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Finite element analysis for the evaluation of the structural behaviour, of a prosthesis for trans-tibial amputees

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ABSTRACT

The finite element analysis (FEA) has been identified as a useful tool for the stress and strain behaviour determination in lower limb prosthetics. The residual limb and prosthetic socket interface was the main subject of interest in previous studies. This paper focuses on the finite element analysis for the evaluation of structural behaviour of the Sure-flex™ prosthetic foot and other load-bearing components. A prosthetic socket was not included in the FEA. An approach for the finite element modelling including foot analysis, reverse engineering and material property testing was used. The foot analysis incorporated ground reaction forces measurement, motion analysis and strain gauge analysis. For the material model determination, non-destructive laboratory testing and its FE simulation was used. A new, realistic way of load application is presented along with a detailed investigation of stress distribution in the load-bearing components of the prosthesis. A novel approach for numerical and experimental agreement determination was introduced. This showed differences in the strain on the pylon between the experimental and the numerical model within 30% for the anteroposterior bending and up to 25% for the compression. The highest von Mises stresses were found on the foot–pylon connecting component at toe off. Peak stress of 216 MPa occurred on the posterior adjusting screw and maximum stress of 156 MPa was found at the neck of the male pyramid.

1. Introduction

The lower-limb prosthesis has come a long way from the days of primitive wooden peg legs to present day electronically controlled prostheses, especially in the last decade, with advancements in anatomy, physiology, material design, computer technology etc. evident in new approaches in prosthetic sockets design, modern prosthetic knee mechanisms, more functional prosthetic feet and advanced manufacturing techniques. As well as an increase in patients' expectations, more and more requirements are placed on the prosthesis, above all on its functionality, reliability and safety.

The prosthesis, in conjunction with an amputee, presents a complex biomechanical system, whose behaviour is influenced by a few factors. In term of the prosthesis, the major factors are prosthesis alignment [1–5], the mechanical properties and alignment of the prosthetic foot [6,7], the length of the prosthesis [8,9] and the weight of the prosthetic components [10]. For the evaluation of prosthetic walking performance, a gait analysis is widely used [11]. Most studies reported in the literature have focused on the assessments of gait kinematics [5,7,8,10], plantar foot pressure [1,3] and contact interface between the residual limb and the prosthetic socket [2,4]. A study of the behaviour of the prosthesis is crucial in the phase of prosthetic component design and of prosthetic alignment.

In connection with the development of modern computational techniques, besides experimental methods, numerical methods have also been increasingly used, especially in the cases where it is necessary to obtain detailed and complex information about the behaviour of the prosthesis or about the interaction between an amputee and the prosthesis. The finite element analysis (FEA) is the most widely used numerical analysis method in lower-limb prosthetics [12].

By means of the FEA, stress and strain distribution in the whole prosthesis can be determined, which is experimentally or analytically difficult. FEA can be used for parametric analysis such as the study of design, material or alignment parameters effect, whereas the prototype or specimen need not be fabricated in contrary to the experimental analysis.

Over the past three decades, FEA has been used to study the interaction between the prosthetic socket and the residual limb. The main goal of these papers was to determine how the socket acts on the residual limb. The stress and the pressure distribu-

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tion in soft tissues over the residual limb at critical phases of the gait cycle were usually obtained. These papers studied the effects of socket and residual limb geometry [13], the material properties of the socket and liners [13–15] and the frictional properties at the interface between the limb and socket [16] with the aim of contributing to the improvement of socket design. The previous FEA models were simplified significantly in terms of geometry, material and boundary conditions [12]. Recently published studies have presented more accurate models including complex geometry models, non-linear material models, interface elements enabling slip-friction conditions [16], automated contact method used in the interface [17], pre-stresses applied on the limb [18] and the effect of inertial loads [19].

In spite of many papers dealing with the FE modelling of socket–limb interaction, only a few have been published about the use of this technique for the study of the behaviour of other lower-limb prosthetic components, especially dealing with prosthetic feet. In Ref. [20], a FE model of the SACH foot was created and then used to address the effect of viscoelastic heel performance as an example of parametric analysis. Another work dealt with the FE modelling and experimental validation of a prosthetic foot to improve footwear testing by computational tests [21]. Papers describing the behaviour of the whole prosthesis have not been published yet. Some approaches can be found in studies analyzing the monolimb, which is a kind of trans-tibial prosthesis having a socket and a shank moulded into one piece of thermoplastic material. In this work, the structural behaviour of the monolimb for different shank designs and stress distribution at the socket–limb interface were studied [22]. In the later works of the same authors, the monolimb was optimized by FE parametric analysis of design factors [23], fatigue life was evaluated based on FEA [24] and experimental fatigue tests were conducted [25].

Previous papers introduced a multidisciplinary approach to FE modelling of lower-limb prosthesis, incorporating gait analysis, reverse engineering, material property testing and structural testing. In the cases where the prosthetic foot was included, ground reaction force (GRF) was found suitable for straight load definition in the FEA. However, in the reported works, it was applied directly to prosthetic foot nodes or on surfaces in spite of a significant geometrical non-linearity of the prosthetic foot–ground interface [20,22]. This fact was reported as the most likely source of error, and incorporation of the prosthetic foot–ground contact and friction elements into the FE model was discussed as possible future work in Ref. [20].

Thus this study was focused on further development of an approach applicable to the trans-tibial prosthesis design and inspection, with the aim to improve the FE boundary conditions and to investigate the stress distribution in load-bearing components of the prosthesis in detail and a new approach to numerical and experimental agreement determination was introduced.

2. Methods

This work dealt with the trans-tibial prosthesis with SureFlex™ foot. In the FEA, a prosthetic socket was not included. Because of the complexity, this study was divided into several phases as shown in Fig. 1. For the assessment of the prosthesis geometric configuration and the load acting on the prosthesis, the foot analysis of an amputee, including ground reaction force (GRF) measurement, motion analysis and strain gauge analysis, was done. The geometry model was created on the basis of information extracted with reverse engineering techniques. For determining the material characteristics of the prosthetic foot components, non-destructive tests and their FE simulations were conducted. In the final phase, the FEA was accomplished and stress and strain distributions were obtained. Agreement between the numerical and the experimental models was assessed by comparison of the FEA and the strain gauges analysis results.

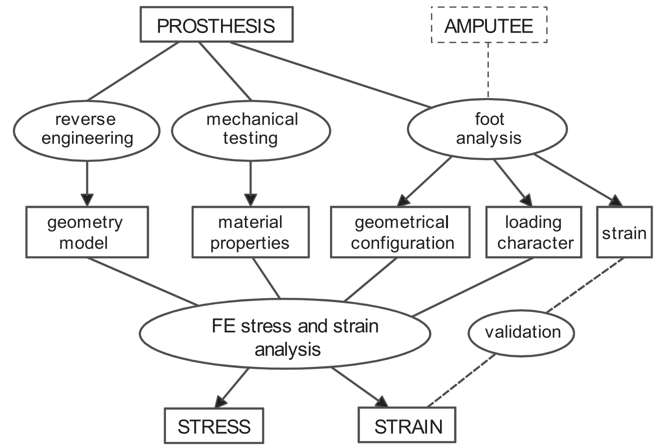

Fig. 1. Flow chart for the approach.

2.1. Foot analysis

The foot analysis was performed on a 32-year-old male right-sided trans-tibial amputee, 80 kg in mass, 182 cm in height, living an active life. His usual prosthetic foot and pylon were replaced by the analyzed ones and was aligned by a prosthetist according to the patient's comments. The patient wore the TSB type socket with SeaFlex (North Sea Plastics) flexible liner and Iceross Seal-in liner (Ossur), providing suction suspension. In the trial, the amputee was asked to walk at a self-selected speed along a 6 m long walkway, while GRF was measured via two AMTI RP 6-5 force platforms placed in the middle of the walkway.

Three-dimensional lower-limb kinematics were collected for the prosthetic side using three cameras with respect to the fixed coordinate system. Passive-reflective markers were placed on the prosthesis at points, corresponding to the following anatomical landmarks; fifth metatarsal head, lateral malleolus, calcaneous and lateral femoral epicondyle. On the basis of the manual markers identification, kinematic data was analyzed using the APAS software so that time-dependent angular displacements of the prosthesis at base planes were obtained. These angles are illustrated in Fig. 7.

For the purpose of measuring the strain acting on the prosthetic pylon, a gauging chain was made up and verified [26]. The gauging chain consisted of a strain gauge system stuck on the prosthetic pylon, a Spider 8—eight data channel gauging card, and a laptop with Beam Spider software and wiring. The strain gauges enabled us to measure linear strain from tension–compression, bending in the medial–lateral (ML) and anterior–posterior (AP) direction and torsion loading. This method provides an accuracy of at least 3%.

A walking trial was repeated four times and the data were averaged with standard deviation up to 10% of the average value for the dominant components of the GRF and strain. Resultant graph displaying GRF, angular displacement and strain acting on the pylon is shown in Fig. 2.

2.2. Geometries

Fig. 3 shows a section model of the prosthetic foot and the foot–pylon connecting components. The geometry model of simply shaped components was created on the basis of manual measurements in 3D CAD software. For the digitalization of the more complex foot cover geometry, an Atos I (GOM International AG) 3D