Exploring the effect of varying internal prosthetic foot degrees of freedom in musculoskeletal modelling

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INTRODUCTION

Prosthetic feet reach their highest range of motion in the sagittal plane. However, modern passive elastic prosthetic feet also allow for movement in the frontal and transverse plane through adapters or split toe designs [1]. That helps to adapt to uneven surfaces or to sideways movements. Previous research has investigated the biomechanics of transtibial prostheses users through 3D inverse musculoskeletal simulations [2]. However, non-sagittal plane movements within the prosthetic foot model were largely neglected.
The aim of this study is therefore to explore the effect of varying degrees of freedom (DOFs) within the prosthetic foot model on knee and hip joint angles and moments in the prosthetic leg during straight running and during turning sideways while running.

METHODS

Kinematic and kinetic data of overground running was collected from 5 transtibial prostheses users [3]. Participants were instructed to run at self-selected speed and hit a force plate with their prosthetic foot during straight running and whilst performing a 45° running turn towards the sound side. All participants wore the same commercially available passive elastic prosthetic foot, which includes a split toe design, matched to their weight (Blade XT, Blatchford, Basingstoke, UK).

A full-body OpenSim model [4] incorporating a 9-segment prosthetic foot (Fig 1) was scaled to each participant. Inverse kinematics (IK) and inverse dynamics were then performed within OpenSim under two prosthetic foot intersegmental joint conditions: (1) modelled as 1-DOF (sagittal plane) hinge joints; (2) modelled as 3-DOF custom joints.

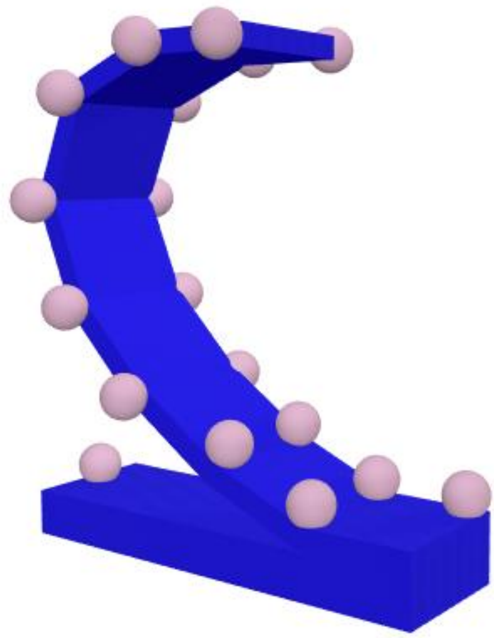

Fig 1: 9-segment prosthetic foot model including model markers.

For both conditions, outputs were investigated during the stance phase. Variation differences in marker errors (IK processes) were compared qualitatively with reference to OpenSim guidelines. Prosthetic-side knee and hip angles and normalised (to body mass) hip and knee joint moment time histories were assessed using a paired t-test in SPM ($\alpha=0.1$; spm1d.org). Whilst normalised peak knee and hip moment differences were analysed via Bland-Altman plots [5].

RESULTS AND DISCUSSION

On visual inspection of the model's movement, resulting from IK, model markers placed on the prosthetic foot matched the experimental markers more closely with the 3-DOF prosthetic foot. That was especially apparent during the 45° running turn.

The IK marker errors were higher in the 1-DOF condition than in the 3-DOF condition. The differences were on average 0.32 (±0.1) mm and 0.62 (±0.3) mm during straight running and the 45° running turn, respectively. These were considered too low to influence the evaluation of the IK processes (threshold for IK evaluation 2cm).

No significant differences were observed in the normalised knee and hip joint moment time histories during straight running and side cutting and peak knee moment differences stayed low.

No significant differences between angle time histories were found across movement tasks with one exception. Knee and hip flexion angle time histories aligned almost perfectly across straight running and 45° running turn (Fig 2). Still, significant differences were found for knee flexion angles during straight running. These differences were however very small (<1°) and therefore considered trivial.

During straight running highest peak differences were found in hip rotation. While no clear trend was obvious, and the mean difference stayed below 5° there was considerable variation per participant with peak differences ranging up to 9°.

During the 45° running turn a pattern was observed. Hip adduction and rotation angles were higher in mid stance for most participants across all trials, with the 3-DOF prosthetic foot, while matching closely during the rest of the movement (Fig 2). Mean peak errors were close to 5°.

The lower IK marker errors in the 3-DOF condition and the hip joint angle differences during the 45° running turn can partly be explained by the ability of the 3-DOF prosthetic foot model to follow the movement of the physical prosthetic foot during sideways loading and partly by modelling constraints. The model's knee joint was constrained to movement in the sagittal plane, the socket-residual limb interface was locked, and soft tissue movement was neglected. Subsequently, the model didn't allow for motion outside the sagittal plane between the hip and the prosthetic foot. The prosthetic foot model's movement and the hip rotation outside of the sagittal plane might have compensated for that and might therefore be exaggerated during mid stance. This relation might become clearer, when making use of a more detailed knee model and/or using a less constrained residual leg-socket interface.

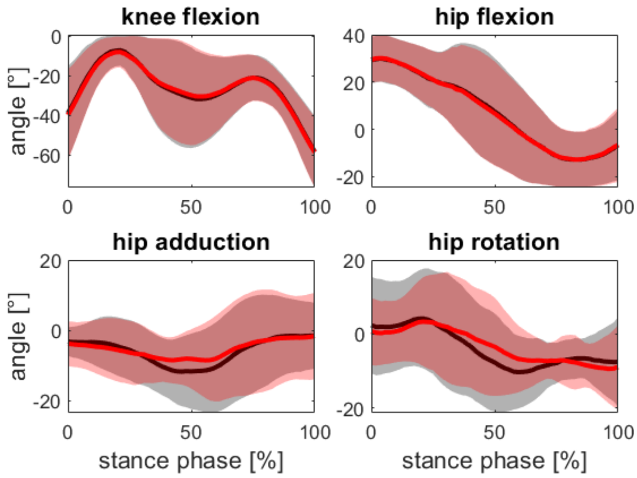

Fig 2: Mean and standard deviation of stance leg joint angles during 45° running turn, in 3-DOF condition (black) and 1-DOF condition (red). Anatomical position is defined as 0°, with positive increase representing extension, abduction, and inward rotation.

CONCLUSIONS

On average, the effect of varying prosthetic foot model DOFs on IK processes, as well as knee and hip joint angles and moments is small. However, the individual variations observed suggest incorporating non-sagittal plane movements within the prosthetic foot during sideways movement should be considered when developing musculoskeletal models to simulate movement and understand individualised cause and effect.

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