Two approaches to estimate foot-ground contact model parameters using optimization techniques

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Abstract

In the last decades, there has been a growing interest in understanding the dynamics of human gait. Usually, the procedure starts in gait analysis labs capturing the subject's motion by means of an optical system and measuring the resultant foot-ground contact force and torque through force plates. After some data processing to get consistent kinematic input data, multibody dynamics techniques are used in order to obtain the equations of motion of the biomechanical system. These are set in a forward or inverse way depending on the goal of the study. The inverse dynamic analysis (IDA) is used to calculate the net joint forces and driving torques that the musculoskeletal system produces during human locomotion. Conversely, the forward dynamic analysis (FDA) determines how the system will move due to the effect of external and internal forces and, hence, it is a predictive method.

If IDA is computed using only kinematic information and anthropometric body segment parameters (BSP), the double support phase becomes undetermined. Generally, this indeterminacy is overcome by the measurement of ground reactions by means of force plates. However, in some cases these devices are not available and, in these conditions, constitutive foot-ground contact models representing the forces as a function of the system state are needed to solve the problem. When the aim of the analysis is to predict motion via FDA, the foot-ground contact model is also needed to simulate the interaction between the biomechanical system and the environment, i.e. the ground. Therefore, in both cases a foot-ground contact model will be necessary.

Some foot-ground contact models have been presented in the literature. They are based on sphere-plane contact elements [7,3], cylinder-plane contact elements [5], or a set of springs and dampers [4]. They are all used to reproduce the relationship between the contact forces (normal and tangential) and the relative foot-ground displacements and velocities. The main problem is how to determine all the parameters of the contact model (i.e., stiffness, damping and friction coefficients, and geometrical properties of the contact elements), and how to relate their values with the actual contact phenomenon.

This work presents a comparison of two approaches used to estimate the parameters of a foot-ground contact model based on sphere-plane contact elements. Both approaches use the optimization method known as Covariance Matrix Adaptation Evolution Strategy (CMA-ES). The objective of the first approach is to adjust the position and size of the spheres and the contact model parameters, so that the resultant foot-ground torsor (force and torque provided by the inverse dynamics) is reproduced by the contact model, that is, minimizing the RMS error between the two torsors using BSP and the motion of both feet as input data.

The aim of the second approach is to adjust the same design variables, so that the predicted motion provided by a forward dynamics optimization can be as faithful as possible to the captured motion. In this second case, the ankle joint torsor (calculated from the same motion capture and a force-based approach that uses force plate data [1]) and BSP are used as input data of the forward analysis, and the objective function involves minimizing the RMS error due to the discrepancies between the captured and the predicted feet motion. Consequently, the information coming from one motion capture is used in two different ways and the same foot-ground contact model parameters are determined for each case.

The foot sole surface is approximated by four spheres (Figure 1) and a continuous contact force model is used. Each foot is divided in two segments with three sphere contact elements for the hindfoot and another sphere for the forefoot contact. The geometric characteristics of the spheres are considered design variables of the optimization process. The relative rotation between the metatarsal bones and the proximal phalanges is modeled through a spherical joint.

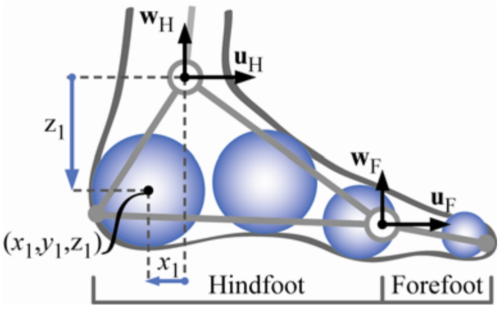

Figure 1: Projection on the sagittal plane of the 3D foot model used.

The normal component of the contact force is estimated using the model proposed by Lankarani and Nikravesh [6], while the tangential forces are calculated through the model proposed by Dopico *et al.* [2]. As mentioned before, the parameters of the contact model in both feet are also considered as design parameters of the optimization process.
As a result, the design variables of the described compliant contact model are the following nine parameters for each sphere: the x and y local position of the sphere center for a given z (in the corresponding local coordinates system [$\mathbf{u}_i$, $\mathbf{v}_i$, $\mathbf{w}_i$]), the sphere radius r, the stiffness K and restitution coefficient c_e of the Lankarani-Nikravesh normal force model [6] and, for the tangential force model [2], the friction coefficient μ, the parameter ν that accounts for the velocity of the stick-slip transition, and the stiffness and damping coefficients (k_s and c_s) of the stiction model.
The human motion data is acquired using 12 OptiTrack FLEX:V100R2 cameras sampling at 100Hz and its software which provides the 3D trajectories of the 36 passive markers attached to the human body. The foot-ground contact torsor (force and torque) is measured using two AMTI AccuGait force plates embedded on a walkway. A 3D computational model of the subject has been developed in mixed (natural and relative) coordinates: it is composed by 18 bodies connected by spherical joints, has 57 degrees of freedom and is defined by 228 dependent coordinates [1].
Since the feet motions are known and the forces introduced by the contact model depend on kinematic data, material properties and geometrical information, only a partial human body model composed by the two feet is required in the optimization process aimed to characterize the contact parameters. The kinematic information (position and orientation of each foot segment) and the dynamic data (ankle and foot-ground torsors) are previously obtained from the whole human body model. The use of this partial model instead of the entire model reduces considerably the computing time required by the optimization procedure.
Finally, the residual value of the two objective functions (RMS error) is compared and the range of variation of the 72 design variables is contrasted. The obtained results provide criteria to assess when to apply one approach or the other in order to estimate foot-ground contact model parameters.

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