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Keywords
muscle, loading, activity, trans-tibial, amputees, during, standing, effects, heel, shoe, height

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Effects of Shoe Heel Height on Loading and Muscle Activity for Trans-Tibial Amputees During Standing

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Abstract: This study accesses the effects of shoe heel heights on loading, muscle activity, and plantar foot pressure of trans-tibial amputees during standing. Five male subjects with unilateral trans-tibial amputation volunteered to participate in this study. Three pairs of shoes with zero, 20 mm, and 40 mm heel heights were used. The loading line of the prosthetic side, the plantar foot pressure, and the surface electromyography (EMG) of 10 muscles were simultaneously recorded. With increasing shoe heel heights during standing, the loading line of the prosthetic side shifted from the anterior to the posterior side of the knee center, the peak pressure was increased in the medial forefoot region, and the peak pressure was reduced in the heel region. The EMG of the medial and lateral gastrocnemius of the sound leg almost doubled and that of the rectus femoris, vastus lateralis, and vastus medialis of the prosthetic side increased to different extents with increasing heel heights from zero to 40 mm. These results show a high correlation with human physical behavior. Changing of the heel heights for trans-tibial amputees during standing actually had similar effects to altering the prosthetic sagittal alignment. The results suggest that an alignment change is necessary to accommodate heel height changes and that prosthesis users should be cautious when choosing shoes.

Key words: shoe-heel height; loading line; plantar foot pressure; surface electromyography (EMG); prosthetic alignment

Introduction

Comfort, function, and appearance are three of the most important criteria used to evaluate the quality of a trans-tibial (TT) prosthesis. The alignment of a prosthesis, defined as the relative position and orientation of the prosthetic components, has been recognized as one key factor that determines the success of a prosthesis for TT amputees[1]. Malalignment always results in deviation of the standing pattern and gait, increased energy cost, pain, and even tissue breakdown at the stump.

After bench alignment, dynamic alignment has to be iteratively done according to the amputee feedback and prosthetist inspection. This is a trial-and-error process, which is very time-consuming and subjective, and relies on the prosthetist experience. Different factors such as the mechanical properties of the residual limb, components used to assemble the prosthesis, and patient history make each amputee’s alignment unique[2]. Quantitative understanding of alignment can provide the prosthetists with guidelines and tools to optimize or measure alignment. Much work has been done to obtain more quantitative techniques and optimization criteria for TT prosthetic alignment[3-6]. These studies provided quantitative information to practitioners that is difficult to obtain through direct subjective observation and enabled rehabilitation outcomes to be...
measured which serves as valuable evidence of treatment efficacy. However, even though the prosthesis is aligned optimally for one amputee, the alignment would be changed if the prosthesis was not used properly or different prosthesis settings were used.

It is well accepted that persons who use lower-limb prostheses are often restricted to the choice of a narrow range of heel heights\(^7\). However, different heel heights are necessary in daily life but they may destroy the alignment settings fitted by the prosthetist. The objectives of this study are to look into the effects of shoe heel heights on the knee loading, muscle activity, and plantar foot pressure for TT amputees during standing to provide quantitative information for biomechanically optimal prosthetic alignment.

1 Methods

1.1 Subjects and prostheses

Five males with unilateral trans-tibial amputation participated in this work. Table 1 outlines the subject characteristics. Each subject signed an approved human subjects consent form. All prostheses were modular patella-tendon-bearing (PTB) with a solid-ankle-cushion-heel (SACH) foot. The subjects had no other concomitant disabilities or skin complications. An experienced prosthettist participated in this study to ensure the function of each participant’s prosthesis.

<table>
<thead>
<tr>
<th>Subject number</th>
<th>Age (years old)</th>
<th>Body mass (kg)</th>
<th>Height (cm)</th>
<th>Prosthetic side</th>
<th>Years since amputation</th>
<th>Cause</th>
<th>Activity level</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>23</td>
<td>57</td>
<td>170</td>
<td>Left</td>
<td>6</td>
<td>Trauma</td>
<td>High</td>
</tr>
<tr>
<td>2</td>
<td>58</td>
<td>70</td>
<td>168</td>
<td>Right</td>
<td>12</td>
<td>Trauma</td>
<td>Moderate</td>
</tr>
<tr>
<td>3</td>
<td>32</td>
<td>75</td>
<td>177</td>
<td>Left</td>
<td>3</td>
<td>Infection</td>
<td>High</td>
</tr>
<tr>
<td>4</td>
<td>44</td>
<td>65</td>
<td>172</td>
<td>Left</td>
<td>11</td>
<td>Trauma</td>
<td>Moderate</td>
</tr>
<tr>
<td>5</td>
<td>28</td>
<td>69</td>
<td>174</td>
<td>Left</td>
<td>2</td>
<td>Trauma</td>
<td>High</td>
</tr>
</tbody>
</table>

Three shoe heel heights were investigated in this study with zero, 20 mm, and 40 mm heel heights. All the shoes were commercially available. The experimental setup is shown in Fig. 1. The loading line on the prosthetic side, the plantar foot pressure, and the surface electromyography (EMG) were simultaneously recorded to describe the effects of heel heights.

The Otto Bock “L.A.S.A.R. Posture” alignment system (Germany) was used to display the loading line on the prosthetic side. This device determined the vertical component of the ground reaction forces acting on its sensing platform. When both feet were on the force platform, the subject’s weight and the location of the weight bearing line could be measured. If only one side, e.g., the prosthetic limb, was on the platform, the force on that leg only and the resultant load line was measured. The standing posture of the subject was studied by measuring the horizontal distance between the load line and each of the four body landmarks at the lateral malleolus, the knee joint centre, the great trochanter, and the middle of the shoulder in the sagittal plane according to images recorded by a digital camera.

The plantar foot pressure was measured using the Pedar in-shoe Pressure Measurement System (Novel Electronics, Munich, Germany) at a sampling rate of 50 Hz. The main components of the measurement system included two insoles (w-9561-951) embedded with 99 sensing elements and a wireless transmitter. The insoles were connected to the wireless transmitter via a serial cable so that the pressure data could be transmitted to a computer. An assistant helped hold the cable pack and device box to minimize its effect on the subject. The subjects were asked to unload each foot in turn to calibrate the insoles to zero. The pressure magnitudes were normalized by dividing them by the body
weight of each subject to allow comparison among different subjects. The surface EMG signal was recorded at a sampling rate of 1500 Hz using pre-amplified bipolar, surface electrodes (Noraxon, Scottsdale, AZ, USA), which were placed on the clean-shaven skin overlying the biceps femoris, rectus femoris, vastus lateralis, and vastus medialis of both legs and the medial and lateral gastrocnemius of the sound leg. The ground electrode was placed overlying the tuberosity of the tibia. The raw EMG data was filtered with a band pass Butterworth filter (10-500 Hz). Full wave rectification and smoothing of the EMG signal were accomplished using root-mean square values over a 50 ms interval.

Since the shoes with the 20 mm heel height were the normal choice according to the subjects’ responses, the prostheses were aligned to be optimal for subjects with this height before the experiments. The alignment was kept the same as the heel height was changed. Each trial was done twice to reduce the uncertainty, with a short break between trials to allow the subjects to get used to the new shoes. Each subject was asked to load the prosthesis to 40% of body weight as measured by the “L.A.S.A.R. Posture” alignment system for each trial. All the data was recorded for 5 s while the subjects were asked to keep the standing posture for about 10 s.

2 Results

2.1 Prosthetic side loading line

The prosthetic side loading line shows how the load acts on the prosthesis side as an indicator of the influence of the heel height. To eliminate the effect of the different body heights of each subject, the loading line shift was divided by the subject height. Figure 2 shows that the five subjects showed a consistent trend with the higher heel giving a large gradient for the whole body. Moreover, the change of the distances between the loading line and the four marks were not linear. The knee joint inclined most as the heel height increased while the other body landmarks inclined less, perhaps due to the rigid prosthetic ankle joint causing the amputees to adjust the posture mainly by flexing the knee joint and then the great trochanter to seek the best posture. Therefore, the amputees adjust the standing posture mainly by moving the lower limb when they lose the standing balance.

Considering the normal 20 mm heel height as the reference case, the data in Fig. 2 also shows that the subjects were more sensitive to the shorter heel height than to the higher height since the shorter heel resulted in greater shifts of the loading line from the four marks. Thus, the subjects deal better with the increased heel height than with the reduced heel height and the shorter heel height may cause greater balance problems than the higher heel during the prosthetic alignment. Thus, amputees should try to use a constant heel height to protect the prosthetic alignment.

Fig. 2 Shift of the prosthetic side loading line with the change of heel heights. Minus values denote that the loading line is anterior to the mark. The different line types refer to the five subjects.

2.2 Plantar foot pressure

The peak plantar pressure on the foot during standing was determined using the Pedar software applied to the insole system. Figure 3 shows the three regions where most of the pressure fell based on the ratios of the foot width and length. Figure 4 compares the peak normalized pressure in the different foot regions for the three test conditions for all five subjects. For each subject, higher heel heights resulted in larger peak pressures in the medial forefoot region. In contrast, the peak pressure in the heel region decreased as the heel height increased. The peak pressures in the lateral forefoot regions did not change significantly with changes in the heel height. The results in Fig. 4 can be explained by the increased heel height inducing shank flexion and forward tilting of the knee joint. The center of pressure was moved forward, which caused the peak pressure to increase in the medial forefoot region but decrease in the heel region. When the heel height was
changed from 0 to 40 mm, the subjects had to change their standing posture to maintain balance. These results corresponded to the shift of the loading lines as shown in Fig. 2.

Fig. 2 Definition of foot regions

2.3 Surface EMG of the lower limb muscles

The EMG signal is widely used to analyze muscles activity. The mean absolute value (MAV) of the EMG was used to show the influence of the heel height changes,

$$\text{MAV} = \frac{1}{N} \sum_{i=1}^{N} |x_i|$$

where $x_i$ is the signal and $N$ is the number of points.

Figure 5 shows the typical muscle activity for the sound and prosthetic sides for the various heel heights (this data is from Subject 1, with the others showing the same trend). During static standing, the EMG signals of the knee muscles of the sound leg did not change much as the heel height was varied, whereas EMG-MAV of the medial and lateral gastrocnemius of the sound leg almost doubled as the heel height was increased from zero to 40 mm because the two posterior muscles which are responsible for ankle plantarflexion, play an important role in standing balance\[8\]. In contrast, the EMG of the knee muscles of the prosthetic side was strongly affected by the heel height. As the heel height was increased from zero to 40 mm, the knee flexion angle increased as a result. The knee extension muscles needed to be more active to maintain the standing balance and knee stability. The results in Fig. 5 show that EMG-MAV of the rectus femoris, vastus lateralis, and vastus medialis all increased as the heel height increased while the EMG-MAV of the biceps femoris was slightly reduced.

Fig. 5 Mean absolute value (MAV) of the EMG of various muscles for the three heel heights (Subject 1). Sound side: 1, medial gastrocnemius; 2, lateral gastrocnemius. Prosthetic side: 3, rectus femoris; 4, vastus medialis; 5, vastus lateralis; 6, biceps femoris.

3 Discussion

The importance of prosthetic alignment to the success of a prosthetic fitting is well known, but work remains to be done in this area. The proper use of prostheses by amputees is also important to ensure achievement of optimal prosthetic function. Much work has targeted at the quantification of alignments to give a growing understanding of the primary goals of alignment to produce more consistent, more objective, and faster techniques for prosthetists. Since prosthesis users have to change shoes in their daily life, the effect of heel height on prosthetic alignment deserves attention. Seelen et al.\[9\] analyzed the effects of prosthetic antero-posterior alignment on the pressure distribution at the stump/socket interface. Their work focused on TT amputees during unsupported stance and gait using a 0.5 cm heel with forefoot wedging. Although Hansen et al.\[7\] reckoned that an alignment change is necessary to accommodate higher or lower heel heights for
changes beyond a small range, only the effects of heel heights on the rollover characteristics of the biological ankle-foot system for non-disabled adult females were studied. In the current study, the loading line on the prosthetic side, the activities of knee muscles on both sides, and the plantar foot pressure were measured for three different heel heights for a TT amputee during standing. The results for the three biomechanical properties correlated well and accorded with human physical behavior. Furthermore, the heel height change for the TT amputee during standing actually caused a change in the orientation angle in the sagittal plane of the socket relative to the ground. These results were comparable with other reports about sagittal alignment in the literature.

Analysis of the forces, moments, and pressures on the biomechanical function of the prosthesis can make prosthetic alignment more objective. The standing alignment for the TT amputee has been defined as biomechanically optimal when the anatomical knee center is 15 mm posterior to the individual loading line of the prosthetic side\(^6\). In this study, the prostheses were adjusted to optimal alignment for a heel height of 20 mm with the loading line on the prosthetic side 6-23 mm anterior to the knee center (Fig. 6a), which agrees with the optimal criterion given by Blumentritt et al.\(^6\)

![Fig. 6 Shift of the loading line for different heel heights](image)

(a) 20 mm heel height  (b) Zero heel height  (c) 40 mm heel height

When the heel height was reduced to zero, the knee joint was hyperextended because of the anterior shift of the loading line, as shown in Fig. 6b. The posterior cruciate ligament had to be strained more to keep the knee joint stable causing some subjects to feel some pain at the popliteal depression region. EMG-MAV variations in Fig. 5 support this trend. The rectus femoris, vastus lateralis, and vastus medialis, which control extension of the knee joint, provided less force when the heel height was zero as shown by the EMG amplitude which was approximately proportional to the muscle force when measured under isometric conditions.

However, when the heel height was changed from zero to 40 mm, the loading line shifted posteriorly to the knee center, giving rise to an external flexion moment to the knee joint. The muscle forces of the rectus femoris, vastus lateralis, and vastus medialis had to increase to maintain the standing balance and joint stability. Although the posterior knee ligament relaxed, the increased energy consumption in the muscles could increase fatigue in the amputee. Long-term standing with the higher heel could then damage the knee joint.

The change of heel heights altered the original optimal alignment setting, as shown in the plantar foot pressure. The foot pressure distribution in this study showed the same trends as the study of the effects of shoe inserts and heel heights on foot pressure, impact force, and perceived comfort during walking for healthy females\(^10\) with the increasing heel height increasing the medial forefoot pressure. The shift of the center of pressure on the foot in the antero-posterior direction when adjusting the alignment angle in the sagittal plane\(^11\) can also explain the foot pressure change with different heel heights as shown in Fig. 4.

## 4 Conclusions

In this study, the effects of heel heights of trans-tibial amputees during standing were investigated based on the loading line, knee muscle activity, and foot pressure. Since the subjects would automatically adjust the standing posture to compensate for the changes of the heel height, direct practical criterion for prosthetic alignment may be difficult to identify. This study emphasizes the importance of proper use of the prosthesis. The findings suggest that an alignment change is necessary to accommodate heel height changes and prosthesis users should be cautious when choosing their shoes. Future work will investigate the effects of heel heights on gait, residual limb/socket interface stress, and foot pressure during ambulation with more subjects.

## References

Former British Prime Minister Tony Blair Speaks to Tsinghua Students on Climate Change

Former British Prime Minister Tony Blair delivered a speech to Tsinghua students entitled “A Global Deal in Climate Change: the Role of Business and New Technologies” on March 24, 2009 in the Lecture Hall of the School of Economics Management.

In his speech, Mr. Blair explained the basis of the global agreement and the major challenges in dealing with climate change. He said the present economic downturn presents a good opportunity to solve this problem. He urged governments to make sure a substantial amount of stimulus investment flows into the new clean energy sector. He also broke down the challenges facing all countries into specific aspects. He said people should realize that the responsibility for tackling climate change is not restricted to government, but every one must shoulder some responsibility.

Mr. Blair then answered students’ questions in the discuss session. The subjects of the questions included new clean energy, the global financial crisis, the London Olympics, and university life.

Tsinghua Vice President Xie Weihe had a talk with Mr. Blair prior to the speech.

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