuring their hysteresis and stiffness at several ankle positions [6], which usually requires special devices as well as prototypes of feet to be tested. On the other hand, the finite element analysis (FEA) can be used as another technique which depends on developing a graphical model of the prosthetic feet to simulate its mechanical testing. The FEA can permit the investigation of the responses of the deformable structures using known boundary conditions [6]. It has been broadly used to evaluate the stresses between the socket and the residual limb [11-13]. Few studies have concentrated on the analysis of the prosthetic foot at the stance phase of the gait cycle [14, 15]. In fact, using the FEA needs data about geometry, boundary conditions of the structure, and the material properties [6, 16]. This technique can be used to evaluate new and/or modified designs of the prosthetic feet. It has the advantage of testing new designs, even before their manufacture.

There are two main standards used in performing the mechanical testing of prosthetic feet, ISO-10328 and ISO-22675 which include the protocols to experimentally test the durability and performance of the lower-limb prostheses [17, 18]. The ISO-10328 prescribed four sequential testing sections: an initial static proof test, an ultimate strength test, a cyclic test and a final static proof test [19]. While ISO-22675, outlines a cyclic durability testing procedures for the lower limb prosthetic devices. It provides M-shape curves, which are the plots of both, the testing forces and the tilting angles of the loading platform upon a prosthetic foot versus time. This plot provides practical testing regimes designed to simulate conditions during the stance phase of prosthetic gait [14].

Tests by different researchers [3, 4, 20, 21] showed that, the load-displacement relationships obtained by testing a prosthetic foot are nonlinear and the prescription of stiffness is still not well-defined [7].

Independent measurements of material and structural properties including stiffness were performed experimentally on different types of prosthetic feet [22]. Only the forefoot portion was mechanically tested, and the forces applied did not reflect the peak loading during gait. All types of feet were then classified into one of four categories: the most stiff, the more stiff, the less stiff and the least stiff [22].

In another study [19], stiffness was determined as the slope of the linear best-fit of the force-displacement curve during the loading portions of the initial and the final proof tests. The tests were performed according to ISO-10328 in the purpose of studying the performance and robustness of three different types of feet.

In order to study the effect of thickness variations of the upper part of the S-region in the Niagara prosthetic foot, on its stiffness behavior [3]. Four models of feet were subjected experimentally to compressive loading at the heel and the forefoot portions separately. Slopes of the secant lines of the force-displacement curves in the intervals of loading between 400 N and 1000 N were only considered. This range was considered to cover the average of the major loading ranges during the gait [3].

Another study was conducted in order to develop a method of characterizing the mechanical properties of the Niagara, SACH and ESAR commercial feet [21]. In this study the mechanical compressive tests, which based on that outlined in ISO-10328 were performed. The force-deflection responses were described in two regions. The initial, was the linear part of the force-deflection curve and denoted as the initial stiffness (S1), while the nonlinear part for the rest of curve up to a design load was denoted as S2. A third value (Sh) was also taken as the average stiffness between the no load and the design load in the curve. This study indicated that, multiple stiffness values were needed to accurately describe the non-linear mechanical behavior of the feet, [21].

Few studies have considered the use of FE analysis technique in evaluating the mechanical behavior of prosthetic feet [6, 15, 20]. A study comprising mechanical tests as that proposed by ISO-10328 were performed on a J-shape foot model [6]. These tests were then emulated using the FE analysis. Stiffness results by either the experiments or the FE curves were allocated as linear stiffness. This study illustrated that the FE analysis of the forefoot loading were matching with those obtained by the mechanical testing.

For the purpose of investigating repeatability of the mechanical tests on the prosthetic feet, a study on a Niagara foot model involved FE modeling were conducted [20]. In this study, the stiffness of foot were described by considering three different values K1, K2 and KH. The value K1 was corresponding to be the slope of the force-displacement curve from the starting zero load up to an elbow point. While, K2 was corresponding to the slope of the curve from the elbow up to 1000 N load, and KH was corresponding to the slope of the line from zero to 1000 N loads.

A methodology for evaluating the mechanical properties of the prosthetic feet was conducted through applying mechanical loads at different angles on the heel and the toe portions of the feet [14]. These loads were extracted from the waveform of the ISO-22675. Another numerical study, using FEA, on a novel design of a Dynamic Energy Return prosthetic foot, based on applying this methodology, was conducted. For this study the displacement profile and stiffness characteristics of the novel foot was compared with those of the Niagara, AXTION and SACH commercial prostheses [15].

Although many of researchers [3, 4, 6, 19-21] followed the ISO-10328 in performing mechanical tests on different types of prosthetic feet. They did not evaluated stiffness by the same way. It seemed from their studies that, no standard method was followed in evaluating the structural stiffness of the prosthetic feet and so a more controlled approach is still needed in prescribing the prosthetic feet stiffness [21]. The aim of this study was to develop an independent methodology which can be standardized for characterizing stiffness of the prosthetic feet. Although the proposed methodology was based on simulating mechanical testing as that prescribed by the ISO-22675 [14]. The stiffness was determined from the force-displacement curve at predetermined fixed levels of loading. Such methodology can be applied for evaluation and comparison purposes of different prosthetic feet designs. In this study the methodology was applied to study the effects of thickness of the S-shape and material variations on the stiffness of a modified model of Niagara foot.

2. METHODOLOGY

The proposed procedures, for the numerical evaluation of the structural stiffness of the prosthetic foot, using the FE tests, consisted of four main stages, Fig. 1. These stages were performed considering two different regimes of loading which were based on using either the standards ISO-10328 or ISO-22675. These stages consisted of creating a the prosthetic foot shape model, selecting its material, setting up the boundary conditions of loading, and performing analysis of the FE test results. The results were compared with that pronounced by Schmitz [20].

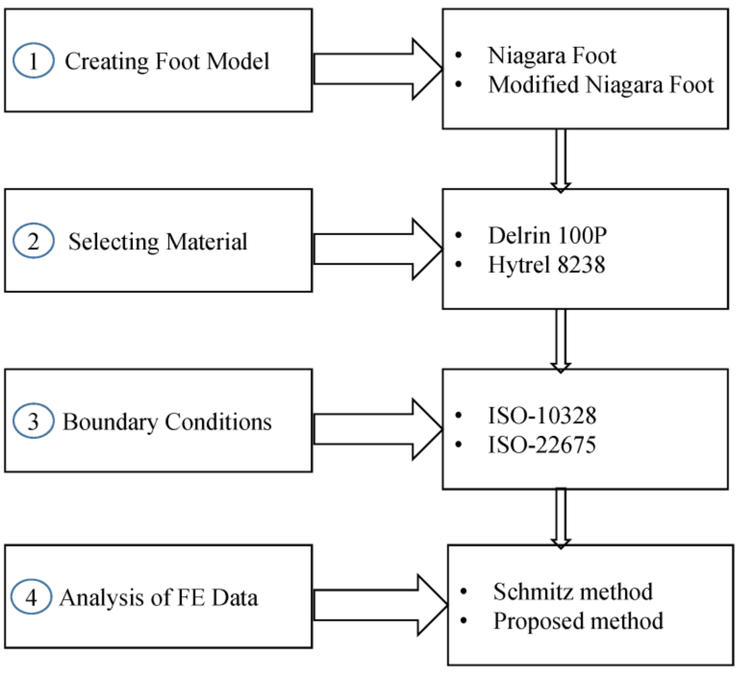

Fig. 1 Overall flow of the methodology.

2.1 The Foot Shape Model

A three-dimensional shape of the Niagara foot (NF) model 1 version 3 was first drawn, with the original geometric dimensions, [20]. Such drawing was prepared using SolidWorks® Premium 2014 Edition software, Fig. 2, then a set of four modified shape models were also created. All models were of identical dimensions except the thickness of the upper part of the S-shape "B", Fig. 3. This thickness was altered by shifting the center of outer semicircle horizontally from the center "O" by distances of 5 mm, 10 mm and 15 mm. The models were nominated M1, M2, M3, and M4 respectively, so the values of "B" were 12.5, 17.5 mm, 22.5 mm and 27.5 mm respectively, Fig. 3 [4].

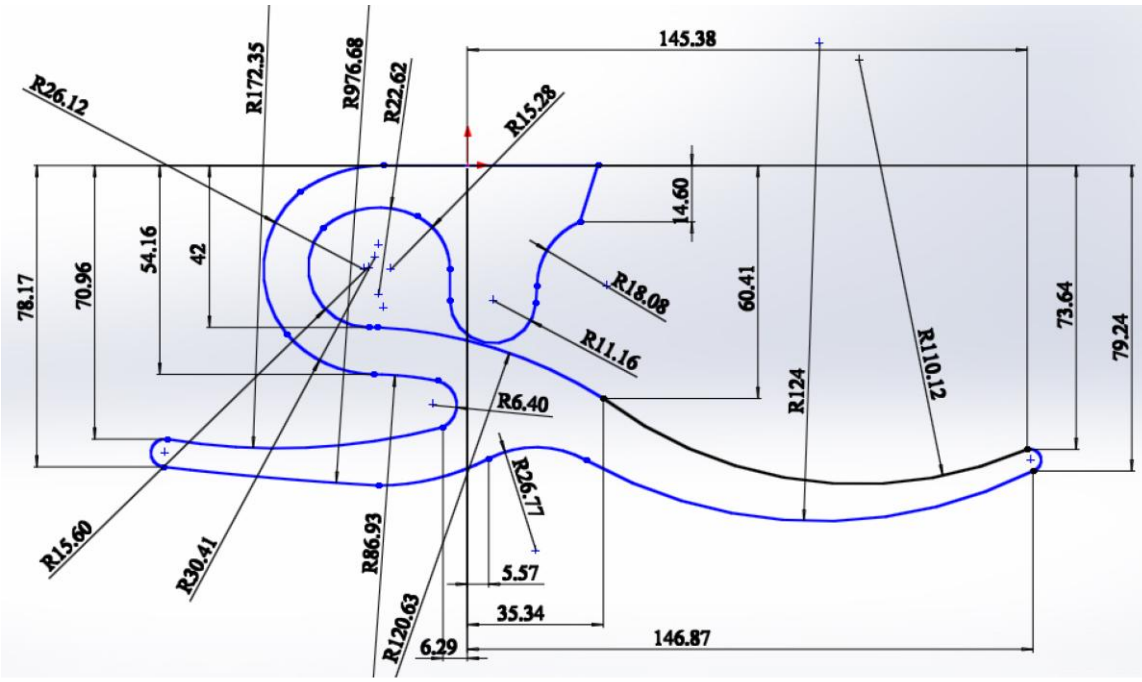

Fig. 2 The basic dimensions of the Niagara foot model (thickness = 60 mm).

2.2 Types of material

Delrin 100P was selected to represent the material of the original Niagara foot (NF) model and the proposed four modified models M1 to M4. In order to study the effect of material on the stiffness of prosthetic foot, Hytrel 8238 was selected to represent another type of material for only the four modified models (M1 to M4). Both the Delrin l00P and the Hytrel 8238 were generally used in different studies [14, 20].

These two materials were assumed to have linear, isotropic, and elastic stress-strain behavior. A summary of the mechanical properties of such materials is shown in Table 1.

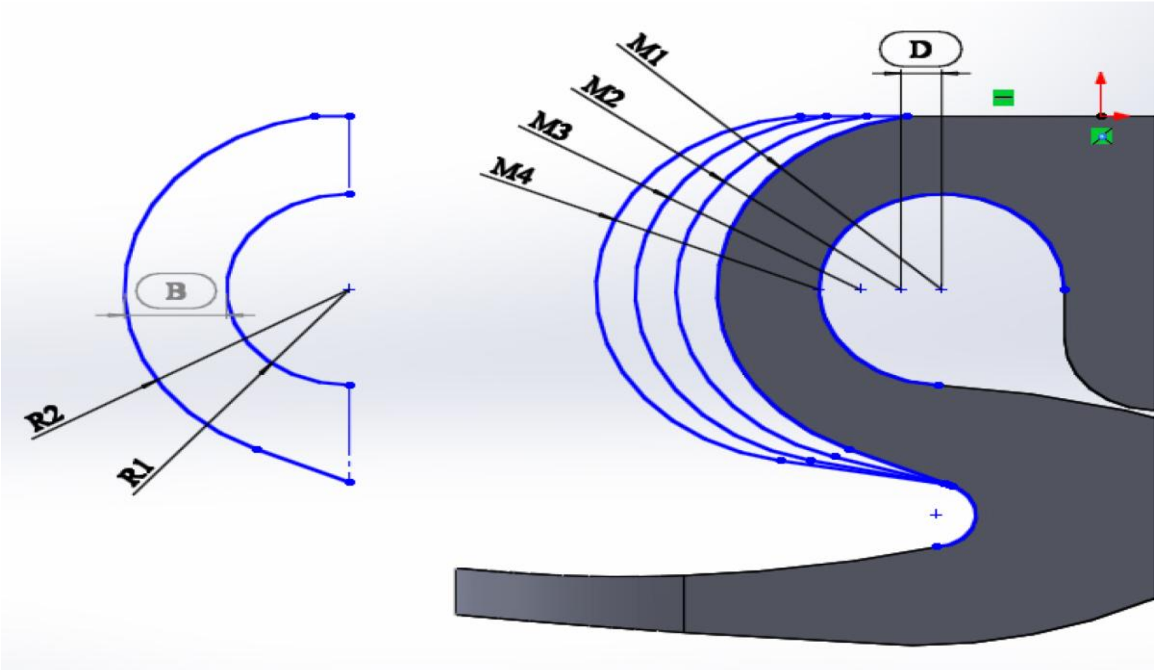

Fig. 3 The Niagara foot models with different S-shape thickness.

Table 1. Mechanical Properties of the Three Materials [23]

Material Name	Mass Density (kg/m^3)	Poisson's Ratio	Elastic Modulus (MPa)	Tensile Strength (MPa)	Yield Strength (MPa)
Delrin 100P	1420	0.35	2900	68	70
Hytrel 8238	1280	0.45	1180	48.3	36

2.3 Boundary Conditions

Simulated mechanical compressive tests were conducted using the simulation package of SolidWorks®. The boundary conditions of these tests were set as that described by the ISO-10328. Where the loading was applied through displacing a virtual platen against the heel and the toe portions separately, at two different angles, Fig. 4. The tests were carried out on each of the four modified foot models (M1 to M4) and on the original Niagara foot model (NF), where the results of these tests were compared with that obtained in [20].

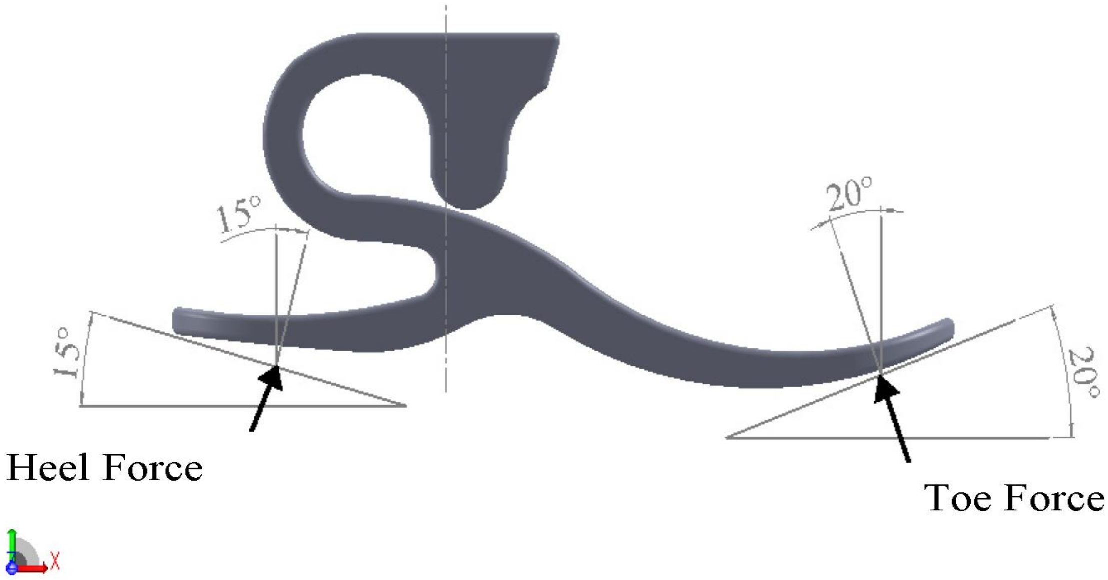

Fig. 4 The direction of the applied forces on the heel and on toe portions (ISO-10328).

The platform of the foot was set to be fixed (immovable in the x, y and z directions) Fig. 5. Contact settings between the platen and either the heel or the toe surfaces were adopted as frictionless without penetration. The prosthetic foot was set to be self-interacted, since it had a special design, that it contained prongs. Another boundary condition for the type of contact between the two prongs and the opposite face was also managed. Such contact was arranged to be surface to surface frictionless contact without penetration, Fig. 5. A coordinating system was also created on the platen in order to control its movement, where remote displacements of the platen were generated to provide loads on the heel and toe modes of loading. The coordinate system was originated (x = 0, y = 0 and z = 0) in the center of the surface of platen. The platen was allowed to move along the “y” direction only without any rotation.

Meshing is a crucial step in the design analysis. The SolidWorks program automatically assigns the appropriate mesh type to the object based on its geometry features. It created a solid mesh with tetrahedral solid elements in the foot solid shape, Fig. 6. For all proposed models of the foot, the SolidWorks assigned different sizes of elements ranged between 6.5 mm and 7.5 mm. These values were readjusted to be 5 mm for all the models. Table 2 shows the total nodes and elements for the original NF and the four modified models.

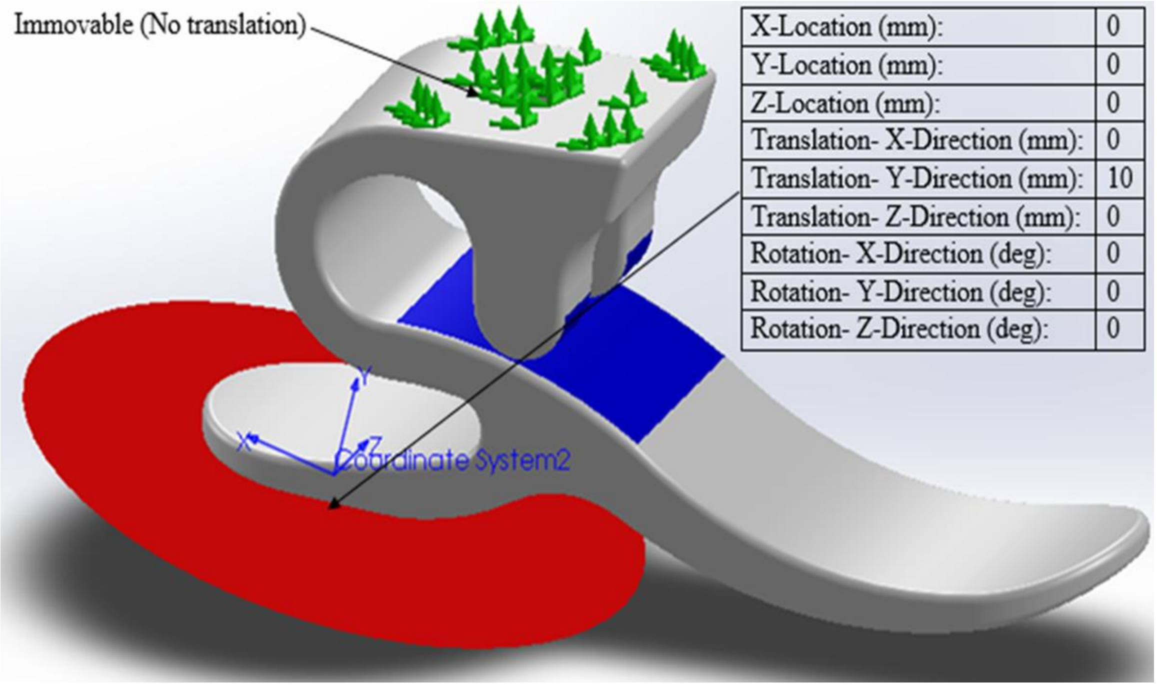

Fig. 5 Setting of the boundary conditions.

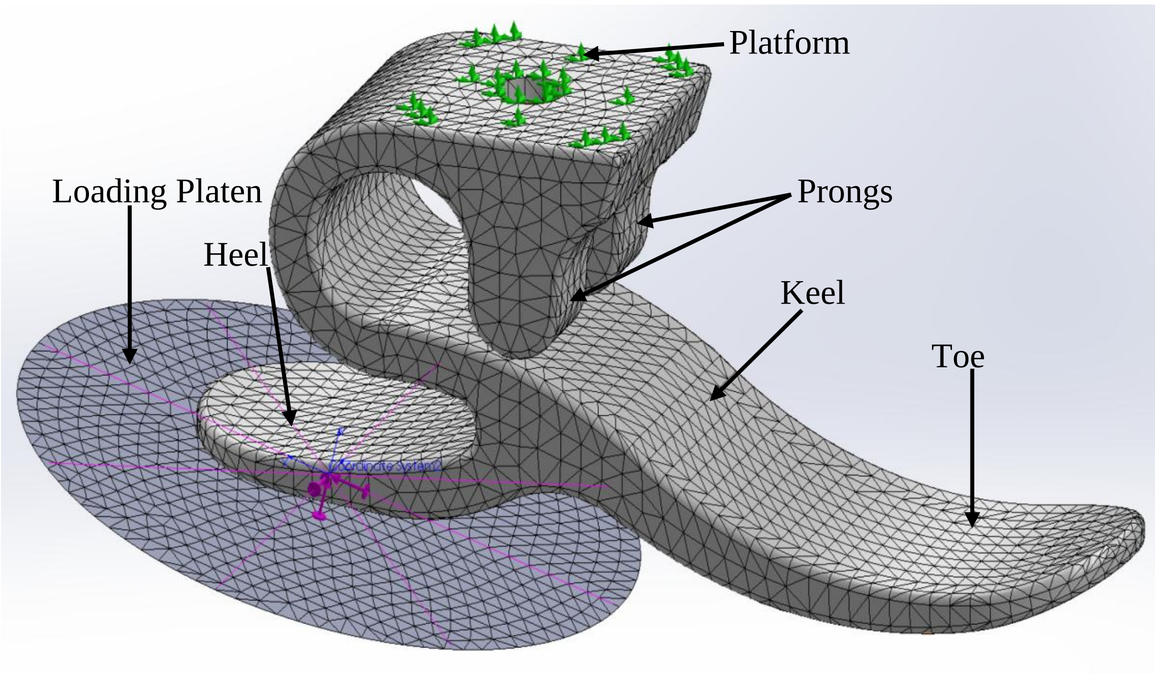

Fig. 6 Meshing of the foot and the loading platen.

Table 2 Meshing data for the models

Model	Meshing Data	
	Total Nodes	Total Elements
NF	35082	21174
M1	34792	21166
M2	35920	21955
M3	37290	22922
M4	38649	23844

Further FE tests were carried out to study the effect of thickness of the S-shape “B” and the material variations on the stiffness behavior of the four modified models (M1 to M4) of the foot. The models were represented by two different materials (Delrin l00P and Hytrel 8238). All feet were subjected to loading conditions conformed to that prescribed by the ISO-22675. The values of forces as well as their corresponding directions were simulating the expected vertical ground reaction loads on the foot (P4 loading curve) that could be occurring during the stance period of gait (600 msec.) [14].

2.4 Analysis of the Finite Element Data

The force-displacement data, obtained from the FEA, of the heel and toe portions, ranged from 0 N up to 1000 N of loading. Two methods were deduced for describing stiffness of the toe and heel portions of the Niagara foot. The first method, considered the stiffness values K1, K2 and KH [20]. The second method, approximated the data of the force-displacement using least square method to obtain a mathematical model. The mathematical expression describing the load-displacement relationship, was tested by applying polynomial functions of 2^{nd} up to 6^{th} order. The 4^{th} order polynomial function was chosen as it fulfilled the best-fit for the data, (R^2 = 0.9999). Stiffness were determined mathematically, as the slope of the force-displacement function at four different selected foot loads (250 N, 500 N, 750 N and 1000 N). The determined stiffness were nominated as SH250, SH500, SH750, and SH1000 respectively, for the heel stiffness. Toe stiffness were also similarly obtained and nominated ST250, ST500, ST750, and ST1000 respectively.

3. RESULTS

Results of the FE tests were divided into two main regime. The first was based on the ISO-10328 standard. The load-displacement data obtained from the FEA were interrelated by two different prescribed methods. The second was based on the ISO-22675 standard. The analysis of results of both regimes dealt with the effect of the independent variables (the thickness “B”, and the type of material) on the stiffness of the heel and the toe portions of the modified Niagara models.

3.1 The First Numerical Testing Regime Based on ISO-10328

3.1.1. Stiffness of the toe

Figure 7 shows a plot of the force-displacement data obtained from the FEA of the toe at 20° angle of loading. The toe initially behaved as a soft component where the stiffness “K1” had remarkably low value. Stiffness progressively increased up to the end of loading. An elbow point, on the curve, was noticed at about 2.5 mm of displacement. After this displacement, stiffness of the toe exhibited rapid increase, which attributes to the closing of the gap between the prongs and keel as shown in Fig. 6. Consequently, the measure of this gap was expected to affect the location of the elbow point. The maximum reached load (1000 N), was corresponding to 16.4 mm toe displacement. Table 3, shows the stiffness K1, K2 and KH of the toe as determined by the FEA results versus the corresponding results in [20].

The new values of stiffness, ST250, ST500, ST750 and ST1000, were determined from the FEA of the toe loading. The data about the Force-displacement relationship were firstly modeled as a 4^{th} order polynomial function, this differentiated to determine the stiffness values (slopes of the curve) were corresponding to the pre-specified loads, 250 N, 500 N, 750 N and 1000 N. Such values were 61.7 kN/m, 73.0 kN/m, 83.4 kN/m, and 100.1 kN/m respectively.

3.1.2. Stiffness of the heel

Figure 7 illustrates a plot of the force-displacement behavior of the heel which resulted from the FE test. The heel, generally, showed higher values of load versus displacements as compared to the toe testing. This was reflected in its stiffness results. The force-displacement curve of the heel test, did not reveal an elbow shape as observed in the toe test. Stiffness of the heel progressively increased with the increase of loading up to the maximum load (1000 N). The displacement of the heel that corresponded to the maximum value of loading, was 8.9 mm. Table 3, shows the stiffness values, K1, K2 and KH of the heel as obtained by the FEA data from this study versus the corresponding values in [20].

Similar procedures were followed, as those explained for the toe, in determining the heel stiffness, SH250, SH500, SH750 and SH1000. The corresponding values obtained were 66.5 kN/m, 156.7 kN/m, 264.9 kN/m, and 367.8 kN/m respectively. Although, these results of stiffness were only for a single loading angle (-15°), but they were more descriptive as they covered a wider range of loading.

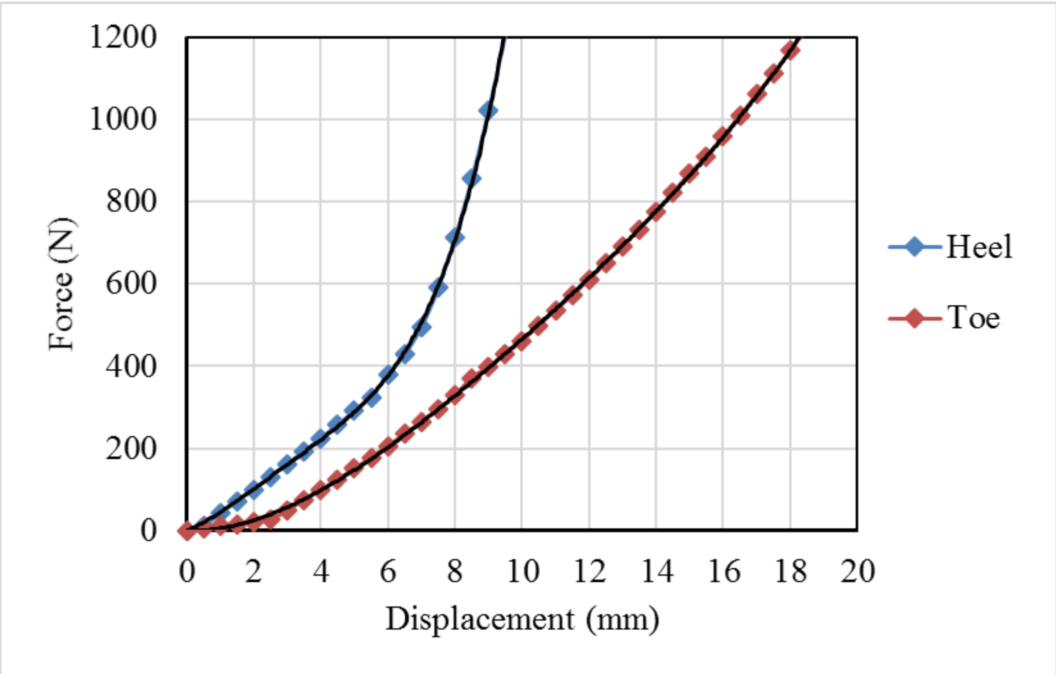

Fig. 7 Force-displacement relationship for the toe and heel mode of loading at 20° and -15° respectively.

Table 3. Stiffness of Toe and Heel of the Niagara Foot at 20° and -15° Respectively

Type of Test		K1 (kN/m)	K2 (kN/m)	KH (kN/m)
Toe	FEA	10.9	69.9	60.9
	FEA [20]	20.3	74.9	64.1
	Mechanical Testing [20]	10.4	68.5	55.1
Heel	FEA	59.0	196.6	111.9
	FEA [20]	139	260.4	174.8
	Mechanical Testing [20]	75.7	224.5	124.0

3.2 The Second Numerical Testing Regime Based on ISO-22675

3.2.1. Effect of thickness B on the displacement and stiffness

The FE testing upon the heel showed that the thickness “B” had a remarkable effect on the foot displacement response, where the increase of thickness “B” with the prongs enhances the resistivity response to the load. While, the thickness “B”, had an unnoticed effect on the toe stiffness, which could be attributed to closing the gap between the prongs and the keel.

Figure 8 shows a plot of the displacement patterns with time, which were extracted from the displacement-angle data. This figure was adopted from the FE testing of the modified models (M1, M2, M3, and M4) and the original model of the Niagara foot (NF), through proposing Delrin 100P as the material of the feet. This plot illustrates the displacement of both the heel and the toe during the simulated stance time (600 msec.). The displacements of the heel increased sharply from 0 mm, (Heel Strike), then incremented at lower rates until reaching peak values (about 6 mm) at 150 msec., while the NF reached 9.4 mm, then gradually decremented back to almost 0 mm by the end of the heel contact (300 msec.). At such instant, the toe was already displaced to about 5 mm and the NF was 7.3 mm, where its displacements continued the increase up to another higher peak value (about 15 mm) at 487 msec., while the NF was 19.5mm, then sharply decremented to 0 mm at the end of the stance at 600 msec., (Toe off). Comparing the two peaks of the heel and the toe, the heel was highly stiff as compared to the toe. The modified models showed lower displacement behavior compared with the NF where the displacement was adversely affected by the thickness “B.”

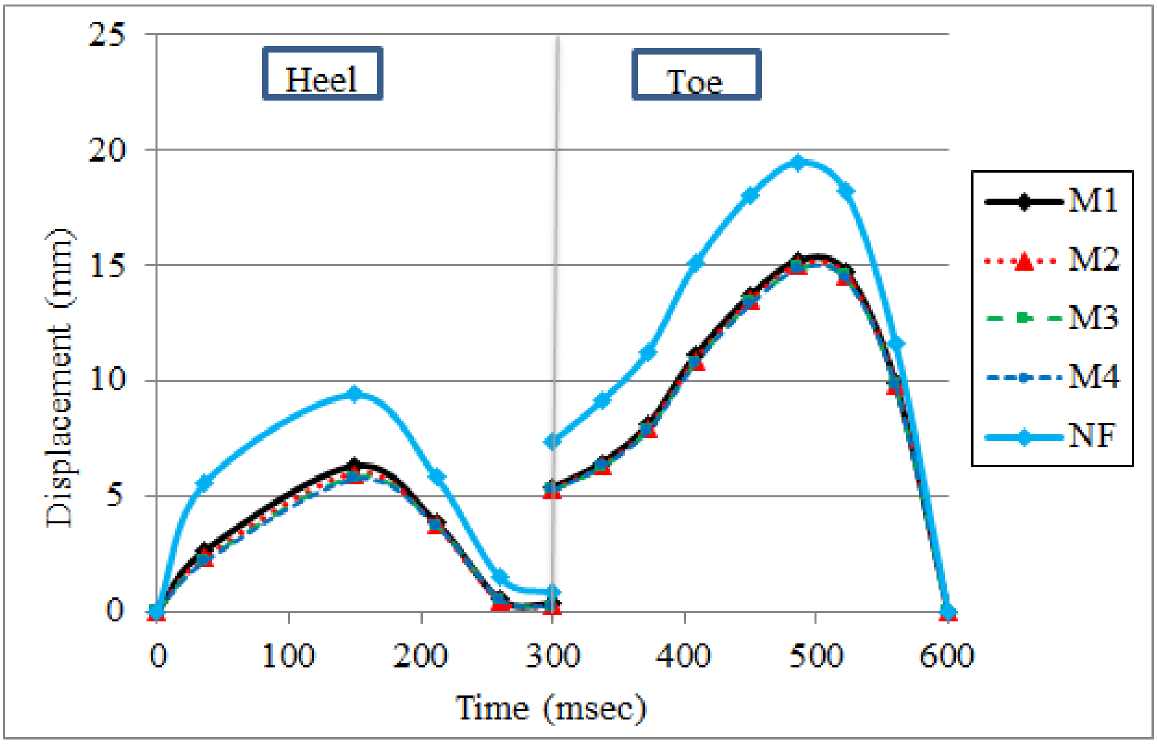

Fig. 8 Prediction displacement of the foot models M1, M2, M3, M4 and NF during the simulated time of stance (Delrin 100P).

Figure 9 and Fig. 10 show the patterns of stiffness change versus time for the heel and toe respectively, where the stiffness data of these plots were determined corresponding to the 15 set points of the virtual stance period. These plots belong the four models (M1 to M4) and the original design of Niagara foot. These special plots, show that the toe of the foot was too soft as compared to the heel. Stiffness of heels were gradually increasing with time starting from 0 up to nearly 200 msec., then steeply up to the end of the heel contact (300 msec.), while stiffness of toes declined from 300 msec. to 600 msec. In general, the stiffness values of both the heel and toe of the NF were less as compared to any of the four models (M1 to M4).

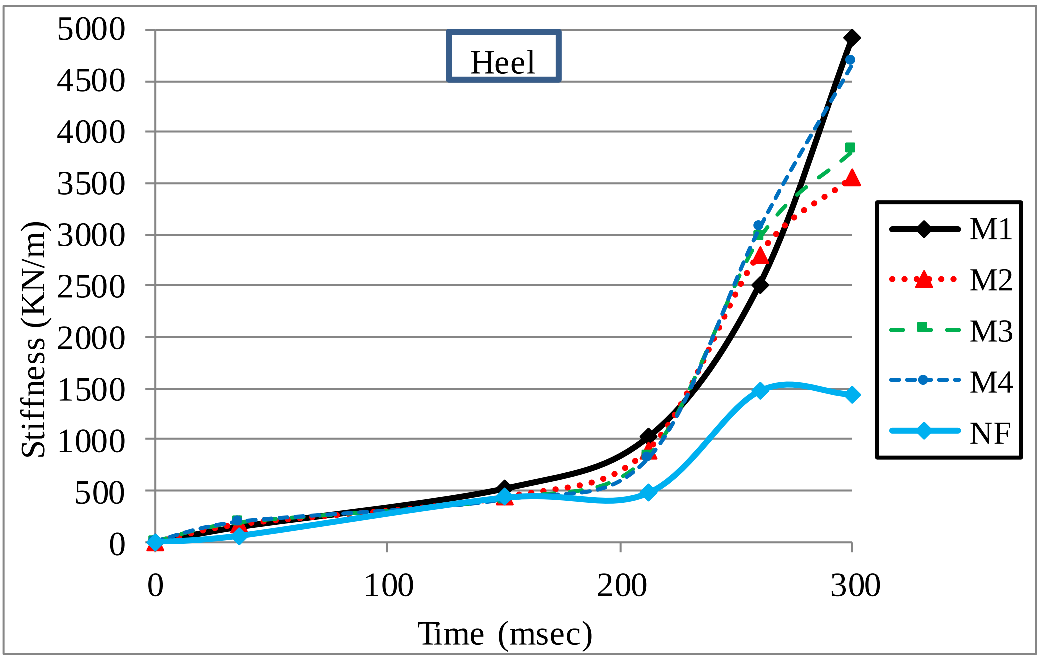

Fig. 9 Prediction stiffness on the heel of the foot models M1, M2, M3, M4 and NF during the simulated time of stance (Delrin 100P).

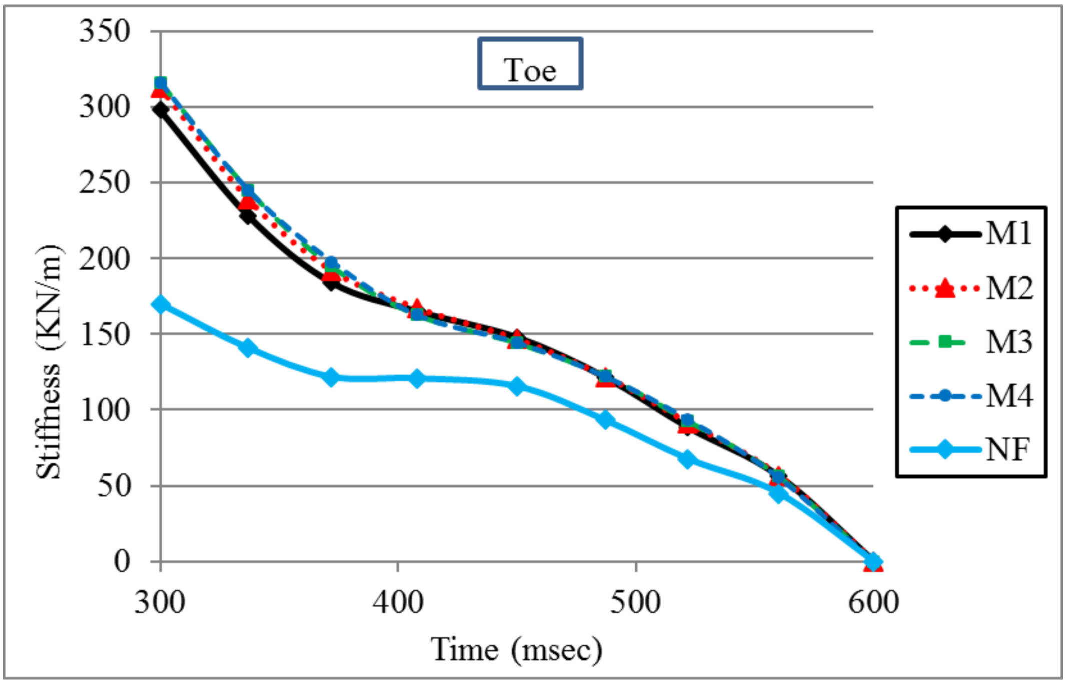

Fig. 10 Prediction stiffness on the toe of the foot models M1, M2, M3, M4 and NF during the simulated time of stance (Delrin 100P).

3.2.2. Effect of material on the displacement and stiffness

Figure 11 shows the displacement-time trends of the heel and toe parts of the foot model M1, which were resulted from the FE analysis of the two imposed materials. The foot of Hytrel 8238, showed higher displacements responses as compared to that of Delrin 100P. Such responses were changing with time, where the effect of material was less pronounced at earliest instants of the heel contact and before the heel rise. Regarding to the effect of material on the toe displacement, the differences between the displacements of the toes were increasing till reaching the peaks then declining with the progression of time.

Figure 12 and Fig. 13 show plots of the stiffness behavior of the heel and toe parts of M1 respectively. These figures illustrated smaller differences in the stiffness between the two materials except in the later parts of the load response to the heel takeoff (From 200 msec. to 300 msec.). The stiffness of the two feet were increasing with the stance time in the heel loading stage, while it was decreasing in the toe stage. The Delrin material showed greater stiffness behavior as compared to the Hytrel material.

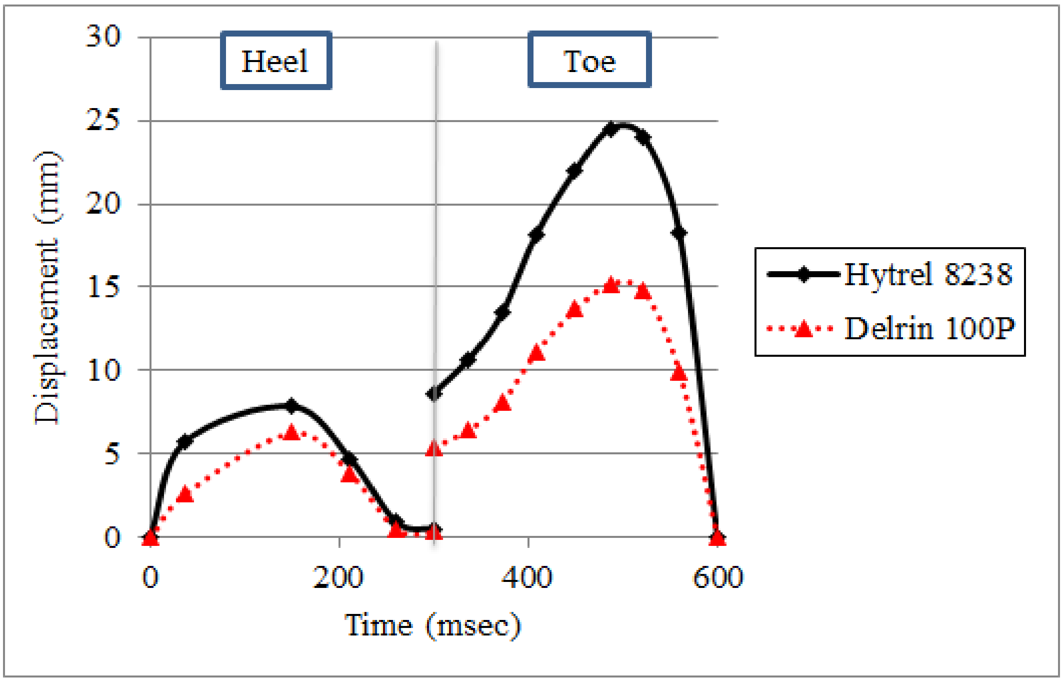

Fig. 11 Predicted displacements of the foot model M1 of the two material Delrin 100P and Hytrel 8238 during the simulated time of stance.

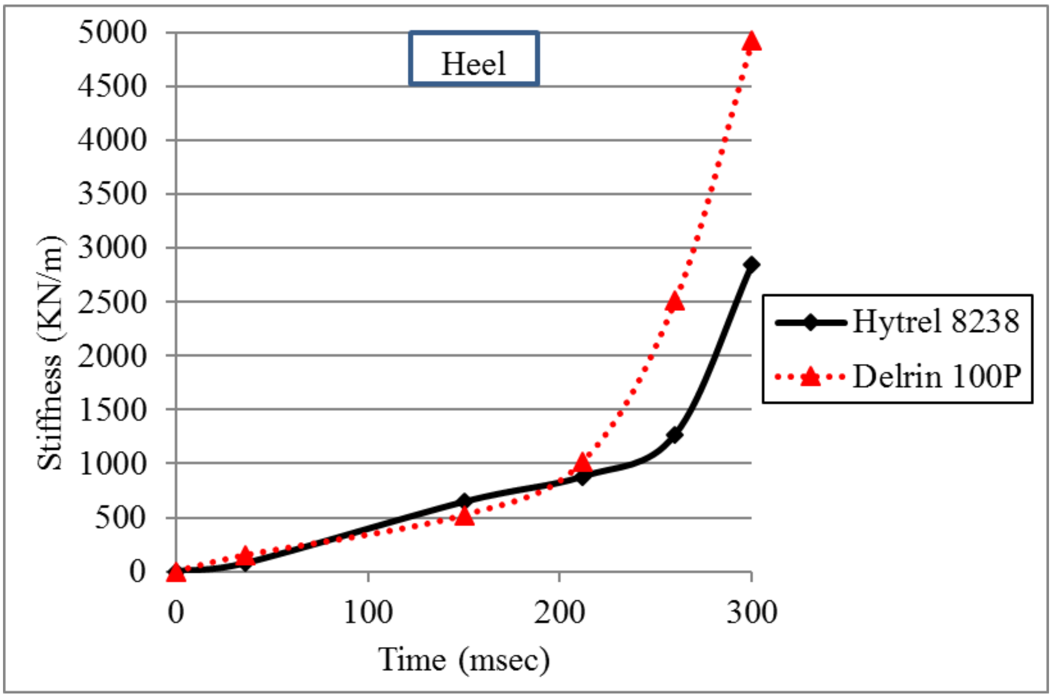

Fig. 12 Predicted stiffness on the heel of the foot model M1 of the two material Delrin 100P and Hytrel 8238 during the simulated time of stance.

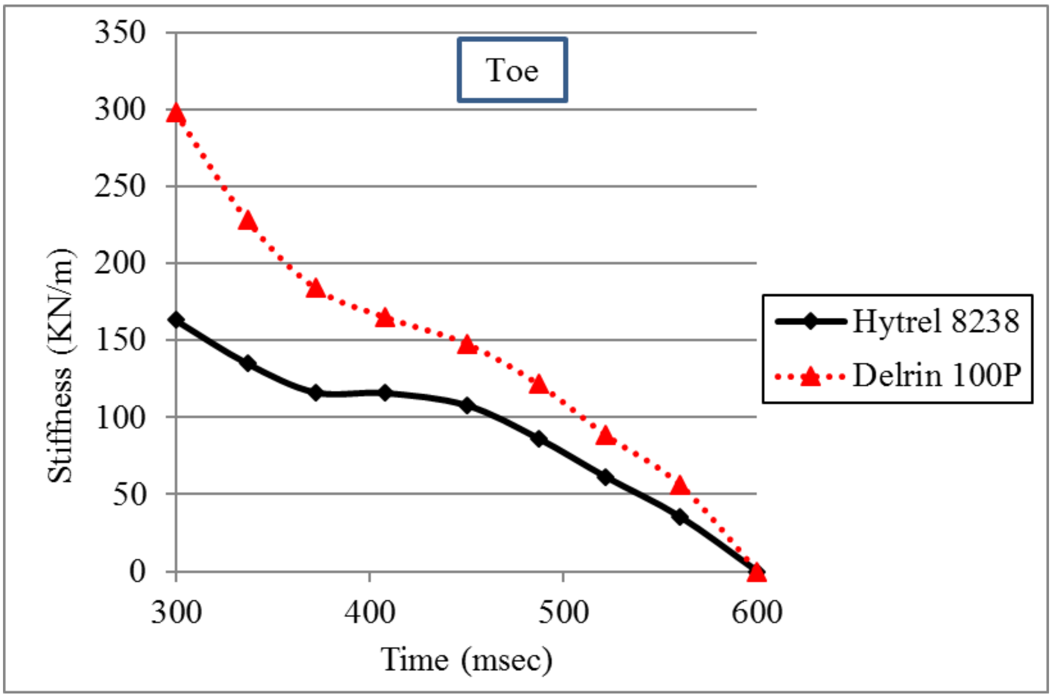

Fig. 13 Predicted stiffness on the toe of the foot model M1 of the two material Delrin 100P and Hytrel 8238 during the simulated time of stance.

4. DISCUSSION

Many studies, that were relevant to prosthetic feet testing, have followed the conditions prescribed in the ISO-10328 in performing their mechanical tests, but stiffness of feet were evaluated by different methods. Some other studies did not follow the ISO-10328 where the toe region at only a single value of angle 12° and with forces applied on the foot less than the maximum proposed load by ISO-10328. Stiffness of toe was considered as the slope of the line that approximated the force-displacement curve [22]. Other study followed the ISO-10328 in performing both mechanical as well as numerical tests. Stiffness of the prosthetic foot was then approximated to be a single value [6]. Another study considered stiffness of foot to be the amount of slope of the best-fit line for the force-displacement relationship. The line was taken for the interval of foot loading between the initial and the final loads in its static proof test [19]. Else, the stiffness has considered to be the slope of a secant line of the force-displacement curve in an independent interval of loading between 400 N and 1000 N [3]. All these studies simplified stiffness of the prosthetic foot to be a single value, which were not sufficient or realistic to describe the stiffness behavior in the most different situations of the foot loading.

A more realistic method of stiffness assessment was assigned by considering three stiffness parameters K1, K2 and KH [20]. Comparing the stiffness values of the toe and heel as determined by the FEA data of this study and those obtained by Schmitz [20], in one side, to the corresponding values of the mechanical tests by his study Table 3. Results showed differences due to some variations in the geometry of the foot together with the conditions of the actual mechanical testing. Also, results of Schmitz's study showed that all the derived stiffness values, achieved the least deviations from those of the mechanical tests. Such values were mainly dependent on personal estimate for determining the location of an elbow point in the force-displacement curve. Moreover, such point was a characteristic of such Niagara foot design caused by closing the gap between the prongs and the keel, and consequently, it might not occur in other different designs. Another method adopting three stiffness parameters, but at different loading angles [21], a criticism may be directed to this method, since it depended on dividing the force-displacement responses into two regions only, which can be affected by the personal judgment.

The proposed method, in this study, covered a wider range of load-displacement behavior, mimicking what could happen upon a real prosthetic foot. As the prosthetic foot was subjected to different levels of load based on the different conditions of the users' characteristics and the walking process. Moreover, these values resulted from the methodology based on the mathematical model. This methodology can be applied to estimate the values of stiffness for any type of prosthetic feet. By finding the equation of the force-displacement relationships, it allows estimating the stiffness at any load point. Also, using such method allows to compare the stiffness of prosthetic feet with different designs and different materials.

The numerical technique of this study can provide estimates of stiffness in a variety of prosthetic foot orientations, using determined set of angles and loads, so that researchers and clinicians can begin to utilize this important property to compare between different prosthetic foot designs or to evaluate their appropriateness with an amputee.

For the evaluation of displacements and stiffness at the 15 critical points, the lower displacement and higher stiffness values indicated that the modified models become stiff compared to the original Niagara foot. Increasing the thickness “B” led to increase the stiffness of the foot especially on the heel compared to other models [14, 15]. Apparently, the stiffness of the heel were affected, not only by the angles of loading, but also from many different factors such as, the location of contact between the loading platen and the heel, and the amount of gap between the prongs and the keel. These two factors were relevant to the profile of the foot .

Actually, the Niagara foot was characterized by the special shape in its rear portion. This portion looks like S-shape, and behaves as a cantilever, where its fixed end was connected to the immovable platform while its free end the heel was subjected to the loading from the movable platen. The loading on the heel caused its deflection, which accompanied the deformation in the rear portion of the foot. Such deformation raised the combined compressive and bending effects upon loading the heel. Consequently, a dramatic increase of the heel stiffness, occurred through loading the foot at the -5° angle, where this increase was accompanied by closing the gap. Although, this gap was also closed during the 0° loading angle, the stiffness were affected, not only by the angle of loading, but also by the changes in the location of contact between the heel and the platen due to deformation of the foot. On the other side, when the loading applied on the heel at any of the other three angles (-10°, -15° and -20°), the gap was kept open, and thus it did not affect such patterns of stiffness. From this observation, we may conclude that, the profile of the foot surface in contact with the ground, contributes with the other structural properties of the prosthetic foot, on the foot stiffness behavior.

The type of material had a considerable effect on both the heel and toe stiffness. The Delrin 100P showed lower displacement and higher stiffness behavior compared to the Hytrel 8238 materials. The modified model M1 using the Hytrel material showed higher stiffness as compared to the Niagara Foot Model which was made of Hytrel [14].

In general the modifications on the Niagara foot model showed that the modified models provide less dynamic performance for the users with greater stability.

5. CONCLUSIONS

To mathematically express stiffness of a prosthetic foot under different conditions of loading in its design stage, a methodology based on the FEA technique and curve fitting was developed. For example, fitting the force-displacement data of the Niagara foot using the conditions specified in this study, resulted a set of fourth order polynomial functions.

Although this study provides a systematic method to assess the stiffness behavior of the Niagara foot at different conditions of loadings (magnitudes and directions), this method can be extended to assess the stiffness behavior of many other types of prosthetic feet. This methodology provides a more accurate explanation for the prosthetic foot behavior as compared to the existed approaches. This procedure includes a simulation for prosthetic foot designers to interactively vary foot geometry and material for different proposed foot designs, then track their effects on their stiffness properties.

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تقدير صلابة القدم الاصطناعية بواسطة النمذجة بالعنصر المحدود

صلابة القدم الاصطناعية ليست قيمه ثابتة حيث لا تتوقف فقط على شكل القدم ونوع المادة، ولكن أيضا على تغير الأحمال أثناء المشي. العديد من الدراسات السابقة قيمت صلابة القدم بطرق مختلفة. والغرض من هذا البحث هو وضع طريقة جديدة لتقييم صلابة القدم تعتمد على استخدام العنصر المحدود لنموذج قدم من نوع نياجارا في ظروف احمال مختلفة والتي تحاكي ما يحدث اثناء المشى. هذا الأسلوب يعتمد على إستخدام معادلات رياضية لبيانات القوة والإزاحة، حيث يتم تحديد الصلابة رياضيا عند أربعة أحمال محددة (٢٥٠ نيوتن، ٥٠٠ نيوتن، ٧٥٠ نيوتن و ١٠٠٠ نيوتن). أظهرت النتائج أن إستخدام الاسلوب المتبع قادر على تحديد صلابة القدم عند اى حمل. وأن السمك "B" وكذلك نوع المادة له تأثير واضح على صلابة القدم الهيكلية حيث ان النماذج المعدلة أظهرت إزاحة أقل وصلابة أعلى مقارنة مع قدم النياجارا وبذلك توفر أداء أقل ديناميكية مع قدر أكبر من الاستقرار. وإقترحت الدراسة منهجية جديدة يمكن استخدامها فى تقييم صلابة القدم الهيكلية خلال مرحلة التصميم.