Assistance Effect of an Evolved Heel-Raising Unit for Walking Disabilities

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***Abstract*—Elderly people and people with diseases such as hemiplegia often have a walking disability, which increases their risk of falling and suffering injuries. In walking, the angular velocities when raising the heel and swinging the toe forward are lower for elderly people than for healthy individuals, because of their lower muscle power. We previously developed a heel-raising unit based on a spring, but that unit lacked a mechanism to release the spring at the optimal timing, as reported in the International Journal on Advances in Life Sciences in 2020. We have since developed an evolved heel-raising unit that works when the heel starts to rise after the whole shoe has contacted the floor. Experimental results demonstrate that the evolved unit's assistance effect is better than that of the previous simple unit. We also report a correlation between body weight and the optimal spring stiffness, and we show that the evolved unit does not affect the individual's walking posture as much as the previous one did.**

Keywords: walking disability; walking-assist unit; heel raising; walking posture; muscle power.

I. INTRODUCTION

As the percentage of elderly people in the world's population is increasing [1], the number of functionally impaired people, such as those with hemiplegia, will also increase. People with such diseases and very elderly people often have walking disabilities, which increase their risk of falling and injuring themselves [2].

Our previous study comparing the walking gaits of hemiplegia patients and healthy students showed that the angular velocities while raising the heel and swinging the toe forward were lower for people with walking disabilities than for healthy people. The reason was the lower muscle power of the hemiplegia patients in our study [3]. This suggests that assistance in raising the heel and swinging the toe forward could help people with walking disabilities to have a gait closer to normal.

Two kinds of foot prostheses, the Solid-Ankle Cushion-Heel (SACH) foot (e.g., 1D10, Ottobock, Germany [4]) and the energy storage and return (ESAR) foot (e.g., Vari-Flex, Össur, Iceland [5]), have enabled foot amputees to improve their gait to be close to normal. In the case of the SACH foot, a properly shaped wedge of cushioning material is built into the heel part of the foot prosthesis to absorb shock when the heel of the shoe contacts the floor and thus assist in raising the heel. The ESAR foot has a kind of leaf spring to absorb the contact shock and assist in raising the heel more effectively than the SACH foot does.

Inspired by the ESAR foot, we previously developed a heel-raising unit that used a spring [6]. The unit was very simple, as it comprised only a conical coil spring and a V-shaped attachment cover made of thin stainless steel, but it did not have a mechanism to release the spring at the optimal timing. As a result, it started to raise the heel as soon as the shoe's heel contacted the floor. Instead, the spring should ideally be released when the heel starts to rise in the gait cycle.

Hence, we have developed an evolved heel-raising unit in which the spring is released when the heel begins rising after the whole shoe has contacted the floor. Experimental results with the evolved unit demonstrated that its assistance effect was notably better than that of the previous unit, and that it did not affect the walking posture as much as the previous one did.

In Section II, we describe the differences in gait between a hemiplegia patient and a healthy person to clarify the required characteristics of an evolved heel-raising unit. Then, in Section III, we explain how the SACH foot and ESAR foot compensate for a lack of kicking power when raising the heel. Section IV describes the structure of the evolved heel-raising unit and the experimental results, before Section V concludes with a summary of the key points.

II. GAIT DIFFERENCES BETWEEN HEMIPLEGIA PATIENTS AND HEALTHY PEOPLE

We analyzed the walking gait cycles of both unimpaired people and people with walking disabilities by using a wearable device (WD) and a Kinect to detect warning signs of falls [3]. Every patient with a walking disability in this experiment had hemiplegia and trained periodically at a rehabilitation facility. To estimate the kicking power and the change in angle between the foot and the floor, we experimentally measured the output data from an acceleration sensor and a gyroscope sensor in a WD mounted on the front of a shoe. A Sony SmartWatch 3 was used as the WD.

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Following Tao et al. [7], we divided the normal walking gait cycle into eight phases as shown in Figure 1: (1) initial contact (heel-strike timing), (2) loading response, (3) mid-stance, (4) terminal stance (toe-off timing), (5) pre-swing, (6) initial swing, (7) mid-swing, and (8) terminal swing.

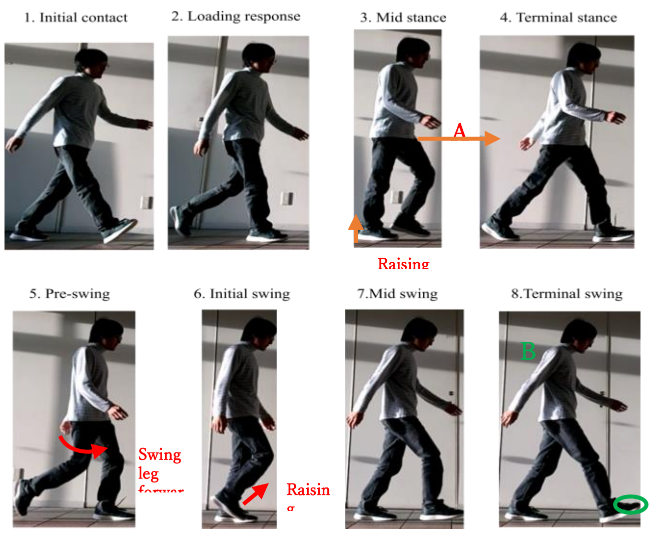


Figure 1. Normal walking gait cycle (right foot).

Figures 2 and 3 show examples of changes in the angular velocity, angle, and acceleration for a healthy participant and one with a walking disability, respectively, over the course of two steps. Each flat region (roughly in the center of each graph) represents the period when the shoe's entire sole touched the floor. The acceleration in this period is roughly 9.8 m/s^2 because the acceleration sensor still measures gravity. The maximum angular velocity at point A indicates the kicking power when raising the heel. This action occurs between the mid-stance and terminal stance, as shown in Figure 1. The minimum angle at point B indicates the toe angle to the floor at the terminal swing.

As seen in Figure 2, the lowest angular velocity at point A for the unimpaired participant was approximately 420°/s. In contrast, as seen in Figure 3, the highest angular velocity at A for the participant with a walking disability was only 250°/s. Thus, the latter participant clearly had a weaker kicking power than the healthy individual when raising the heel, which indicates a clear difference in gait.

Similarly, the highest angle at point B in Figure 2 for the unimpaired person was approximately −18°, whereas the lowest angle at B in Figure 3 for the participant with a walking disability was approximately −8°.

Tables I and II list the averages and standard deviations (SDs) of the measured angular velocity at point A and angle at point B. The angular velocity at A was clearly different between the unimpaired participants and those with walking disabilities. There was also a measurable difference between them in the angle at point B, although the ranges of those values sometimes overlapped.

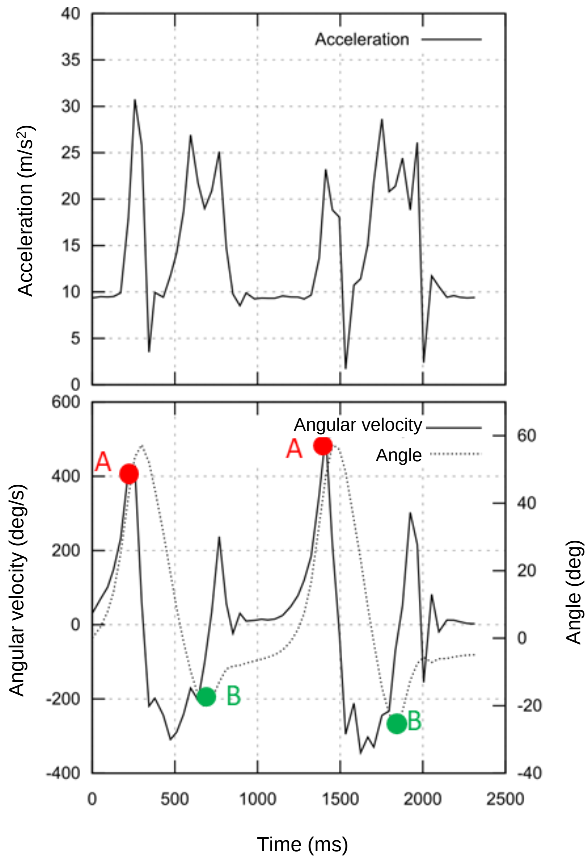


Figure 2. Examples of the angular velocity, angle, and acceleration for an unimpaired participant.

Thus, the individuals with a walking disability had a weaker kicking power when raising the heel than that of the healthy participants. They also had difficulty raising their toes during the terminal swing phase. Accordingly, if additional power could compensate for the difference in kicking power when raising the heel, an individual with a walking disability could walk with a gait closer to that of a healthy person.

TABLE I. ANGLE VELOCITY AT TERMINAL STANCE

Participants	Average (deg/s)	SD (deg/s)
Unimpaired participants	509.36	18.91
Participants with disability	342.06	86.52

TABLE II. ANGLE AT TERMINAL SWING

Participants	Average (deg)	SD (deg)
Unimpaired participants	-17.76	8.02
Participants with disability	-7.45	8.02

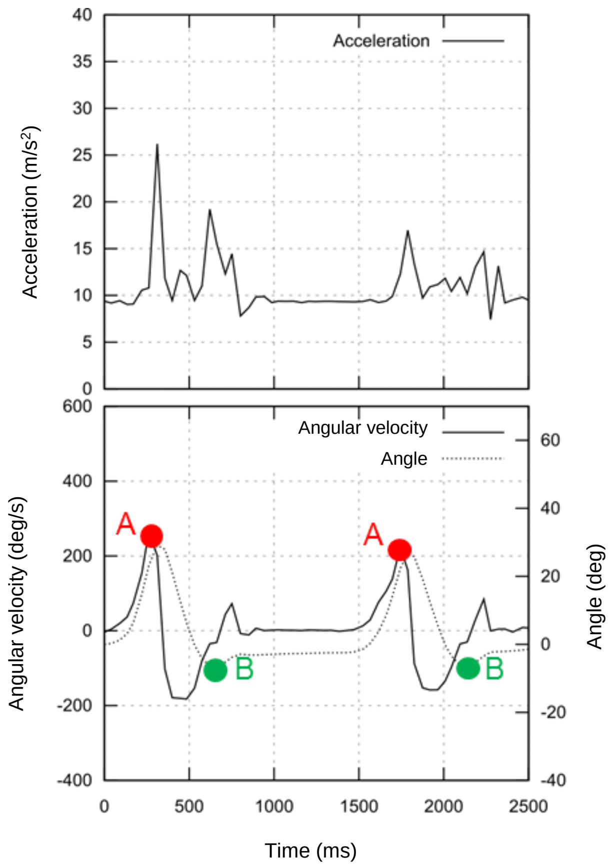


Figure 3. Examples of the angular velocity, angle, and acceleration for a participant with a walking disability.

III. Walking Assistance Mechanisms in Passive Foot Prostheses

There are two types of walking assistance mechanisms in passive foot prostheses. First, the SACH foot [8], shown in Figure 4 [4], was designed to provide shock absorption and ankle action characteristics close to those of a normal ankle without the use of an articulated ankle joint. The SACH foot's action is achieved by the use of two functional elements: a properly shaped wedge of cushioning material built into the heel, and an internal structural core or keel shaped at the ball of the foot to provide a rocker action. A primitive version was first developed toward the end of the 1800s.

Second, the ESAR foot [5], shown in Figure 5, has a kind of leaf spring at its heel and an adequate roll-over shape like that of the foot, which generates the kicking power when raising the heel and increases the dissipated energy during the step-to-step transition in a walking gait. Wezenberg et al. reported that the ESAR foot was more effective than the SACH foot in reducing metabolic energy while walking [9], and Houdijk reported that it evolved the step length symmetry [10].

The cushioning material of the SACH foot or the leaf spring of the ESAR foot compensates for a lack of foot muscle to kick off the floor when raising the heel. This suggests that a shoe with a mechanism to raise the heel would enable elderly people with low muscle power to walk more easily.

Figure 4. Examples of the SACH Foot (1D10, Ottobock, Germany).

Figure 5. Examples of the ESAR Foot (Vari-Flex, Össur, Iceland).

IV. Spring-Assist Unit for Walking Disabilities

A. Structure of heel-raising unit

With the goal of developing a heel-raising unit, we first developed a shoe, shown in Figure 6, to assist individuals with walking disabilities. This shoe had a coil spring and a leaf spring that enabled the wearer to raise the heel more easily. The coil spring force was 15 kg, and the shoe had a roller to prevent the toe from accidentally tripping.

In experiments with this shoe, we noticed a correlation between body weight and the most effective spring power, which would affect the walking posture. Moreover, every participant felt that the timing of generating the spring reaction force was too early for smooth walking.

Figure 6. Heel-assist shoe prototype.

We thus focused on clarifying the correlation between body weight and the optimal spring stiffness to understand its effect on walking posture. Specifically, we developed the heel-raising unit shown in Figure 7. Its mechanism was very simple, as it comprised only a conical coil spring and a V-shaped attachment cover. We adopted the conical spring so that it could be thinner. When stepped on, it provided a spring power of 3, 5, 9, or 11 kg. The attachment cover was made of thin stainless steel.

We measured electromyography (EMG) and posture data from healthy students and elderly participants by using an iEMG [11] and a Kinect [12]. The results demonstrated a linear correlation between body weight and the optimal spring stiffness, and the spring-assist unit did not affect the walking posture.

These studies were reported in the International Journal on Advances in Life Sciences in 2020 [6]. However, the device shown in Figure 7 did not have a mechanism to release the inner spring when the heel started to rise just after touching the floor in the initial contact phase.

Figure 7. Two views of the simple heel-raising unit.

We think that the inner spring should optimally release as soon as the heel rises, and this paper focuses on clarifying the effect of optimizing the spring-release timing. We thus developed the evolved heel-raising unit shown in Figure 8. This model was fabricated with a metal 3D printer. Even though it was made of titanium to be light, its weight was 210 g. The evolved unit comprises a top part A, bottom part B, inner spring, and release timing control part. The top part is placed in the heel part of a shoe. The release timing control part comprises a slider (C) U shaped rod (D), and L shaped rod (E) as shown in Figure 9.

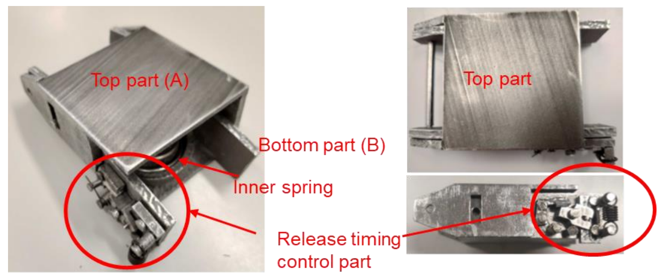
Figure 10 illustrates the configurations for locking and releasing the inner spring. Before touching the floor, the inner spring is in the released state, as shown in Figure 10(a). When the wearer touches the heel to the floor, the top part A goes down, part C slides up, part D rotates counterclockwise, and part E rotates clockwise. Finally, as shown in Figure 10(b), the corner of part E latches with a notch in the top part A.

Next, when the wearer raises the heel from the floor, part C slides down, part E rotates clockwise, and part E rotates counterclockwise. Finally, the corner of part E releases from the notch, as shown in Figure 10(a), and the top part A raises the heel.

B. Experimental setting

Next, we built the evolved heel-raising unit into the heel parts of both left and right shoes, as shown in Figure 11.

To compare the difference in the assistance effect between the simple heel-raising unit and the evolved unit, we measured iEMG and motion data from healthy participants using both of them. Each participant walked with and without springs having power of 3, 5, 7, 9, and 11 kg, in the space shown in Figure 12. Wireless EMG sensors were attached to the gastrocnemius of the right leg to gather the iEMG data. The motion data was measured at the head and mid-hip to analyze the effect on the walking posture, which was measured with a Kinect.



(a) Bird's-eye view (b) Top and side views

Figure 8. Various views of the evolved heel-raising unit.

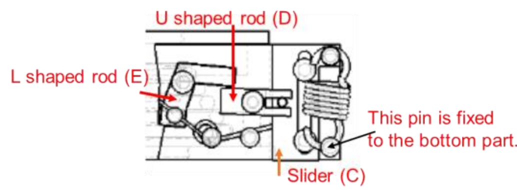
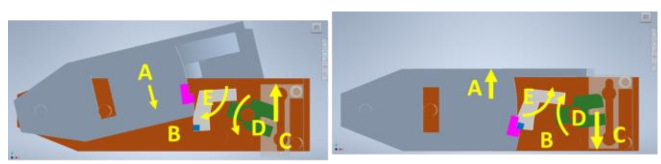


Figure 9. Structure of the release timing control part



(a) Released state (b) Latched state

Figure 10. Configurations for locking and releasing the inner spring.

(a) Spring released state

(b) Spring compressed state

Figure 11. Pair of shoes incorporating the evolved heel-raising unit.

Figure 11. Measurement space.

C. Evaluation of assistance effect

Figure 13 shows the iEMG values for each participant with each spring stiffness, and Figure 14 shows the improvement rates (IRs) due to the heel-raising unit. Here, the IR was calculated as follows:

$$IR = 100 \times (B - A)/B;$$

A: iEMG at most effective spring stiffness;
B: iEMG without heel-raising unit.

The IRs were different for each participant, but those with the evolved unit were clearly larger than those with the simple unit. Note that the measured iEMG values for participant C were much larger than those of the other participants; however, the reason for this remains unclear.

Figure 15 shows the spring stiffness at the lowest iEMG value with respect to the participant's body weight. In contrast to the case of the simple unit, the results for the evolved unit did not show a linear correlation between body weight and the optimal spring stiffness. However, because the number of participants was not enough for appropriate evaluation, we will need to measure such data for more participants.

Finally, Figure 16 shows the average peak-to-peak ranges (A-P2P) for the participants' top/bottom (T/B) and left/right (L/R) motions at the head and mid-hip. Although there were differences among the participants and differences between not using the heel-raising unit, using the simple unit and the evolved unit regarding the order of A-P2P, the differences appeared random and showed no obvious trends.

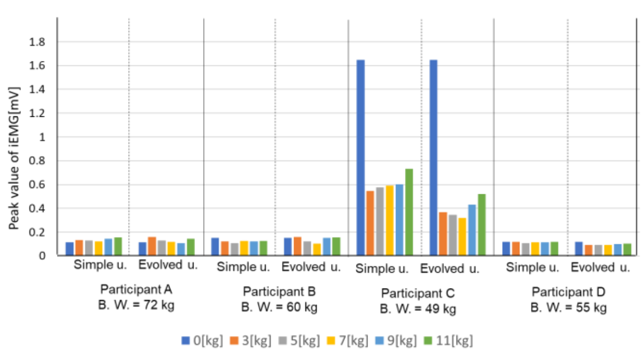


Figure 13. Measured iEMG vs. spring stiffness for each participant.

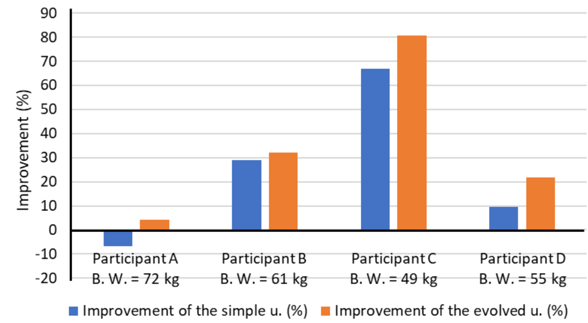


Figure 14. Improvement rate due to heel-raising unit.

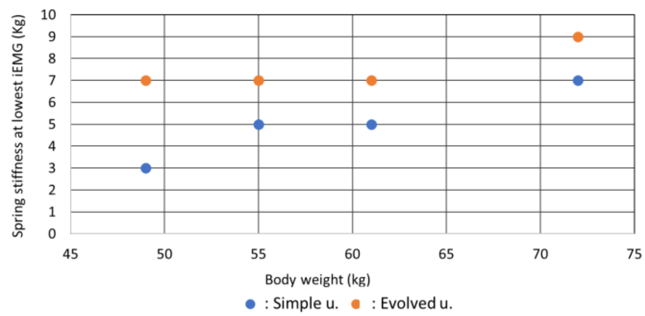


Figure 15. Spring power at lowest iEMG vs. participant body weight.

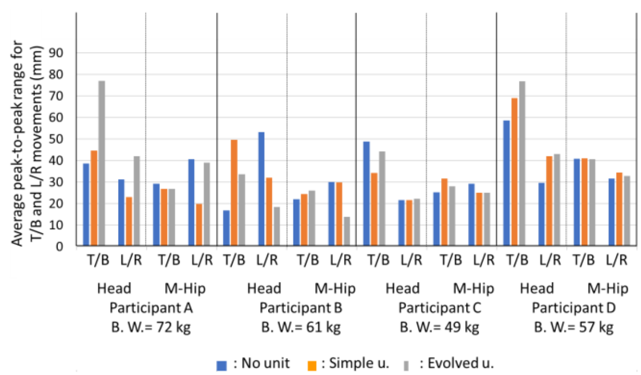


Figure 16. Average peak-to-peak range for T/B and L/R motion over two steps [mm].

The results also demonstrate that the evolved unit had little effect on the walking posture as much as the simple unit did

V. CONCLUSIONS

It is difficult for elderly people to raise their heels because of their low muscle power. As a result, most shuffle their feet when walking and sometimes stumble or trip over something and fall down. We have thus spent nearly 10 years studying how to prevent such falls and developing a heel-raising unit to compensate for low muscle power. In this work, we developed an evolved heel-raising unit, in which a spring in the unit is released when the foot shifts from the whole shoe contacting the floor to the heel rising. Experimental results with the evolved unit demonstrated that its assistance effect was notably better than that of our

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previous unit, whose structure did not control the spring-release timing. Moreover, the evolved unit had little effect on the walking posture as much as the previous one had. This experiment included a low number of participants who were all healthy students. We will thus need to measure data from many elderly people to correctly evaluate the evolved unit's assistance effect. Another problem is that the prototype model in these experiments was heavy and sometimes broke or did not work well, since the model was fabricated with a metal 3D printer. A consumer model will need to be lighter and more reliable, and to have a lower cost. After developing lighter and more reliable model, we would like to measure data from many elderly people to reconfirm assistance effects of the evolved heel-raising unit.

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