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MODELING AND ANALYSIS OF AN ANKLE-FOOT ORTHOSIS (AFO) USING MULTIBODY METHODOLOGIES

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ABSTRACT: *Biomechanics is the scientific domain which deals with the study of biological systems, such as the human body, using physical concepts and mechanical engineering methodologies. It allows the development of new medical devices and provides a quantitative analysis of the subject being studied. In the present work, the effect of an ankle foot orthosis (AFO) was studied on a healthy male subject. For this purpose, a biomechanical multibody 2D-model was developed in code MOBILE. The model was made of 9 rigid bodies connected by 9 frictionless hinged joints. Three additional degrees-of- freedom (DOFs) were added so the model can move freely in the plane. Kinematic data acquired in a gait lab were used as time functions to drive the joints and a foot model was designed based on three Hunt-Crossley's spheres-plane contact model. The measured ankle kinematics was successfully reproduced using forward dynamics principles, for the stance phase period. In a first approach, barefoot kinematics was reproduced to define the foot model properties by adjusting manually the foot parameters and fitting the ankle angle. The ankle moment obtained in the gait lab was used to power the ankle joint. Then, the ankle-foot orthosis was added as a linear torsional spring element acting at the ankle joint and the moment powering the ankle joint was diminished. A manual optimization process was performed in order to fit the ankle ankle and it was concluded that the AFO reduces the muscle moment developed at the ankle in 15% and it can be simulated as a spring with k = 50 N.m/rad.*

1 INTRODUCTION

Biomechanics of the human musculoskeletal system is an interdisciplinary field with many applications in the physical and biological study of the human movement [1]. In this context, a number of pathologies due to neurological, vascular or orthopedic disorders affects the ankle motion causing an abnormal gait. Frequently, these disorders lead to lack of control of the lower limbs, and a necessity for lowering the pressure on the feet or need for support [2]. In particular, a large number of patients suffers from dropfoot which is a neuromuscular disorder characterized by a steppage gait that affects the patient's ability to raise their foot at the ankle, and is further characterized by an excessive and uncontrolled plantarflexion, an inability to point the toes towards the body (dorsiflexion) or move the foot at the ankle inwards or outwards. The dropfoot motion leads to toe dragging during the swing phase of the gait cycle and results in pain and weakness. Moreover, numbness may accompany loss of function [3, 4]. A suitable solution for these patients is to be fitted with an AFO, a brace or a splint to provide stability to the ankle-foot complex, restore normal motion or to constrain and inhibit abnormal motion [3, 4]. AFOs can be used to improve the base of support of patients with balance perturbation but they also improve ankle kinematics during stance phase, increase step and stride length, decrease cadence and energy costs in walking, while improving walking, running and jumping skills [5-7].

Despite the great variety of published work on modeling the normal and the pathological gait, regarding muscle activation [8] and trying to understand pain [9], the characterization of the mechanical properties of AFOs [10-13] and their effect on the pathological gait [5, 14, 15], there are only a few articles about modeling the human gait with an AFO and the effect of this medical device on the human gait. Normally, a 2D-multibody system is developed in the sagittal plane and the joints are either rheonomic or scleronomic according to the authors' objectives [2, 12, 16, 17]. However, there is a remarkable lack of knowledge on how AFOs affect the muscular moments [18].

2 BIOMECHANICAL MODEL

In this research work, a biomechanical multibody model of the human body was developed using the object oriented multibody simulation library MOBILE [19]. The model (see Fig. 1) is confined to the sagittal plane, since during the gait cycle, the greater range of motion occurs in this plane [2]. It consists of a pair of legs, each one made up of 4 rigid links (thigh, shank, foot and toes) and an additional body to represent the head, arms and trunk (HAT) inertia properties. The bodies are constrained by ideal revolute joints simulating the hip, knee, ankle and metatarsal articulations (angles φ_h^R, φ_{kn}^R, φ_{an}^R and φ_{mt}^R for the right leg and φ_h^L, φ_{kn}^L, $\varphi_{an}^L$and φ_{mt}^L for the left leg), as well as the bending of HAT with respect to the pelvis (angle φ_H). Three additional DOFs were added at the pelvis, such the model can move freely in the plane (two translations x_p, y_p and one rotation φ_p). Thus, the model has a total of 12 DOFs. It is prepared for gait analysis and kinematic measurements can be used to drive the joints with time functions.

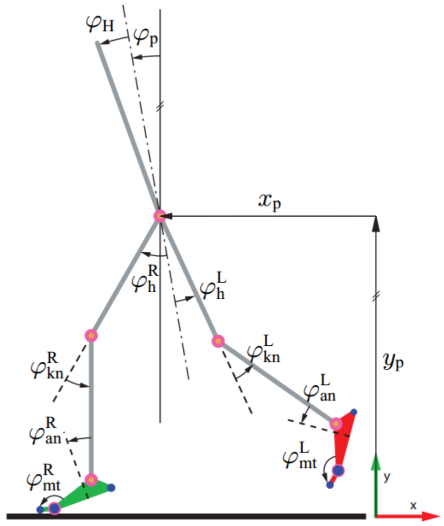

Fig. 1 Graphical representation of the two-dimensional multibody model.

Link lengths were obtained from biometric data and masses, location of the centers of mass and moments of inertia were calculated using the Winter's coefficients table [1] (Table 1).

Table 1 Properties of the rigid links

Rigid link	Length (mm)	Location of the center of mass* (mm)	Mass (kg)	Moment of inertia in the plane (kg/m^2)
HAT	556.71	348.50	44.070	3.36017
Right thigh	416.79	180.47	6.500	0.11780
Right shank	435.66	188.64	3.0225	0.05232
Right foot**	141.00	70.50	0.845	0.00379
Right toes	50.31	21.78	0.0975	0.00***
Left thigh	409.72	177.41	6.500	0.11384
Left shank	429.07	214.54	3.0225	0.05075
Left foot**	141.00	70.50	0.845	0.00379
Left toes	44.56	22.28	0.0975	0.00***

* These distances are relative to the proximal extremity, i. e., the extremity that is closer to the human body center of mass.

** This foot segment refers to the distance between the ankle and the metatarsal joints.

*** The toes moments of inertia were set to zero because it is a small segment with a very small mass.

3 Data Acquisition

Kinematic data were acquired from a healthy male subject with 65 kg of weight and 1.82 m height. The non-pathological movement was acquired by a VICON® MX 13 motion capture system with seven cameras, 2 force plates (AMTI® OR6-7-2000), and 2 camera recorders. The sampling rate was 100 Hz and the measurements were performed with the subject barefoot and using AFOs on both feet.

4 Foot Model

Since the early beginnings of forward dynamics human gait simulations, foot models have been a major concern for biomechanical engineers. The contact force computation is usually computationally heavy and a simple, yet accurate, foot contact model is very desirable [20].

4.1 Existing Models

Several foot models using different geometries and contact and friction properties have been employed. The most representative are described below.

Millard et al. (2008) proposed two simple contact models based on sphere-plane contacts [20]. The first model consists of only one rigid body with two spheres at the heel and the metatarsal joint. The second one is composed of two rigid bodies connected by a revolute joint with a torsional spring-damper system representing the metatarsal articulation. The supplementary body was added to improve the normal ground force profile and has a sphere at the toe tip. During simulations, normal forces were computed using the Hunt-Crossley point contact model [21]. Friction forces were initially computed using a Coulomb friction model which was replaced by a bristle friction model [20].

Moreira et al. (2009) presented a 3D-foot contact model made of two rigid segments, connected by a revolute joint with a torsional spring-damper system [22]. The model included a total of 9 spheres (6 at the plantar surface and 3 at the toes). The normal forces were computed using the Hunt-Crossley model and the friction model included the standard Coulomb friction and a viscous friction component. This model showed promising dynamics results and proved to be appropriate for simulation purposes.

As most of the foot-contact models are based on sphere-plane approaches, lately, Kecskeméthy (2011) presented an alternative [23]. This model is made of two bodies connected by a revolute joint with a spring-damper element and used two cylinder-plane contact elements for the forefoot and the heel contact. Normal forces are computed using the Hunt-Crossley model and the tangential forces (sticking and sliding) were computed using the Coulomb's law of friction.

3.2 Contact Model employed and Forces Computation

The contact model developed in the present work uses a set of three sphere-plane interactions similar to Millard's model. The contact model used between the feet and the floor is the Hunt-Crossley's contact model with a sphere-plane geometry (see Fig. 2) [21]. A spring-damper was added at the metatarsal joints in order to provide more adaptability to the feet. In this way, the feet have a greater freedom of movement than when they were guided by time-functions and simulations showed the model was more stable.

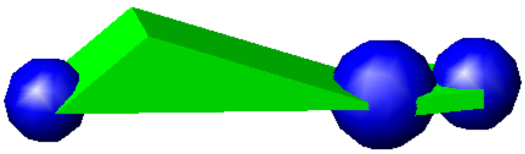

Fig. 2 Foot geometry with identification of the spheres position used in the contact model

The foot geometry was created based on the markers position used during the data acquisition in the gait lab. The radius of each sphere was defined as the minimum height reached by the corresponding marker during the measurements.

The model computes the normal forces (F_N) using the well-known Hunt and Crossley contact model [21], and the tangential friction sticking forces ($F_{T,st}$) were computed by adapting the Hunt-Crossley mode. The transition from sticking to sliding is characterized a vanishing friction saturation ΔF_T (Eq. 1), where the term μ_{st} represents the coefficient of sticking friction. The sliding friction formula ($F_{T,sl}$), were computed according to equations (2) where μ_{sl} represents the coefficient of static friction.

$$\Delta F_T = \mu_{st} F_N - |F_{T,st}| \tag{1}$$

$$F_{T,sl} = \mu_{sl} F_N \tag{2}$$

4 SIMULATION PROCEDURE

The main goal of this research work was to reproduce, by forward dynamics, the ankle motion during stance phase with and without AFO, in order to define the orthoses properties. The simulation was first performed for the barefoot condition to test the foot model and determine the contact properties.

In this approach, the pelvis pose, x_p, y_p and φ_p, both metatarsal joint angles φ_{mt}^R and φ_{mt}^L, and the ankle joint angle of the stance foot φ_{an}^R are set as scleronomic variables (i.e., treated as generalized coordinates moving under the effect of forces according to the equations of motion) while all other variables are driven with time functions obtained in the gait lab. Furthermore, a spring-damper system is applied to the metatarsal joints (with angles φ_{mt}^R and φ_{mt}^L) and the ankle joint of the stance foot (φ_{an}^R) is powered by the ankle moment obtained in the gait lab. Thus, the problem corresponds to a multibody system with six degrees of freedom (x_p, y_p, φ_p, φ_{mt}^R, φ_{mt}^L and φ_{an}^R) subject to six rheonomic constraints (φ_h^R, φ_{kn}^R, φ_h^L, φ_{kn}^L, φ_{an}^L and φ_H).

4.1 BAREFOOT SITUATION

Contact parameters, as well as metatarsal spring-damper properties were manually tuned in order to fit the ankle angle obtained in the gait lab. Fig. 3 shows the right ankle angle plotted against time for (a) a barefoot measured gait (blue dotted line) and (b) a forward dynamics simulation of the gait (red continuous line) using the set of parameters presented in Table 2 and 3.

Despite the good correlation between the two curves, there is a discrepancy in the first 0.05s which represents the time that the simulation takes to stabilize after heel strike (HS). This discrepancy is due to the high forces developed at HS. From this instant on, the plot coincide during most of the stance phase. During pre-swing (starting at *t=0.5s*), a small offset is noticed and toe off (TO) is reached 0.04s sooner and with and an angle 2.3 degrees small than the measured gait which can be explained by the simplicity of the foot model, mainly in the forefoot area.

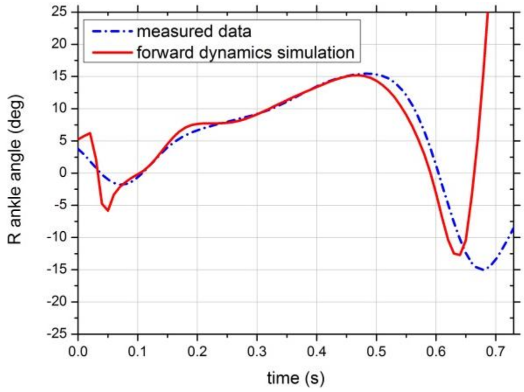

Fig. 3 Ankle kinematics (a) measured (blue dotted line) and (b) obtained by forward dynamics (red line). Results refer to the barefoot trial.

Table 2 Contact parameters of the foot

Contact parameters	Heel sphere	Metatarsal and toe tip spheres
Normal stiffness c_N (N.m$^{-1.5}$)	6.5x10^5	1.0x10^9
Tangential stiffness c_T (N.m$^{-1.5}$)	6.5x10^5	1.0 x10^9
Normal coef. of restitution e_N	0.4	0.2
Tangential coef. of restitution e_T	0.01	0.1
Exponent in the force-deflection function	1.5	1.5
Static/sticking friction coef. μ_{st}	0.5	0.9
Dynamic/static friction coef. μ_{sl}	0.4	0.8

Table 3 Metatarsal spring-damper system parameters

Contact parameters	Heel sphere
Spring constant k	22.5 N.m/rad
Damper constant c	0.0 N.s/m

4.2 ORTHOSES SITUATION

In a second step, the set of parameters obtained in section 4.1 was used to compute the forward dynamics of the gait with AFOs. Hereby, the ankle moment was modified by a factor α which represents the percentage of muscular activation needed to maintain a normal gait using an AFO, comparing to the barefoot case. Moreover, an extra linear torsional spring with stiffness k was applied at the ankle simulating the AFO's stiffness.

The foot parameters (Table 2 and 3) were maintained since the contact properties are the same whether a patient is wearing an orthosis or not.

Orthosis spring-damper properties and coefficient α were tuned for the best ankle angle fit which is represented by the red continuous line in Fig. 4. The corresponding set of parameters is presented in Table 4 and Fig. 5, which shows the moment applied at the stance ankle joint.

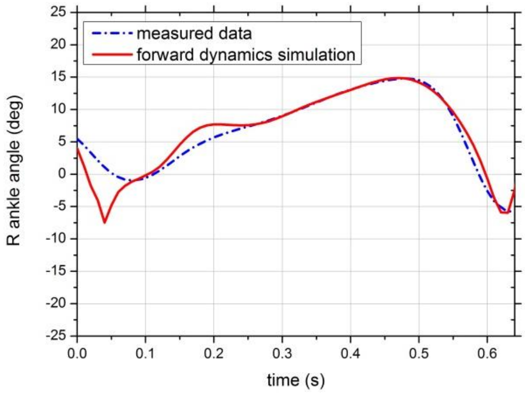

Fig. 4 Ankle kinematics (a) measured (blue dotted line) and (b) obtained by forward dynamics (red line). Results refer to the trial with AFO.

Again, the initial instants does not give a good approximation because of the HS impact and the simulation takes a little time (aprox. 0.1 s) to stabilize. Nevertheless, the remaining parts of the curve are very similar to the curve obtained in the measured gait, even until TO (*t=0.63s*).

Table 4 Spring-damper system parameters of the AFO

Contact parameters	Heel sphere
Spring constant k	50.0 N.m/rad
Damper constant c	0.0 N.s/m

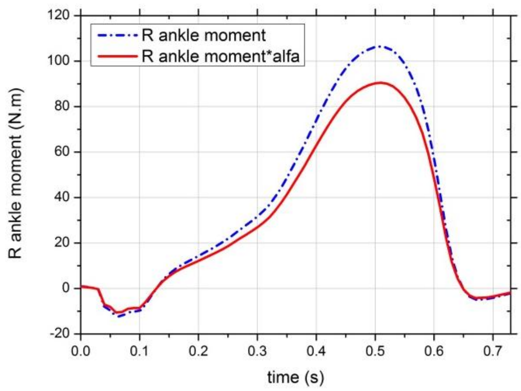

Fig. 5 Moment applied at the ankle for the (a) barefoot simulation (blue dotted line) and (b) the simulation with AFO with α = 85% (red continuous line).

Fig.5 presents the ankle moment applied at the ankle joint over time for (a) a barefoot measured gait (blue dotted line) and (b) a forward dynamics simulation with a spring simulating the AFO (red continuous line). It demonstrates that using orthoses in both feet, the muscle activation is only 85% of its value barefoot, thus demonstrating the support of the orthoses which is about 15% of the total moment developed at the ankle.

5 CONCLUSIONS

In the present work, a biomechanical model 2D-model was developed in MOBILE. The model was created in the sagittal plane and is made by 9 rigid bodies constrained by 9 frictionless revolute joints and has 3 additional DOFs added at the hip.

A simple contact model was developed in the foot-floor interface with three sphere-plane contacts. The model is prepared to solve forward dynamics problems and–the ankle kinematics obtained in the gait lab was positively reproduced, with and without orthosis. Thus, the model was validated and proved to be appropriate for this study, since it is efficient and simple to define.

AFOs can be successfully computed as torsional springs acting on the ankle.

Forward dynamics simulation proved that the spring constant is about *k = 50 N.m/rad* and that it reduces the muscle activation at the ankle in about 15% in a healthy subject.

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