Journal of Biomechanics 42 (2009) 1746–1753

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Comparative roll-over analysis of prosthetic feet

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ARTICLE INFO

Article history:
Accepted 3 April 2009

Keywords:
Prosthesis
Foot
Material testing
Inverted pendulum model
Curvature

ABSTRACT

A prosthetic foot is a key element of a prosthetic leg, literally forming the basis for a stable and efficient amputee gait. We determined the roll-over characteristics of a broad range of prosthetic feet and examined the effect of a variety of shoes on these characteristics. The body weight of a person acting on a prosthetic foot during roll-over was emulated by means of an inverted pendulum-like apparatus. Parameters measured were the effective radius of curvature, the forward travel of the center of pressure, and the instantaneous radius of curvature of the prosthetic feet. Finally, we discuss how these parameters relate to amputee gait.

1. Introduction

Given the design of a prosthetic foot, how does it match the function of a biological foot? In human plantigrade gait the foot rolls over the ground during each step, analogous to a wheel (Hansen et al., 2004a), while the stance leg acts like an inverted pendulum (e.g., Winter, 1995a,b). The center of mass (CoM), located roughly at the level of the pelvis, travels in a series of arcs (Fig. 1A). The overall motion can be described by a rocker-based inverted pendulum model, which, in contrast with a simple inverted pendulum with a peg, allows the center of pressure (CoP) to travel forward along the curved foot (Fig. 1A–D). It has been shown that able-bodied people have curvatures of about 30% of the leg length during walking (McGeer, 1990; Hansen et al., 2004a). The energy cost of walking is lower on long, smoothly curved feet than on flat or pointy feet. It has been reported that the metabolic costs are optimal when walking on a foot with a curvature of 30% of the leg length (Adamczyk et al., 2006). Moreover, foot curvature has been demonstrated to be surprisingly invariant to walking speed, loads carried, and shoe heel height (Hansen et al., 2004a; Hansen and Childress, 2005, 2004). On the other hand, considerable differences were found between gait initiation, steady-state walking, and gait termination (Miff et al., 2008). Here, the orientation of the curvature changed from 'flexed' to 'neutral' to 'extended' (Fig. 1C) through active muscle control.

The major challenge when designing prosthetic feet is to substitute the actions of the biological counterpart as efficiently as possible by means of a passive dynamic device. In this endeavor the design of prosthetic feet has become increasingly sophisticated in recent years. From this development arises an immediate need for a standard test method not only quantifying the mechanical properties of prosthetic feet, but further allowing a seamless translation of mechanical test data to performance data of gait experiments in individual patients.

A wide range of mechanical properties of prosthetic feet have been studied, such as stiffness (Van Jaarsveld et al., 1990; Lehmann et al., 1993), natural frequency (Lehmann et al., 1993), energy recovery and hysteresis (Van Jaarsveld et al., 1990; Postema et al., 1997; Geil et al., 2000), viscoelasticity (Geil, 2002), and material fatigue (Toh et al., 1993). While these studies have addressed important properties of prosthetic feet, the resulting qualitative characteristics have often lacked a seamless translation into amputee gait characteristics due to separate measurement techniques. Furthermore, the nature of loading applied during testing is very different to that in walking. During testing the loading was increased at a constant ground angle, while during actual gait the ground angle changes during roll-over. In a similar method, introduced by Hansen et al. (2000), the complex actions of the ankle–foot system are captured in an overall motion. Through quasi-static loading with a custom-made jig the authors estimated the roll-over shape, i.e. the effective

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doi:10.1016/j.jbiomech.2009.04.009

Nomenclature

CoM	center of mass
CoP	center of pressure
F	force (N)
GRF	ground reaction force (N)
l	pendulum length (m)
m	mass (kg)
s	forward travel (mm)
s'	instantaneous radius of curvature (mm), i.e., first derivative of forward travel (s)
x, y	x-axis forward, y-axis vertically upward
x_0	center of curvature (mm)
α	shank angle (deg.)
ρ	radius of curvature (mm)

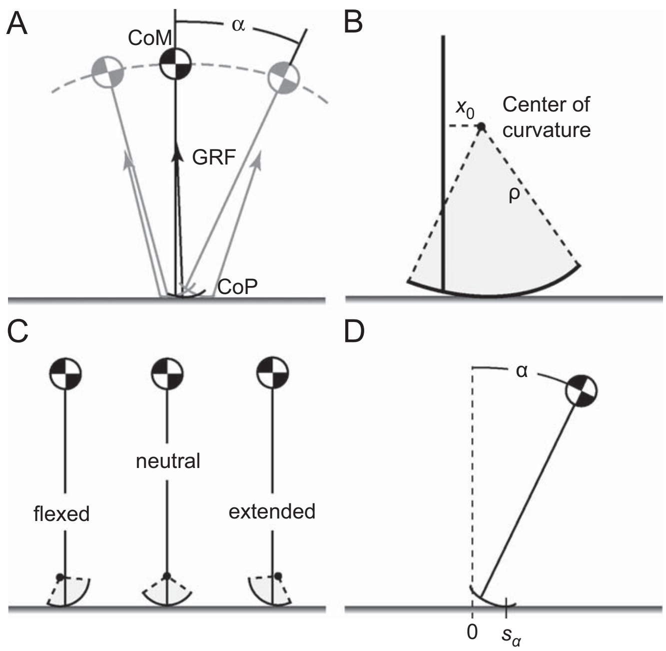

Fig. 1. Rocker-based inverted pendulum model. (A) Model of the leg as inverted pendulum supporting the center of mass (CoM). A ground reaction force (GRF) acts on the CoM, originating at the center of pressure (CoP). (B) Roll-over characteristics of a foot approximated by a constant radius of curvature (ρ), and fixed horizontal position of the center of the curvature (x_0). (C) The position of the rocker center (x_0) affects the orientation of the foot curvature. Able-bodied people actively manipulate the position of the effective rocker center by means of muscle control. A posterior shift of the center of curvature results in a more 'flexed' orientation; this can be observed in able-bodied people during gait initiation. During steady-state walking the curvature maintains a 'neutral' position. An 'extended' orientation, i.e. an anterior shift of the center of curvature, is produced during the last step of gait termination. (D) General foot model with a variable curvature; $s(\alpha)$ is the location of the CoP on the ground as a function of shank angle α. The zero position is defined as the horizontal position of the shank at $\alpha = 0$. The instantaneous radius of curvature (s') is the first derivative of the forward travel (s) with respect to the shank angle (α).

curvature of prosthetic feet. The loading apparatus was modified in a later study (Hansen et al., 2004b); here a small weight was attached to the prosthetic foot and rolled over while a technician applied additional loading. This method has practical limitations with respect to the maximal load that can be applied. In addition, a rather short pendulum length was used so that the moments acting on the foot may be different to those during amputee gait; this might affect the effective roll-over shape. In this study we extend the work of Hansen et al. (2000, 2004b) by introducing a novel inverted pendulum-like apparatus allowing continuous measurements. The technical requirements for the device testing method presented here are basically the same as those for a standard gait analysis, making intricate custom-made loading jigs and other costly solutions redundant.

The purpose of this study is to determine the roll-over characteristics of a range of prosthetic feet and to discuss their biomechanical implications for prosthetic gait. Further, we will quantify the effect of different types of shoes on these characteristics. We expect the differences between the prosthetic foot models to be stronger than the effects imposed by the various shoes.

2. Methods

We designed an experiment in which we investigated the roll-over characteristics of a number of prosthetic feet in combination with different shoes by means of an inverted pendulum-like apparatus. Roll-over shapes of these foot–shoe combinations were simulated for a hypothetical subject with a body mass of 70 kg and a body height of 1.80 m.

2.1. Apparatus

The inverted pendulum-like apparatus consisted of a shaft, with a prosthetic foot attached to the lower end and a mass (m) of 70 kg mounted to the upper end of the shaft (Fig. 2). The pendulum length (l), the distance between the foot sole and the CoM of the added weight, was 0.98 m, a typical leg length for a person of 1.80 m body height. A custom-made rig provided lateral guidance during testing with a minimum of friction; further it restricted the anterior–posterior range to predefined limits.

Seven prosthetic feet of three different manufacturers were included in this study (Table 1): Endolite (Esprit, Navigator), Össur (Flexfoot, Vari-Flex), and Otto Bock (1C40, 1D10, 1D35). All feet were right sided and sized 270 mm. Each prosthetic foot was tested under 4 different conditions: (1) without shoe, (2) with a men's leather shoe, (3) with a running shoe, and (4) with a hiking boot.

The prosthetic feet were mounted in neutral alignment, i.e. the feet were flat on the ground when the shaft was vertical under zero-load. The alignment was adjusted for shoe heel height to obtain a vertical standing position of the shaft under zero-load.

The ground reaction forces (GRF) and the position of the center of pressure were measured with an AMTI (a) force plate, sampled at 1000 Hz. Two reflective markers were placed on the shaft. The markers were tracked by an eight-camera VICON motion analysis system (b) at a sampling frequency of 100 Hz. The data was then further processed with Matlab (c). This delivers, amongst others, shank angle (α) and ankle moment data.

2.2. Experimental procedure

The experimenter applies a horizontal force to the top weight necessary for an angular velocity of ± 10 deg/s, making the foot roll over from heel to toe and back. The measurement range was -15° to 20° with respect to the absolute vertical, corresponding to heel-contact and toe-off. The testing procedure was repeated three times for each foot–shoe combinations. (A video of the experimental procedure is available as Supplementary Material.)

2.3. Data analysis

2.3.1. Effective radius of curvature (ρ)

During part of the stance phase a biological ankle–foot system acts like a smoothly curved solid object. The CoP progresses forward from heel to toe, similar to that of a rolling wheel with a particular radius. Hansen et al. (2004a) proposed a method that allows estimating the effective 'ankle–foot roll-over shape' of an ankle–foot system from CoP data. The strength of this method is its universal applicability, ranging from a simple rolling wheel, to deformable objects and complex multi-joint systems. By transforming successive CoP location data from a laboratory-based coordinate system into a shank-based coordinate system the effective curvature geometry can be determined. The resulting ankle–foot roll-over shape is similar to a circle and reflects the overall motion of the system. The radius