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Effect of different mounting angles of prosthetic feet dedicated to sprinting on reaction forces

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Abstract

Prosthetic sprinting feet made of carbon fibre reinforced plastics for transtibial amputee athletes are widely used by hobby athletes and in professional competition. However, so far work done to assess static and dynamic properties of prosthetic feet dedicated to sprinting did not take into account different mounting angles of the prostheses onto the shaft. In this research two sprint prostheses (Otto Bock Sprinter feet) with low (P1) and mid stiffness (P3), used for athletes of high activity level in Paralympic sports were subjected to compressive loads on a motor driven static test bench under quasi-static loading conditions at different mounting angles (0, 5 and 18 degrees) and vertical and horizontal reaction forces were measured. The energy return did not show unambiguous dependence on mounting angles. The results showed that both vertical and horizontal stiffness decreased as the mounting angle increased, which was unexpected and requires further examination.

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Keywords: prosthesis, running prosthesis, mechanical properties, hysteresis, horizontal forces

1. Introduction

In transtibial amputee sprinting athletes are using special sprinting prostheses which are attached to the shank stumps via an interface (socket) and have a carbon fibre reinforced foot in the form of a blade attached which acts similar to a spring and is compressed during the early stance phase returning the stored energy during the push off phase. However not 100% of the stored energy are returned as it partially dissipates as heat or sound [1].

Carbon sprint feet are manufactured for different weight classes and allow different mounting angles on the socket. Each athlete can adapt the mounting angle - depending on manufacturer and model - in different steps. In the literature, ground reaction forces (GRF) have been assessed with individual athletes [2–6], however, so far only little work has been done to assess static and dynamic properties of prosthetic feet dedicated to sprinting [7,8]. Far more research was conducted on prosthetic feet for everyday life [9–12].

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The results of past research are not generally applicable and do not take into account different mounting angles of the prostheses onto the socket. The properties (i.e. stiffness and hysteresis) of different prosthetic feet in different phases of the gait cycle (i.e. different loading angles) were investigated by [10] but only in feet for daily living.

The literature mentioned above almost exclusively reports one single stiffness value for prostheses (i.e. in vertical direction). However, it seems appropriate to consider the mechanical behaviour in horizontal direction too. Although the mounting angle does not directly influence the foot's mechanical properties, it is considered to highly affect the direction of the ground reaction forces during sprinting. It is expected that (1) a prosthesis for athletes with heavier body weight (BW) has a higher stiffness (k); (2) that the vertical reaction force (and hence vertical stiffness) of all prostheses decreases as the mounting angle increases; (3) the horizontal (forward) reaction force (and hence horizontal stiffness) increases as the mounting angle increases; and that (4) the resulting force (and hence resulting stiffness) acting on a given prosthesis remains unaffected by the mounting angle. Furthermore it is expected that (5) the resulting energy return of the prosthesis is not affected by the mounting angle.

2. Methods and materials

For the experiments, two sprint prostheses of the same model with different stiffness parameters were used for testing. Both samples are *Otto Bock Sprinter* feet (Duderstadt, GER) designed for paralympic sports with low (P1) and mid stiffness (P3), respectively. P1 was used for athletes with low ($\approx 55kg$) and P3 for athletes with mid BW ($\approx 80kg$) and high activity level. Along with these samples, different attachment brackets were used which allowed for different mounting angles between shaft and prosthetic foot. The company offers three brackets with different angles which were used for the measurements: 0°, 5° and 18°.

The static tests were conducted on the static test bench of the University of Applied Sciences Technikum Wien. The position of the prosthesis relative to the ground was kept constant by using an arrester plate on the front tip of the prosthesis, still allowing for different contact points between prostheses and ground, as shown by [7] that a fixed distal end enhances the results. On the surface of the cross beam as well as on the vertical face of the arrester plate, teflon strips were attached to generate a very low friction coefficient and thereby minimise the influence of a possible stick-slip effect. The position of the motor/spindle unit - usually freely moving horizontally - was also fixed to allow for recording of horizontal GRF.

The prosthetic foot was attached to the end of the spindle and aligned vertically using a digital level such that the front tip was almost touching the arrester plate. A 3D force sensor (K3D120, ME-Messsysteme GmbH, Hennigsdorf, GER) was used to record the reaction forces (F_x, F_y, F_z) during loading; positive F_y were facing backwards and positive F_z upwards. Furthermore a laser distance sensor LDS 85/705 (ELTROTEC Sensor GmbH, Uhingen, GER) was positioned below the load cell such that the laser beam was directed on the underside of the load cell, thereby measuring the compression of the prosthesis during the tests. All data were recorded synchronously using a 11-bit A/D converter (NI-DAQ 6008, National Instruments, Austin, USA) at a recording frequency of $100Hz$ with a LabView (National Instruments) application.

Each sample (P1, P3) was tested ten times with each of the available bracket angles (0°, 5° and 18°). For being able to record the data continuously, the speed of the motor/spindle unit was kept low ($2mm/s$) in order to maintain quasi-static conditions. Initially it was intended to load the prosthesis up to a vertical force (F_z) of approximately 3 times the BW of the athletes the prostheses were manufactured for, as these are the magnitude of GRFs reported in literature [2]. However, it was observed that the resulting horizontal forces generated bending moments on the spindle which caused the spindle to stop. As the magnitude of these bending moments depended on the mounting angle, different maximum vertical forces could be applied for different mounting angles. Hence the prostheses were loaded between 1000N and 1400N of F_z (Figure 1). After having reached the maximum test force, an unloading sequence was initiated using the same spindle velocity as for the loading.

The raw data were processed with Matlab R2012a (The Mathworks, Natick, USA). After reading the data from all files and applying calibration parameters from the respective sensors' manuals, the following steps were performed on the raw data for calibration and filtering: the data were filtered using a zero-lag digital forward-reverse moving average filter (window width 50 samples); force offsets were removed by calculating the mean of the first 10 values of all force channels and subtract the mean from the values ($F = F - \frac{\sum_{i=1}^{10} F_i}{10}$). Furthermore all data before and after ground contact of the prosthesis were removed by setting a force threshold ($F_t = 10N$) to remove data from all channels where