Effects of prosthetic design parameters on running performance of a unilateral transfemoral amputee

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Abstract
Carbon fiber running-specific prostheses (RSPs) are widely used among lower-limb amputee runners. However, which prosthesis provides the best performance for runners remains unknown. For this purpose, a computational model of the human body with a prosthesis was created and the effect of the prosthetic parameters on performance was investigated. First, motion capture systems were used to collect motion data from amputees. Furthermore, marker and force plate data were obtained to create a digital human model. Kinematic data such as limb lengths and joint angles were calculated using marker data. Afterward, the inertial properties were estimated to conduct inverse dynamic analyses. After building a computational model of amputee sprinting, the joint positions and ground reaction forces (GRFs) were compared with the experimental results. The design parameters of the prosthesis were introduced to understand the effects of the prosthesis on motion and performance. The response surface method was used to express motion adaption regarding the geometry and stiffness of the prosthesis. Hip and knee sagittal joint angles were updated based on the response surface method to simulate joint motion adaptations of the worn prosthesis. Additionally, average horizontal velocity, horizontal velocity change over one gait cycle, vertical and horizontal impulses were considered as performance functions. An evaluation parameter was proposed to generalize the idea of performance. The moment of the prosthetic knee and the closest point of the prosthesis to the ground during the swing phase were defined as design constraints to consider knee buckling and prosthetic leg tripping, respectively. The effect of the design parameters on the performance and constraint functions was also investigated and a method to determine and design a suitable prosthesis for an individual was proposed. It was revealed that proper selection and design of prostheses represent an important way to increase performance.

Keywords: Amputee running, Running-specific prosthesis, Prosthetic parameters, Running performance, Multilink model, Motion capture

1. Introduction

Running-specific prostheses (RSPs) have been developed over the last two decades and it has been shown that shape/stiffness regulation of RSPs may provide better sprint performance for individuals with lower extremity amputation (Nolan, 2008). Previous studies utilized mass-spring (Hobara et al., 2013), multibody (Fey et al., 2013), and finite element (Rigney et al., 2015) models to investigate the effect of prosthesis on the running mechanics, where contact time (Beck et al., 2017), leg stiffness (Beck et al., 2017) (Sano et al., 2017), and mechanical energy flows (Rigney et al., 2017) were evaluated. Although these studies identified how the running mechanics varied by wearing an RSP, little is known about how RSPs should be designed for individuals with lower extremity amputation.

To understand the effect of RSP design (e.g. shape and stiffness) on biomechanics and sprint performance, a
computational multibody model of the runner with an RSP was developed together with biomechanical data. For example, previous studies used ground reaction forces (GRFs) from force-instrumented treadmill (Grabowski et al., 2010) and joint kinematics and kinetics from 3-D motion capture systems (Sepp et al., 2019). However, computational methods with experimental data often lead to complicated design strategies. For example, it is difficult to determine the location of a representative RSP ankle joint to carry out inverse dynamic analyses (Baum et al., 2010). Therefore, the present study considers the RSP as a multi-link structure consisting of rigid links with rotational springs and dampers to include its deformation characteristics. Furthermore, as GRFs could also be affected by the mechanical properties of RSPs, digital human models to estimate GRFs (Murai et al., 2018) and contact models (Jackson et al., 2016) have been used in forward and inverse dynamic analyses, respectively.

Another methodological issue to consider in RSP design is the definitive evaluation of sprint performance. Although metabolic cost, joint loading (Fey et al., 2012), and gait asymmetry (Yang et al., 2012) were common parameters in previous studies, we used GRF impulses and horizontal velocity as performance indexes to determine the effect of RSP on sprint performance. Furthermore, we considered reducing the risk of “knee buckling” (defined as the sudden loss of postural support across the knee at the time of weight bearing) induced by the amount of internal knee flexion moments (Namiki et al., 2019) and tripping at the mid-swing phase for RSP design.

The purpose of this study was to establish a design methodology based on prosthetic parameters by considering the effect of the design on the motion of joints and the runner’s performance. A computational human model has been built, and inverse dynamic analyses have been carried out. Horizontal velocity and GRFs were calculated using a computational human model with different RSP and generated motions based on the prostheses worn by the runners. The effect of each design parameter on performance and constraints were discussed. This study proposes a comprehensive parametric design method for RSPs to improve the running performance in individuals with lower extremity amputations.

2. Methods and modeling

The sprinting performance of a unilateral amputee was investigated through the proposed experiment-based computational RSP design methodology as shown in Fig. 1. First, the design space of the RSP was determined by the design of experiments and the computational model of a sprinting unilateral amputee was created. Afterward, the response surfaces were modeled for the joint motion depending on the prosthetic design parameters. Inverse dynamic simulations of a sprinting amputee were conducted with various RSP designs. Finally, the sprinting performance was evaluated depending on the RSP design parameters.

![Flowchart of the methodology.](image)

2.1 Subject characteristics

A female unilateral transfemoral amputee (Age 21 years old, Height 1.56 m, Mass 58.3 kg) participated in this study. The participant was informed about the experiment and the research purpose and informed consent was obtained. The Ethical Committee of the University of Tokyo/Office for Life Science Research Ethics and Safety approved the research. The participant performed trials using a single prosthetic knee joint (3S80, Ottobock, Duderstadt, Germany) and nine different RSP blades which were originally manufactured for the research purposes. The nine different RSP blade designs were determined based on the 4-factor 3-level orthogonal array according to the design of experiments. The 4 factors were design parameters of the RSP blade which will be explained in '2.3 Computational model of a human with an RSP'.
2.2 Experimental procedures and data collection

After performing adequate warm-up exercises, the participant carried out a maximum sprint on a straight 40-meter indoor track. To collect the 3-D coordinate positions of the body and GRFs, a motion capture system (20 cameras, VICON MX, sampled at 200 Hz) and seven force plates (AMTI, sampled at 1000 Hz) were positioned approximately 22-m from the starting line. Helen-Hayes marker set (van Sint Jan, 2007) was adopted to place markers on the parts of the body, prosthetic knee and RSP. Trials in which the participant stepped within the boundaries of the force plates were accepted as successful trials. Four successful trials were obtained for each prosthetic blade. Sufficient rest time was provided between each trial.

2.3 Computational model of a human with an RSP

The raw GRFs and 3D marker positions were filtered using a fourth-order zero-lag low-pass Butterworth filter with cut-off frequencies of 75 and 20 Hz, respectively (Namiki et al., 2019). Afterward, the human body was discretized into rigid bodies and joints. As the main motion during running occurs in the lower extremities, the legs were well discretized, but the head-arms-trunk model was used to model the upper body for the sake of simplicity. The RSP was also discretized into rigid links and rotational spring dampers (Rigney et al., 2016). The mass and inertial properties of each limb were estimated by using the anthropometrical measurements chart (Winter, 2009). The joint positions were defined as simple relations between markers as shown in Fig. 2 and local orientation axes were created using markers on each segment.

Euler angles were calculated by using local coordinate systems between two rigid bodies. The body and marker locations were reconstructed after calculating joint angles, because the length of each varies over time due to soft tissue movement, measurement errors, etc. The length of each limb was adjusted to remove kinematic inconsistencies (Pàmies Vilà, 2012) and mean values of the curves were assumed as the lengths of each limb. MATLAB® (The Mathworks, Natick, USA) was used to process the data to reconstruct the human body and motion, and to calculate the joint angles.

The RSP was discretized into six rigid bodies connected by revolute joints and spring-damper elements between rigid bodies to include elasticity effects in the structure. The mechanical properties of the RSP (E91 Runner, Ottobock, Duderstadt, Germany) were investigated by the force-displacement characteristics obtained by compressive testing. Force was applied at the mounting position of the RSP and the horizontal motion of the tip was limited using a fixture. Afterward, the stiffness of each spring between the rigid segments was adjusted by considering the design parameters of the RSP, stiffness in the vertical direction, height, width, and toe length as shown in Fig. 3. Additionally, nine running blades were designed to investigate the effect of the abovementioned parameters on human behavior. The stiffness and damping parameters in the prostheses were updated for each blade.

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Fig. 2 Marker positions and rigid links on human body. Blue circles, red crosses and black lines represent marker locations, joint locations and body segments, respectively.
2.4 Inverse dynamic analyses

A computational model of the human body was created using the joint definitions, joint angles, and inertial properties obtained in the previous steps as explained in section 2.3. First, we carried out inverse dynamic analyses by using experimentally obtained GRFs to determine kinematical accuracy. Ground reaction forces and joint angles were imposed as inputs of the inverse dynamic analyses. Joint moments and the positions of the joints were obtained. Modeling and analyses were carried out using Altair MotionView and Altair MotionSolve. Kinematic measures, such as joint angle and length of limbs, and stiffness and damping coefficient of the RSP were validated by inverse dynamic analyses with experimental GRF data. As the damping of the prosthesis is a dynamic characteristic of the structure, damping coefficients of the spring-damper elements were also adjusted in this step. Since the aim of the research was to improve the sprinting performance through the design of the RSP, the position of the RSP was important. The height of the RSP blade was a design parameter and it could change depending on the RSP design. The knee joint was connected to the RSP by a rigid bar, and it was not dependent on the design parameters. Therefore, the global positions of the right and left knee were used to evaluate and validate the model. According to Fig. 4, the model shows a similar behavior with the experimental results. Asymmetry of motion can be observed by considering the positions of the prosthetic and sound leg knee. Larger displacement occurred in the knee joint position of the prosthetic leg compared to that of sound legs. The differences between the experimental and the computational models occur due to soft tissue movements, kinematic constraints, measurement errors, and the fixed center of pressure point on RSP and foot.
Inverse dynamic analyses with a contact model were carried out to predict the runner’s sprint performance with a RSP that was not worn by the subject. The impact contact model was used to define the foot-ground and prosthesis-ground interactions. Spheres were used to define the contacting geometries of the foot and prosthesis. Joints angles were used as inputs of the inverse dynamic analyses with contact model. GRFs and the joint moments were obtained. The model was validated by comparing the experimental and computational GRFs and knee positions as was done in previous inverse dynamic analysis. According to the results shown in Fig. 5 and Fig. 6, the contact time, maximum peak force in the vertical and horizontal directions for the prosthetic and sound leg, and the curve trends of GRFs express similar characteristics. Besides, the positions of the knee joints have similar trends with the experimental results. The error in the results of inverse dynamics with contact model is lower than the error in inverse dynamics with experimental GRF in some regions. Because the center of pressure can move due to the contact model, and modeling accuracy increases.

Fig. 4 Comparison of knee joint positions obtained by experiment and computational model via inverse dynamic analyses with experimental GRF data.

Fig. 5 Comparison of the experimental and the computational GRFs in horizontal and vertical directions. First, the prosthetic leg touched ground, and the sound leg followed second contact. Finally, the prosthetic leg touched ground again. Experimental vertical, $F_{v,exp}$, and horizontal ground reaction forces, $F_{h,exp}$, were compared with computational vertical, $F_{v,sim}$, and horizontal ground reaction forces $F_{h,sim}$. 

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Understanding the effect of each shape/stiffness parameter of RSPs on design and motion facilitates increasing runner performance. The response surface method was used to learn how the motion of the joints and the performance of runners change depending on the RSP. Analyses of the various prostheses in the design domain were conducted and various results were acquired.

The motion of the human body adapts itself regarding to whichever prosthesis is used; this is important to understand how motion will change. The 3D marker data of all trials (nine blades with four trials for each blade) were collected and local coordinate systems were defined on each segment by using 3D marker positions. Joint angles were calculated by using Euler angles between local coordinate systems of the segments. The motion of the joints was assumed as periodic and the period was calculated as one gait cycle. Afterward, the motion of the joints was normalized by time to evaluate the characteristics. Second-order polynomial response surfaces were created to approximate the effect of each parameter on period and joint motion in the sagittal plane. Finally, the approximated motion of humans was obtained by considering the parameters of the prosthesis used. The effects of each parameter on the prosthetic knee joint angle were as given in Fig. 7.

![Fig. 6 Comparison of knee joint position obtained by experiments and computational model via inverse dynamic analyses with contact model.](image)

2.5 Response surface method for motion and performance

The response surface method was used to learn how the motion of the joints and the performance of runners change depending on the RSP. Analyses of the various prostheses in the design domain were conducted and various results were acquired.

The motion of the human body adapts itself regarding to whichever prosthesis is used; this is important to understand how motion will change. The 3D marker data of all trials (nine blades with four trials for each blade) were collected and local coordinate systems were defined on each segment by using 3D marker positions. Joint angles were calculated by using Euler angles between local coordinate systems of the segments. The motion of the joints was assumed as periodic and the period was calculated as one gait cycle. Afterward, the motion of the joints was normalized by time to evaluate the characteristics. Second-order polynomial response surfaces were created to approximate the effect of each parameter on period and joint motion in the sagittal plane. Finally, the approximated motion of humans was obtained by considering the parameters of the prosthesis used. The effects of each parameter on the prosthetic knee joint angle were as given in Fig. 7.

![Fig. 7 Variations of prosthetic knee joint angle in the sagittal plane with respect to the RSP design parameters: stiffness, width, height and toe length in normalized time. The interval [-1, 1] represents the minimum and maximum values in the design domain for each design parameter.](image)
After creating the response surfaces of motion, new computational models were created. When the shape and stiffness of the prostheses changed, the stiffness and damping ratio of the joints in the multi-link prosthesis model were updated to run the analyses correctly. Finally, computational models in the design domain were created and inverse dynamic analyses of all RSP designs with contact model were repeated to obtain responses. Four different objective functions were created to evaluate the sprint performance of the running trials. As one of this study’s main aims is to develop the sprint performance of runners, performance-related design functions were defined. These functions were horizontal velocity change over one period, average horizontal velocity, and the horizontal and vertical impulse of GRFs. After obtaining the outputs of the horizontal position, and the velocity of the runner’s center of mass, and GRFs for all the models in the design space, response surfaces were created to hold information about the effect of each RSP design parameter.

3. Results and discussion

The joint motions for any RSP design parameters were estimated by using response surface of the joint motion. Inverse dynamic analyses with the ground contact model were carried out to estimate the sprinting performance. Figure 8 shows the response surface of the horizontal velocity change over one period. An increase in both stiffness and width in the design domain improves the runner’s sprint performance, while height and toe length have less effect on the sprint performance in unilateral amputees. Therefore, bending stiffness distribution along the blade, contacting area, and joint position of the multi-link model of the RSP were adjusted by determining the RSP parameters to improve the sprinting performance.

Fig. 8 Change in horizontal velocity over one period with respect to the RSP design parameters. The interval [-1, 1] represents the minimum and maximum values in the design domain for each parameter.

The horizontal velocity change over one period and the average horizontal velocity showed similar responses; however, the behavior of GRF impulse-related functions differed slightly. Therefore, an evaluation parameter was proposed to investigate the effect of both vertical and horizontal impulses, $I_{ver}$ and $I_{hor}$ as single value. A bouncing ball analogy was used to derive an evaluation parameter by simplifying the motion of the runner as shown in Fig. 9. It was assumed that a longer distance after each contact would increase performance. The human body was considered as mass $m$, and initial velocities located on the horizontal and vertical axes, as $u$ and $v$, respectively. In the vertical direction, velocities before and after contact were assumed to have the same magnitude, but in the opposite direction. Aerial time, $t_{aerial}$, was obtained by calculating the velocity in the vertical direction as follows in Eq. (1).

$$t_{aerial} = \frac{2u}{g}$$

$$I_{ver} = \frac{2m}{g}$$

Afterward, the change in velocity in the horizontal direction, $v_{new} - v_{old}$, was calculated using the horizontal impulse. The horizontal velocity after contact, $v_{new}$, was written as follows in Eq. (2).

$$v_{new} = v_{old} + I_{hor}$$
If initial velocities were assumed to be the same, the final velocity would be directly related to the change of velocity. Then, horizontal distance after contact was given in Eq. (3) according to bouncing ball model.

\[
\text{Horizontal distance} = \frac{I_{\text{ver}}}{mg} \times \left( \frac{I_{\text{hor}}}{m} + v_{\text{old}} \right) \tag{3}
\]

Consequently, the evaluation parameter was obtained by omitting the initial horizontal velocity term from the horizontal distance as given in Eq. (4). It was revealed that vertical and horizontal impulses should be considered together to understand performance.

\[
\text{Evaluation parameter} = \frac{I_{\text{ver}} I_{\text{hor}}}{m^2 g} \tag{4}
\]

The importance of the evaluation parameter was shown by comparing four different models with different horizontal and vertical impulses. The evaluation parameter was calculated, and the relation between the horizontal velocity change over one period, and the evaluation parameter shows that the evaluation parameter is useful to simplify impulsive responses, as shown in Fig. 10. It was concluded that the change in shape and stiffness of the RSP affect the athlete’s performance. Therefore, the choice of prosthesis is important for individuals to perform well.
Conversely, knee buckling can be a risk for transfemoral amputees who use artificial knee joints; therefore, knee buckling risk should be considered in the RSP design and optimization process. Knee buckling occurs when a moment exists in the direction of knee flexion in the stance phase. The relative position of the artificial knee with respect to the center of pressure and GRF vectors determines the direction of the joint moment around the knee joint. Additionally, toe length parameter changes the bottom half of the shape of the prosthesis. Thus, the effect of RSP toe length on prosthetic knee moment was investigated by using inverse dynamic results with ground contact model. As can be seen in Fig. 11, reducing the toe length causes a smaller moment in the opposite direction of knee flexion. In other words, it is easier to observe knee buckling if the RSP toe length is reduced. However, increasing the toe length may cause a different type of trouble, such as tripping during the swing phase. As it may create a fall risk, the closest position to the ground was obtained. According to Fig. 12, when the toe length increases, the trajectory of the RSP becomes closer to the ground. Therefore, an increase in toe length also increases the risk of touching down to the ground while swinging.

![Knee moment caused by ground reaction forces](image1)

**Fig. 11** Amount of the moment around the knee joint decreases when toe length is reduced. Therefore, the possibility of knee buckling increases with lowering toe length.

![Prosthesis Position](image2)

**Fig. 12** 9 different model for prostheses were created computationally and the closest point of prosthesis to ground was obtained for each model. 1st order linear fitting was applied and it was revealed that high level toe length increases the tripping risk.
Although the response surface of the joint angles was useful to consider the effect of the RSP design parameters on the motion, the accuracy of the experiments affected the accuracy of the response surface of the joint motion. This limitation of the methodology could be reduced further by increasing the number of the experiments. Another limitation of the proposed methodology is the limited design space of the RSP. According to the proposed methodology, the dimensions and the stiffness of the RSP are investigated depending on the base shape. In addition to the limitations related to base shape, the response surface may not represent outside of the limits of the design parameters. Thus, the shape alternatives of the RSP blades are limited.

4. Conclusions

This study proposes a methodology to design RSPs to improve the sprinting performance of runners. It was revealed that the design of the RSP affects running kinematics, sprinting performance, and the potential for knee buckling and tripping. A 3-D model of a unilateral transfemoral amputee was created. Inverse dynamic analyses of the runner were carried out and validated by using the joint positions and GRFs. Afterward, joint angles for all trials were calculated by using 3-D motion capture data. Changes in human behavior due to different prosthesis design parameters were approximated using the response surface method and new computational models were generated using estimated motion data. The sprint performance was evaluated by acceleration-related functions. Knee buckling and tripping during the swing phase of the prosthetic leg were considered as limitations. Finally, the case-optimal prosthesis regarding four different performance functions was obtained, and it was revealed that an increased stiffness and width give better performance for the runner. While a high toe length increases the risk of tripping when using the prosthetic leg, a low toe length increases the risk of knee buckling. In conclusion, the prosthesis has a significant effect on the sprint performance of an athlete with RSP and the choice of a prosthesis for a runner is important. In particular, running asymmetry requires that unilateral amputee runners develop different strategies to run faster and maintain their balance depending on their prosthesis. The response surface method can be used to understand how the shape and stiffness of the prostheses affect the motion of the joints to which they are fitted.

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