EFFICIENT COMPUTATIONAL METHOD FOR ADJUSTING THE STIFFNESS OF INDIVIDUAL 3D-PRINTED INSOLES

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Introduction

Diabetes-adapted insoles are crucial for adequate and consistent distribution of plantar pressure. The International Working Group of Diabetic Foot recommends adjusting the stiffness of the insoles based on the plantar pressure distribution during gait [1]. Most Finite-Element (FE) models that simulate gait are too complex for clinical practice by implementing detailed geometries and muscle forces [2]. Therefore, the aim of our present study was to introduce an efficient computational method to adjust the stiffness of individual 3D-printed insoles during gait.

Methods

Medical image data of a subject (female, 27 years old) without diabetic foot syndrome was used to generate the individual left foot and a simplified skeleton geometry (one geometry). The shape of the sole was adapted to the subject's foot by means of CAD, for which 3D scan data of the foot was used. The designed insoles were then additively manufactured from the thermoplastic copolyester with a gyroid filling structure and a filling of 20 %. A 3D gait analysis including sole pressure measurement was carried out with the additively manufactured insole. The data from the pressure measurement was used to validate the FE model. Afterward, the ground reaction forces and marker trajectories obtained from the 3D gait analysis are imported into a musculoskeletal multibody (MMB) model (AnyBody Technology, Aalborg, Denmark) to calculate ankle joint reaction forces and moments during the gait. Subsequently, the FE model was created in Abaqus (Dassault Systèmes, Vélizy Villacoublay, France) from the geometries and boundary conditions obtained. Two time points were analyzed, the early and late mid-stance phases (Fig. 1).

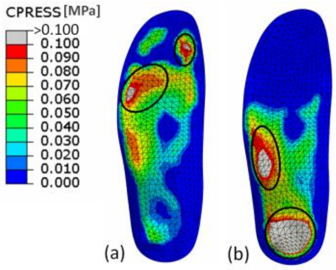

Figure 1: Plantar pressure distribution [MPa] of the left insole (view from above) during the first peak (a) and low point (b) of vertical ground reaction forces and the partitioning of the insole.

The stiffness of the individual insole was adjusted in areas where high plantar pressures occurred by applying soft insole plugs (>100 kPa). Three different Young´s moduli were analysed in these areas (0.5 MPa, 1.0 MPa, 1.5 MPa) (Tab. 1).

Results

Validation shows a difference of 234 kPa between the experimental and simulated plantar peak pressure at the first high point and 30 kPa at the low point of the vertical ground reaction force. Adjustment in stiffness in areas with plantar pressure greater than 100 kPa resulted in a plantar reduction of approximately 16% to 26% by using a Young's modulus of 0.5 MPa.

	Percentage reduction in peak plantar pressure [%]		
Area	1.5 MPa	1.0 MPa	0.5 MPa
Heel	-8.16	-13.01	-22.96
Lateral Midfoot	-4.44	-8.15	-16.30
Lateral Forefoot	-6.78	-10.17	-16.95
Toe	-11.30	-17.39	-26.09

Table 1: Percentage reduction of the peak plantar pressure depending on the Young´s moduli 0.5 MPa, 1.0 MPa and 1.5 MPa of the corresponding soft insole used compared to the peak plantar pressure.

Discussion

Due to the observed deviation between the experimental and the numerically calculated peak pressure in the plantar, it is hardly possible to determine the quantitative reduction of plantar pressure by FE modeling. However, the areas of high plantar pressure can be identified, enabling the adjustment of the stiffness during gait by means of parameter analysis. By calculating joint reaction forces and moments using MMB modeling as boundary conditions, our presented method is time-efficient compared to FE models, where muscle forces are applied complexly.

References

1. Bus et al, Diabetes Metab Res Rev, 36:1-e3269, 2020.
2. Behforootan et al, Med Eng Phys, 39:1-11, 2015.

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