Proposal of Spring Assist Unit for Walking Disabilities

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Abstract— **People with diseases, such as hemiplegia, and latter-stage elderly people often have a walking disability, which increases their risk of falling and injuring themselves. The magnitude of the angle velocity during the kicking the floor phase (raising the heel) and swinging the toe forward phase of walking is lower for disabled people than for healthy persons due to lower muscle power. We have developed a spring assist unit that fits in the heel of a shoe and helps disabled people raise their heel when beginning to walk. Experimental results demonstrated that it substantially assists gait walking, that there is a correlation between body weight and optimal spring stiffness, and that the spring assist unit does not affect the person's walking posture.**

Keywords-walking disability; walking assit unit; spring; 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, or latter-stage elderly, often have walking disabilities, which increase their risk of falling, and consequently injuring themselves [2].

Our study of comparing the walking gait between hemiplegia patients and healthy students shows that the magnitude of the angle velocity during the kicking the floor phase (raising the heel) and swinging the toe forward phase of walking is lower for disabled people than for healthy persons; this is because of the lower muscle power for hemiplegia patients compared to the healthy students [3]. This means that assisting with raising the heel and swinging the toe forward while walking could help disabled people and could enable them to have a close to normal gait.

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]) are provided for foot amputees to improves their gait so that it is close to a normal gait. They help the wearer raise their heels and take their toes off. Unfortunately, such prostheses cannot assist walking disabilities.

We previously developed a prototype shoe in which a coil spring was built into the heel part of the shoe, and a leaf spring was built into the half backward of sole. Experimental results demonstrated that it reduced the magnitude of muscle power needed to raise the heels and swing the toes forward. They also demonstrated that there would be a correlation between body weight and optimal spring stiffness, and that the spring assist unit would affect walking posture.

We have now developed a spring assist unit that is built into the heel of a shoe. Experimental results using springs with four different degrees of stiffness demonstrated that there was surely a correlation between body weight and optimal spring stiffness, and that the unit did not affect walking posture, unlike the prototype shoe. This means that the assist unit does not have any negative effects.

After introducing two kinds of foot prosthesis, the SACH and ESAR feet, in Section II, we describe in Section III the differences in gait between a hemiplegia patient and a healthy person to clarify the characteristics that need to be addressed. A prototype shoe into which a coil spring and a leaf spring are built and the experimental results are described in Section IV. The structure of the spring assist unit and the experimental results are described in Section V. Section VI concludes with a summary of the key points.

II. WALKING ASSISTANCE MECHANISM IN PASSIVE FOOT PROSTHESIS

Since there are no walking assistance devices for walking disabilities, such as latter-stage elderly people and hemiplegia patients, walking assistance devices for foot prosthesis are introduced in this section. There are two types of walking assistance mechanisms in passive foot prosthesis.

The SACH foot [6], shown in Figure 1, 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 action of the SACH foot 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. Its primitive form was developed toward the end of the 1800s.

The ESAR foot, shown in Figure 2, has weak push-off power and adequate roll-over shape of the foot, which increases the energy dissipated during the step-to-step transition in gait. Wezenberg et al. reported that the ESAR foot was more effective than the SACH foot in reducing metabolic energy while walking [7], and Houdijk reported that it improved the step length symmetry [8].

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

Figure 2. Examples of ESAR Foot (Vari-Flex, Össur, Iceland).

III. DIFFERENCES IN GAIT BETWEEN HEMIPLEGIA PATIENT AND HEALTHY PERSON

We analyzed the walking gait cycles of unimpaired people and those with disabilities to walk using a Wearable Device (WD) and a KINECT to detect warning signs of falls [3]. Every walking disability in this experiment had one-side paralysis, and trained periodically at a rehabilitation facility. We experimentally measured the output data of an acceleration sensor and gyroscope sensor in a WD mounted on the front of a shoe to estimate the kicking power and change of angle between a foot and the floor as shown in Figure 3. In this measurement, Smart watch 3, SONY was used as a WD.

Figures 4 and 5 show examples of changes in acceleration, angle velocity, and angle for an unimpaired participant and one with a walking disability, respectively. Data for two steps are plotted. Each flat period (roughly the center period) in these figures represents when the entire shoe sole touched the floor. The maximum angle velocity at timing A indicates the kicking power when raising the heel, and the minimum angle at timing B indicates the angle to the floor at terminal swing.

The lower angle velocity at A in Figure 4 is about 420 deg./sec. On the other hand, the higher angle velocity at A in Figure 5 is about 250 deg./sec. Thus, the participant with a walking disability clearly has a weaker kicking power when raising their heel compared with that of the unimpaired participant, indicating a clear difference in terms of gait.

The higher angle at B in Figure 4 is about -18 deg. On the other hand, the lower angle at B in Figure 5 is about -8 deg. And, the swinging speed of walking disability is slower than that of healthy participant. Thus, the participant with a walking disability expressed difficultly when raising their toe at the terminal swing phase.

(a) WD: SmartWatch 3, Sony (b) WD mounted on foot

Figure 3. Measuring device and WD mounting method.

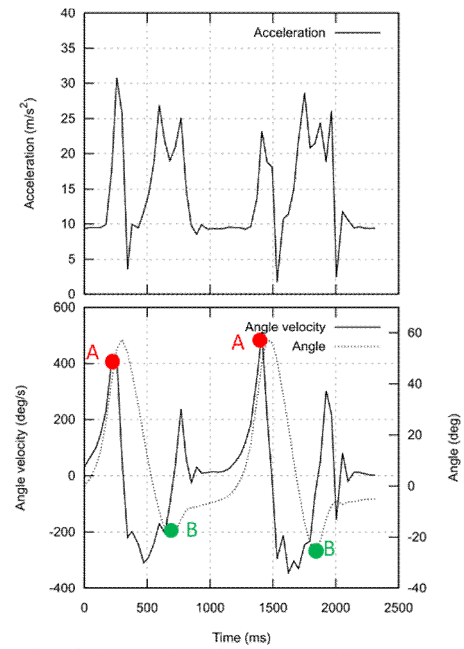


Figure 4. Angle velocity, angle, and acceleration for unimpaired participant.

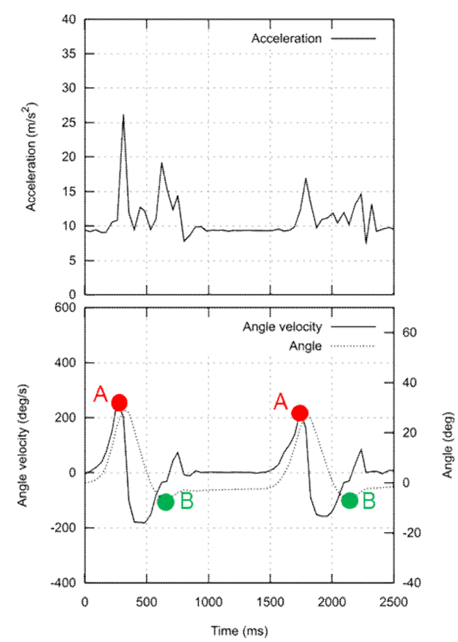


Figure 5. Angle velocity, angle, and acceleration for participant with walking disability.

TABLE I. ANGLE VELOCITY AT THE TERMINAL STANCE

Participant	Average (deg./s)	SD (deg./s)
Unimpaired participant	509.36	18.91
Participant with walking disability	342.06	86.52

TABLE II. ANGLE AT THE TERMINAL SWING

Participant	Average (deg.)	SD (deg.)
Unimpaired Participant	-17.76	8.02
Participant with disability	-7.45	8.02

Tables I and II list the averages and Standard Deviations (SDs) of measured data for angle velocity at timing A and angle at timing B. The angle velocity at timing A is clearly different between unimpaired participants and those with walking disabilities. There is a big difference between them in the angle at timing B; however, this value would have sometimes overlapped each other.

IV. PROTOTYPE OF SHOE TO ASSIST PEOPLE WITH WALKING DISABILITIES

As described in Section III, people with a walking disability, such as those who suffer from hemiplegia, clearly have a weaker kicking power when raising their heel and swing power when swinging their toe forward. We have developed a shoe, shown in Figure 6, that assists with walking disabilities. This shoe has a coil spring and leaf spring to enable a user to easily raise their heel. The spring force of the coil spring is 15 kg. The shoe has a roller to avoid the toe accidentally tripping.

We compare the kicking power (angle velocity) when the heel is raised between a normal shoe and our proposed assist shoe worn by a stroke patient. The data is shown in Figure 7. The kicking power with the assist shoe is lower and more stable than that with a normal shoe.

We then measured a group of 8 students who were asked to walk as if they had a disability while wearing a normal shoe and the assist shoe. Measured data is shown in Figure 8. In every participant except one, their kicking power with the assist shoe was lower and more stable than that with the normal shoe. Authors also examined, and sensed that the shoe compensated to raise their foot slower with weaker power than the normal shoe and its compensation power was stable. Measured data in Figures 7 and 8 indicate the above senses.

We measured the integrated ElectroMyoGram (iEMG) readings for two walking disabilities to confirm the effect of the assist shoe. We used the wireless EMG logger from Logical Product Corporation [9]. The wireless EMG sensors were attached to the gastrocnemius of the right leg as shown in Figure 9. The sampling rate was 500 Hz. Measured data is shown in Figure 10. The results for the assist shoe are lower than those with a normal shoe for both people. The compensation effect of the proposed assist shoe is also confirmed with the iEMG.

Figure 6. Assist shoe prototype.

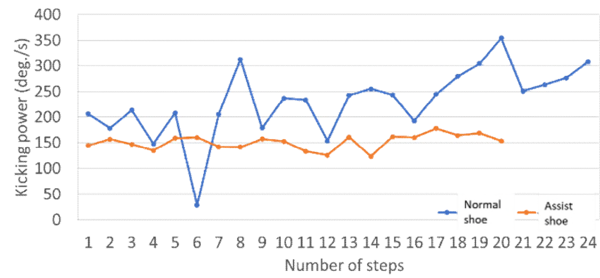


Figure 7. Kicking power when heel is raised with normal and proposed assist shoes for a stroke patient.

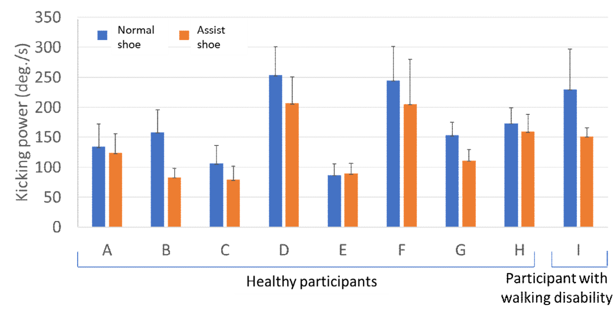


Figure 8. Kicking power when heel is raised with normal and proposed assist shoes.

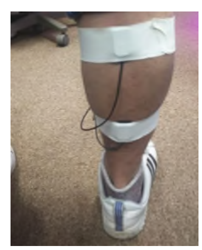
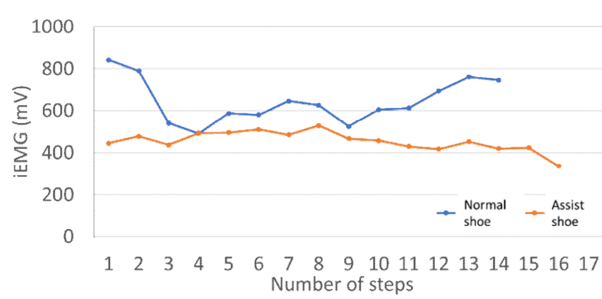


Figure 9. EMG sensors placement.



(1) Person with walking disability #1

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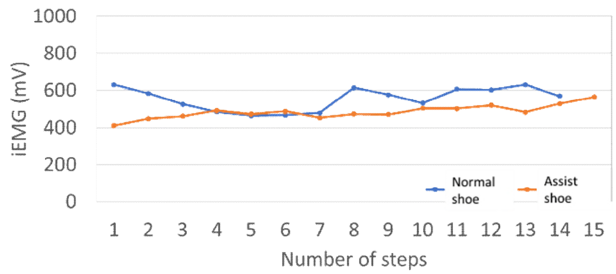


(2) Person with walking disability #2

Figure 10. iEMG data when heel was raised.

It is clear that the proposed shoe compensates for muscle weakness. However, most evaluators including authors felt that the timing to generate a spring reaction force is too early to walk smoothly; the timing at which the knee comes out in front of the ankle is best, and they had to change their gait motion to use a spring power effectively. And, we noticed that there would be a correlation between the body weight and most effective spring power, and would affect the walking posture.

The prototype shoe has a toe roller. However, as it is difficult to have walking disabilities intentionally trip over an obstacle, we could not quantitatively evaluate it.

V. SPRING ASSIST UNIT FOR WALKING DISABLED PEOPLE

A. *Structure*

As described in Section IV, every participant felt that the timing for generating spring reaction force was too early for walking smoothly. We thus focused on clarifying the correlation between body weight and optimal spring stiffness; the effect on walking posture; and developed the spring assist unit shown in Figure 11. Its mechanism is very simple as it comprises only a conical coil spring and a V-shaped attachment cover. We adopt the conical spring to be thinner when stepping on of which spring power is 3, 5, 9, and 11 Kg. The attachment cover is made of thin stainless steel.

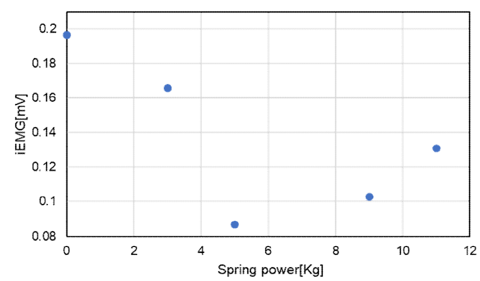
Figure 11. Two views of spring assist unit (heel-up spring).

B. *Assistance effect*

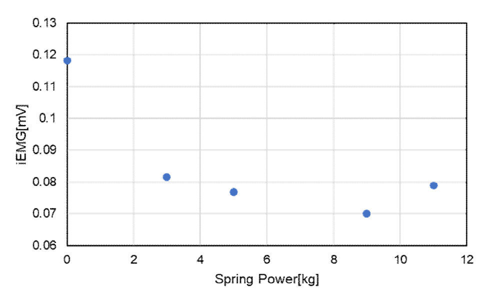
The prototype assist shoe shown in Figure 6 had a coil spring and a leaf spring and was made for the right foot. In contrast, the spring assist unit shown in Figure 11 was built into the heel part of the right and left shoes, as shown in Figure 12. To measure the assistance effect, we had the participants wear shoes with each of the spring stiffnesses and walk straight for 6 m while we measured the iEMG. We also measured the motions of the head and mid-hip to analyze the effects on walking posture. Wireless EMG sensors were attached to the gastrocnemius of the right leg, as shown in Figure 9. The participant's posture was measured using a MS-Kinect [10]. The participants were ten healthy students. In the near future, we plan to measure the same data for persons with a walking disability.

Examples of the measured iEMG vs. spring stiffness for two participants (A and B weighing 57 and 70 kg) are shown in Figure 13. The iEMG values are lower for every spring stiffness than without the spring assist unit. The value was the lowest at the specified spring stiffness. The value for Participant A was lowest at 5 kg, and that for B was lowest at 9 kg. The spring stiffness at the lowest iEMG vs. participant body weight is shown in Figure 14. The spring stiffness magnitude at the lowest iEMG is linearly bigger, a participant gets more weight.

Figure 12. Pair of shoes with built-in spring assist units.



(a) Participant A (weight 57 kg)



(b) Participant B (weight 70 kg)

Figure 13. Examples of measured iEMG vs. spring stiffness.

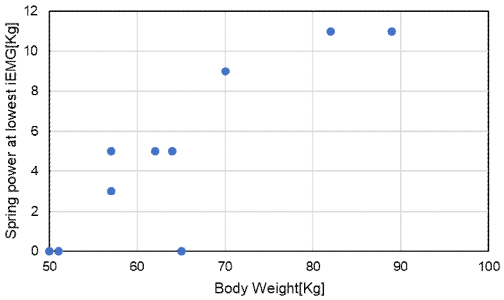
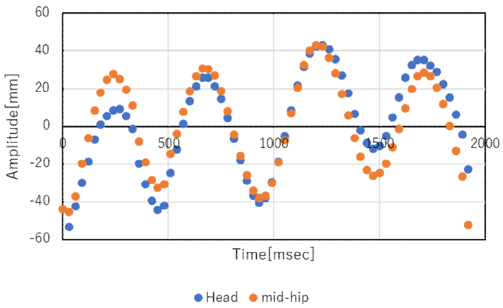
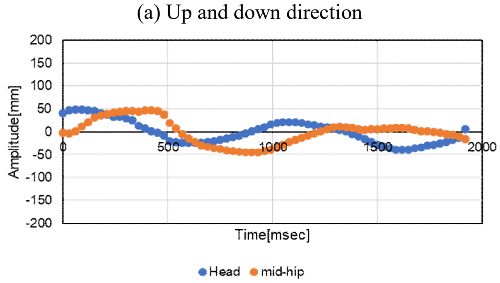


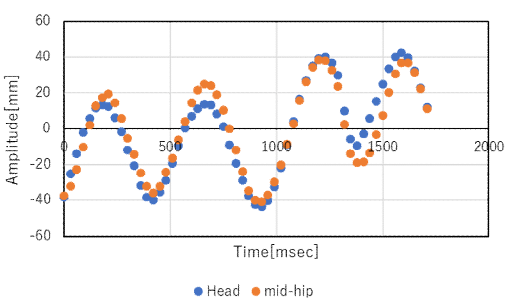
Figure 14. Lowest iEMG vs. participant body weight.



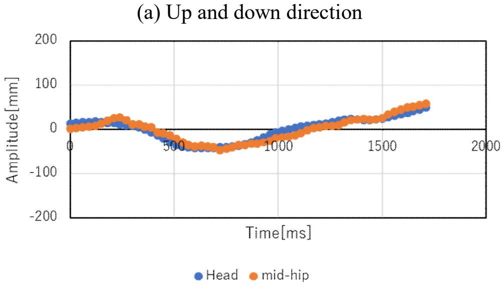
(a) Up and down direction



(b) Right and left direction
Figure 15. Motion of Participant C without a spring assist unit.



(a) Up and down direction



(b) Right and left direction
Figure 16. Motion of Participant C with a spring assist unit.

The measured positions of the head and mid-hip for Participant C are shown in Figure 15 for walking without a spring assist unit and in Figure 16 for walking with the most effective spring assist unit. The changes in the up and down motion are shown in graph (a), and the changes in the right and left motion are shown in graph (b). Without a spring assist unit, the up and down motion of the head and mid-hip clearly changes like a sine wave, as shown in (a) of Figure 15. The right and left motion of the head and mid-hip also changes, but not clearly like a wave as shown in (b) of Figure 15; and the cycle period does not differ from that of the up and down motion. There are not big differences between without a spring assist unit and with a spring assist unit

The average range of each step's peak in the right and left motion (LR) and up and down motion (UD) for each participant is shown in Table III. Although there are differences between participants and spring stiffnesses, the differences are random, with no obvious patterns.

TABLE III. AVERAGE RANGE OF PEAK TO PEAK IN LR AND UD MEASURED OVER TWO STEPS [MM]

Spring power			0[Kg]	3[Kg]	5[Kg]	9[Kg]	11[Kg]
Paticipant A	Head	LR	47.28	34.10	28.54	21.49	30.29
		UD	36.04	33.48	41.07	44.84	42.06
	Mid-hip	LR	32.20	17.11	22.65	17.94	17.44
		UD	29.66	25.80	28.71	22.32	33.61
Paticipant B	Head	LR	47.51	72.87	76.96	71.06	81.35
		UD	13.46	14.68	16.01	18.51	19.18
	Mid-hip	LR	18.83	24.34	22.50	26.19	29.52
		UD	25.87	26.81	28.95	22.72	25.00
Paticipant C	Head	LR	65.83	57.55	56.28	69.76	69.74
		UD	67.89	58.62	62.61	66.90	66.97
	Mid-hip	LR	33.35	33.74	29.23	35.00	32.37
		UD	60.87	67.42	68.07	65.31	54.42
Paticipant D	Head	LR	38.53	41.79	48.00	30.45	51.25
		UD	34.61	37.23	34.01	27.94	30.43
	Mid-hip	LR	31.46	35.64	44.38	23.28	40.88
		UD	38.68	48.60	45.03	45.38	26.01
Paticipant E	Head	LR	34.44	53.31	77.44	63.40	67.88
		UD	30.45	23.43	36.62	21.16	34.64
	Mid-hip	LR	39.86	68.98	71.58	79.92	83.73
		UD	38.71	38.20	34.67	32.90	37.70

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A participant who tested the assist shoe shown in Figure 6 and the pair of assist shoes shown in Figure 12 commented that "I had to step on the shoe to walk smoothly in the shoe shown in Figure 6, whereas I did not feel any effect of the spring units when walking with the shoes shown in Figure 12. I could walk smoothly without any additional actions."

We conclude that the spring assist unit does not affect walking posture.

VI. CONCLUSION

It is difficult for people with walking disabilities to raise their heels because their muscle power is lower. Therefore, most of them shuffle their feet when walking and sometimes stumble over something and fall down. The spring assist unit we developed for walking disabilities enables them to easily raise their heels and walk smoothly. The iEMG was measured to analyze the assistance effect. The iEMG values for every spring stiffness were lower than those without the spring assist unit. The iEMG value was the lowest at the specified spring stiffness; the magnitude of the spring stiffness at the lowest iEMG was linearly bigger and the body weight was greater. These results demonstrate the assistance effect of the spring assist unit and that there is a linear correlation between body weight and the optimal spring stiffness.

We also measured the position of the head and mid-hip with and without the spring assist unit for spring stiffnesses of 3, 5, 9, and 11 kg. There were differences among the participants and among the spring powers, including no spring. However, the differences were random, without any obvious patterns between them. The results also demonstrate that the spring assist unit does not affect walking posture.

We would like to launch the commercial version in a near future.

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