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Design of an ankle prosthesis for swimming and walking

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Abstract

This study presents a particular mechanical device for lower limb amputees, a prosthetic ankle, designed for swimming and partially for walking in two possible configurations. Experimental static tests are carried out for a compliance according to the standard, in the configuration for walking which is considered the most severe in terms of loading. In parallel, a numerical model is created and validated by a comparison to the experimental test in terms of displacements and strains. The validated FE model is then used for an optimization of the geometry and of the used materials, and to provide a further simulation for the configuration of swimming.

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Keywords: Biomechanics; trans-tibial prosthesis; sport device.

1. Introduction

Lower limbs amputations are physically but also psychologically a hard shock for the patients. For their rehabilitation, the following recovery of the muscle activity and a full reinstatement in a social life, mechanical or more complex robotic devices are thought and designed. The possibility to practice sports is an important note to offer to the amputees, especially for young or more active persons.

In this light, the device presented in this paper is though for being applied to trans-tibial amputees. They are a particular category of lower limbs amputees, still having the possibility of the knee movement. Object of the present study is a mechanical ankle prosthesis to be used basically for swimming (Fig. 1.a), made of different components, shown in Fig. 1.b.

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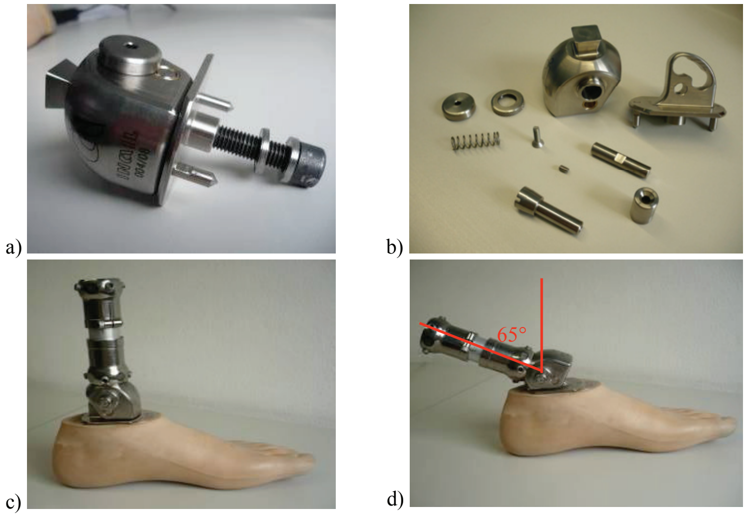


Fig. 1. The ankle prosthesis object of study: full configuration (a) and components (b.). The ankle prosthesis with the foot in walking (c) and swimming configurations (d).

It presents two possible configurations, guaranteed by a relative inclination between the tibia axis and the foot, shown in Fig. 1.c,d. The first configuration (Fig. 1.c) is related to the walking, having the prosthetic foot plane perpendicular to the pipe, used to simulate the tibia. Although the prosthetic ankle is designed for swimming, that is the athlete has to mount this device only at the swimming pool rather than at the seaside or on a boat, this configuration for walking is however needed.

The second configuration (Fig. 1.d), for swimming, has a relative angle of 65°. This angle represents an average position assumed by a foot during the swim. It should be mentioned that this configuration has to be used when the athlete is already in the water.

In general, sport prostheses are designed for specific applications and are not suitable for any other purpose. This prosthesis, on the other hand, is designed to allow the combination of prosthetic feet available in commerce, which are also compatible with common flippers. Hence, the advantage of this ankle device is its versatility: the amputee can move from the first configuration to the other one without changing the prosthesis. Due to its particular application, the ankle is required to be water and seawater compatible, together with a sufficient mechanical strength to hold mechanical loads expected during its use. From these considerations, the manufacturer (INAIL) decided for the use of titanium (Ti6Al4V), because of its peculiarity to create a passivation layer and so to prevent corrosion.

In the following paragraphs, experimental tests carried out according to the available standard and numerical simulations on the prosthesis in the two different configurations are described.

2. Experimental static tests according to ISO 10328

Experimental tests on the ankle-foot prosthesis are carried out in the walking configuration: this is the most severe configuration between the two possibilities. During the short phase of walking, indeed, the load is completely due to the weight of the athlete, who can even wear diving cylinders. In the swimming phase, on the other side, the prosthesis is loaded in hydrostatic configuration, and the global loads are estimated to be less severe.

With the aim of compliance, the standard to be taken into account is the ISO 10328 [1], which sets out the procedures for carrying out experimental tests on all types of lower limb prosthesis for walking (trans-tibial and trans-femoral). The standard identifies two loading configurations at the maximum stress levels undergone by the limb during normal walk (shown in Fig. 2): condition I is related to the instant of maximum loading occurring early in the stance phase of walking (called *heel contact*), while condition II is related to the instant of maximum loading occurring late in the stance phase of walking (called *toe-off*) [2]. ISO 10328 indicates, for each load configuration, three possible load levels depending on physical parameters, locomotion characteristics and other factors of the patient. These three categories are:

– P3: for an amputee till 60 kg of weight;
– P4: for an amputee till 80 kg of weight;
– P5: for an amputee till 90 kg of weight.

The load level chosen for the tests is related to the P5 category, that is the most critical one. For the same reason, the prosthetic foot size (SACH model [3], made of an inner wooden core and a covering in rubber and foam) used for the test is the biggest available in commerce (European size 46), generating the highest bending at the ankle.

In the specific case of the prosthesis object of this study, *principal static proof tests* are required. Due to the particular nature of this device, only static tests are planned, since it is not designed for common walking and a fatigue test in walking configuration is not meaningful. The principal static proof test consists in a fixed load cycle that involves three different types of forces: stabilizing (50N), settling test (1024N for condition I and 920N for condition II) and proof test (2240N for condition I and 2013N for condition II). The increase in load has to be between 100 and 250 N/s.

Fig. 2. Schematization of the stance for the normal walk (a) and related interpretation of the ISO 10328 standard (b).

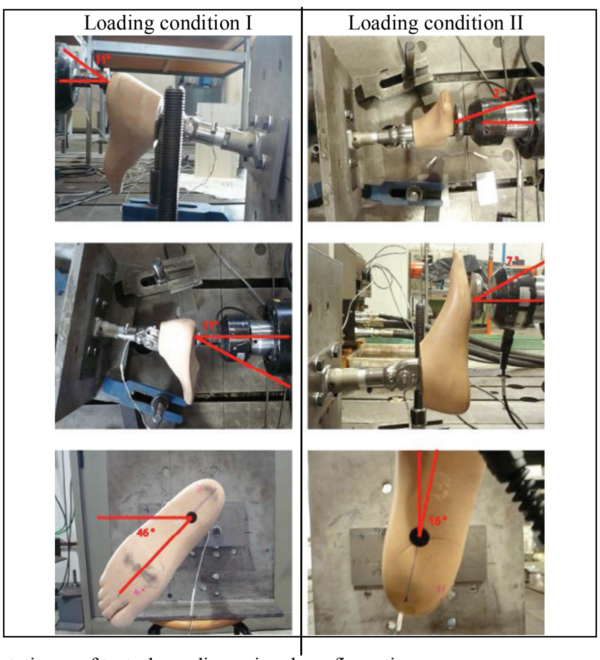


Fig. 3. Experimental principal static proof test: three-dimensional configuration.

The load application is performed by a single hydraulic actuator, because the two load configurations (I and II) can be implemented separately. According to the ISO standard, to perform the principal static proof test, the load line has to be three-dimensional. Thus, to ensure a correct placing of the sample during tests, the prosthesis axis is tilted with respect to the hydraulic piston one. The configuration used for tests is shown in Fig. 3.

During the experimental tests, in load control, piston displacements are recorded in the two different test configurations. Also, a strain gauge was applied to a structural part of the prosthesis where cables are more easily linked: the chosen position is shown in Fig. 4. During experimental tests also these strain values are collected, and will be used in numerical simulation for comparison and validation of the model.

The results of the static tests on the foot prosthesis are shown in Fig. 5 in terms of applied control force (Fig. 5.a) and piston displacements (Fig. 5.b). It is possible to notice that, even if the applied loads are very similar for the two configurations, the resulting displacements are deeply different. In configuration I displacements are almost three times the displacements in configuration II: this is due to the different stiffness of the prosthetic foot, resulting from all its inner materials.

The final result of the static tests was positive, since the residual displacement after the loads application is less than 5 mm, in accordance with the standard requirements. So, a compliance with the ISO standard is reached.

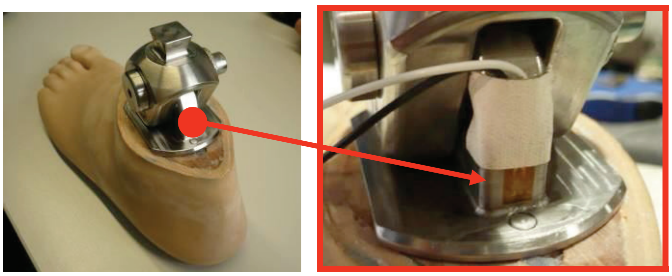


Fig. 4. Strain gauge position on the prosthesis.

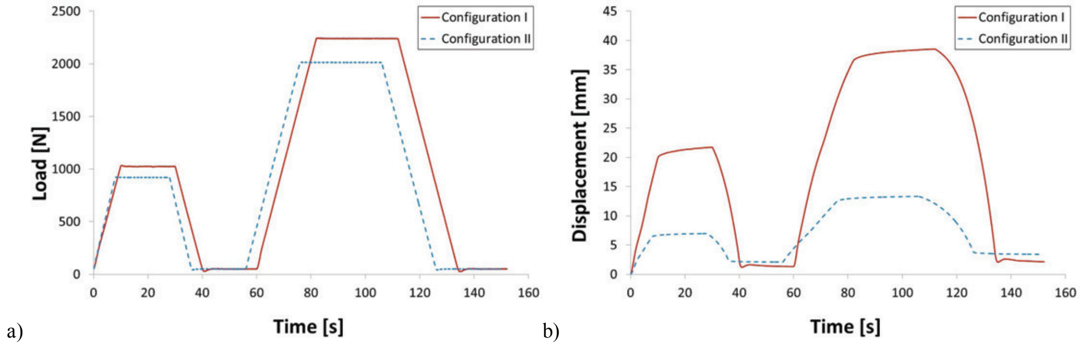


Fig. 5. Experimental result: applied load (a) and resulting displacement (b) trends, from the load cell and the hydraulic actuator.

3. Numerical simulations and optimization of the prosthesis

Aim of these numerical simulations is multiple: first of all, the availability of a validated FE model allows an optimization of the designed part of the prosthesis. In this particular case, the optimization can be performed considering both the materials to be used and the geometry. The two possible points of modification from the original prosthesis are linked, since for both of them the first aim is to reduce the stresses involving the structure and, if it is possible, also the cost of the device. The specific use, which the ankle is designed for, highlights that during the amputee walking it is important to take into consideration the mechanical loading conditions. On the other side, during the amputee swimming, the environmental factors become important. Mechanical and chemical conditions have therefore to be taken into account in the design and compliance of the ankle prosthesis.

A second goal of the numerical simulations is the verification in swimming configuration, not performed experimentally. Using the FE model, it is possible to make a stress analysis and to compare it to the configuration for walking.

A FE model appears therefore a powerful tool in this stage of the work. In order to create and validate the FE model, two steps of numerical simulations are followed: on the foot unit and on the foot and the ankle device.

Aim of the analysis on the foot unit is to obtain an equivalent stiffness of the foot. The geometry of the prosthetic foot is obtained by means of a triangulation 3D laser scanner and the reverse engineering technique. Since the prosthetic foot is made by a wooden part and a rubber foam, the model is divided in two simplified zones: the first one is near the ankle and has the mechanical properties of wood, the second one is the part in contact with the ground. The mechanical answer to different load levels is therefore different since the behaviour is globally non linear. To avoid such simulations, not object of interest for the ankle manufacturer, a linear behavior is used to describe the foot equivalent stiffness: the idea is therefore to deal with the prosthetic foot as a "black box". Equivalent stiffness is found when the displacements are equal to the experimentally measured ones. Calculated stiffness values are 13000 MPa for the upper part (similar to wood), and 2 or 60 MPa for configuration I or II (equivalent to rubber foam). The reason of this wide range of stiffness is related to the different answer of the foot unit during compression tests in the two configurations. A comparison in terms of displacements between experimental test and numerical simulation is shown in Fig. 6 for configuration II.

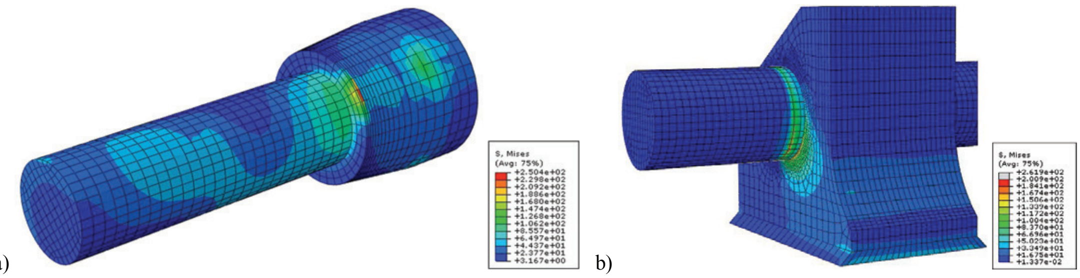
Fig. 6. Comparison between experimental and numerical displacements, for the foot component and *toe-off* loading condition (II) – scale 1:1.

The second type of analysis is related to the assembly of foot and ankle device, joined together by rigid coupling. The ankle is made of different sub-components, all made in titanium. The behaviour of these parts is modelled as linear. In this second type of simulations, comparisons for the models' validation are provided by means of comparisons not only with experimentally measured displacements, but also with strains measured by the applied strain gauge. Some simplifications are made in the model geometry compared with the real ankle, but no one that can substantially modify its behaviour. In Tab. 1 the comparison between the numerical (ε_{num}) and experimental (ε_{exp}) strains is shown. The values of the numerical strains are calculated in correspondence of the strain gages applied to the ankle. The comparison is proposed at two different force levels, for both the loading conditions I and II. It is evident that the percentage differences between the experimental and numerical strain values are very reduced in correspondence of the maximum applied load during the tests.

Results in terms of Von Mises stresses are shown in Fig. 7 for two particular components of the ankle, where stress is found to be the highest. The strength of these parts under the static loads is ensured.

Table 1. Comparison between experimental and numerical results in terms of strains.

Loading condition	**Applied force [N]**	ε_{exp} [με]	ε_{num} [με]	**Diff. [%]**
I	*1024*	-96	-106	-10.4
I	*2240*	-120	-121	-0.8
II	*920*	129	141	-9.3
II	*2013*	257	258	-0.4



a) b)

Fig. 7. Stress field in two components of the ankle device: Von Mises stress.

After the validation of the numerical models, these become a reliable and secure tool to perform optimizations on the designed device. In particular, considerations can be proposed by varying the geometry and materials of the ankle prosthesis designed by INAIL, called model 1.

An optimized geometry is studied (model 2), paying attention to the region of maximum Von Mises stress. In particular, a modification of a fillet radius deeply changes the stress value, as indicated in Tab. 2. In this table, the maximum Von Mises stress is compared to the ultimate tensile stress (UTS), and the safety factor is calculated as their ratio. In model 2 the maximum stress is almost an half with respect to model 1. This consideration gives space to the use of different cheaper materials, instead of titanium. For instance, three other materials, resistant to the corrosion, are considered: an inox austenitic steel (AISI310), an Aluminium alloy (Al7075) and an inox duplex steel (1.4462 or X2CrNiMoN22-5-3). As indicated by the safety factor in Tab. 2 for each analysed material, the choice of Al alloy resulted too restrictive, while it is possible the use of the two inox steels, as it is possible to see from the calculated safety factors.

Table 2. Comparisons of the maximum Von Mises stress in the ankle from numerical simulations – loading condition II. Model 1: as designed by INAIL, model 2: optimized geometry.

Numerical model	**Max VM stress [MPa]**	**Diff. [%]**	**UTS [MPa]**	**Safety factor**
1_Ti6Al4V	535.1	-	920	1.72
2_Ti6Al4V	301.9	-43.6	920	3.09
2_AISI310	223.4	-58.3	500	2.24
2_Al7075	312.3	-41.6	370	1.18
2_Duplex	315.4	-41.1	450	1.43

A last step for the numerical analysis is the comparison between the two configurations for which the ankle prosthesis is designed, for walking and swimming. No experimental tests are performed in the swimming configuration, but only numerical analyses, considering the longest flipper commercially available and two configurations (kick up and down). Fig. 8 shows the most stressed component, in the kick up and down loading. The maximum Von Mises stress is 105.6 MPa, much lower than the walking configuration for the different models reported in Tab. 2. This is a further proof than the compliance to standard must be checked in the walking configuration for the presented prosthesis [4].

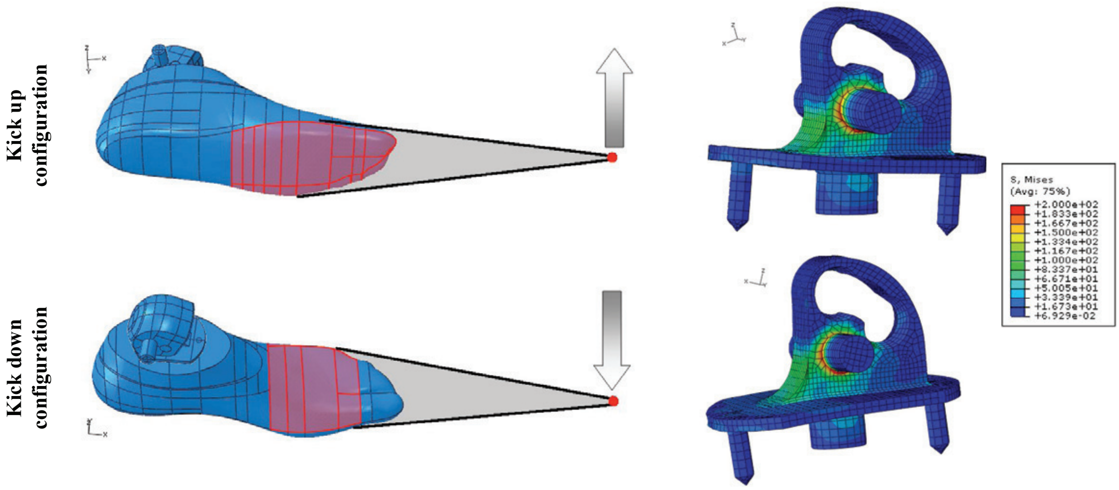


Fig. 8. Swimming configurations: kick up and down, and Von Mises stress in the most loaded component of the prosthesis.

4. Conclusion

In this work experimental and numerical simulations on an ankle prosthesis for walking and swimming were presented. Compliance with ISO 10328 standard has been positively verified by experimental tests. In parallel, a numerical model was developed and validated by the experimental measures of displacements and strains. The numerical model was used for optimization of the geometry and the used materials, and also for a further verification in the swimming configuration. In all these analyses, the device has presented adequate static strength to the loads required by the standard.

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