Spring Assist Unit for Individuals with Walking Disabilities

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***Abstract*-Latter-stage elderly people, as well as people with diseases such as hemiplegia, often have a walking disability, which increases their risk of falling and suffering injuries. The magnitude of the angular velocity during both the heel rise phase of walking (kicking the floor) and swing phase (swinging the toe forward) is lower for disabled people compared to healthy individuals, owing to their 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 demonstrate that it substantially assists in walking and normalizing gait. We also report a correlation between body weight and optimal spring stiffness, and that the spring assist unit does not affect the individual's walking posture.**

Keywords-walking disability; walking assist unit; spring; walking posture; muscle power.

I. INTRODUCTION

This paper is an extension of the paper initially presented at the Twelfth International Conference on eHealth, Telemedicine, and Social Medicine [1]. In this paper, we discuss additional experiments that we conducted with elderly people to show the effectiveness of our proposed system.

As the percentage of elderly people worldwide increases [2], so too will the number of functionally impaired people, such as those with hemiplegia. People with such diseases, as well as latter-stage elderly people often have walking disabilities that increase their risk of falling and consequently suffering injuries [3]. Tao *et al.* reviewed gait analysis using various wearable sensors [4]. They indicated that gait analysis using wearable sensors is useful for estimating fall risk, although they did not describe the gait of people with walking disabilities.

Our study compared the walking gaits of hemiplegia patients and healthy students. We showed that the magnitude of the angular velocity during the phase in which the subject kicks the floor (heel rise phase) and swings the toe forward (swing phase) is lower for disabled people than for healthy counterparts. This is likely due to the lower muscle power of hemiplegia patients [5]. Thus, assisting the raising of the heel and swinging the toe forward while walking could help disabled people walk with a gait closer to that of younger and healthier individuals.

The Solid-Ankle Cushion Heel (SACH) foot (1D10, Ottobock, Germany) [6]) and the Energy Storage And Return (ESAR) foot (Vari-Flex, Össur, Iceland)[7]) are provided for foot amputees to improve their gait. These prosthetics help the wearer raise their heel and lift their toes. Unfortunately, such devices cannot assist individuals with only walking disabilities.

Mooney *et al.* reported result showing that the ankle exoskeletons does not exclusively reduce positive mechanical power at the ankle joint, but also mitigates positive power at the knee and hip as show in experiments with six participants without walking disabilities [8]. Leclair *et al.* developed an unpowered ankle exoskeleton using flexible air spring [9], although they did not conduct demonstration experiments for the people with disabilities. We also previously developed a prototype shoe in which a coil spring was built into the heel and a leaf spring was built into the back half of the sole. Experimental results demonstrated that our device reduced the magnitude of muscle power needed to raise the heel and swing the toes forward. We also demonstrated that there may be a correlation between body weight and optimal spring stiffness and that a spring assist unit would affect walking posture.

We have thus developed a new shoe incorporating a spring assist unit built into the heel. Using springs with 3 kg to 11 kg of stiffness, our results demonstrate that there is surely a correlation between body weight and optimal spring stiffness and that the spring unit did not actually affect walking posture, unlike the previous prototype shoe. The main scientific contribution of this paper is that our newly developed device reduced the magnitude of muscle power without negative effects. This contribution was shown through an experiment targeting elderly people.

After introducing two kinds of foot prostheses, 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 in order to clarify the necessary characteristics to be addressed. A prototype shoe that incorporates a coil spring and a leaf spring is then described and evaluated as a walking assistance device in Section IV. The structure of the spring assist unit and the preliminary experimental results gathered from healthy young students are described in Section V. We then validate the effectiveness of our system using an experiment with elderly people in Section VI. Section VII concludes with a summary of the key points.

II. WALKING ASSISTANCE MECHANISM WITH PASSIVE FOOT PROSTHESES

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

The SACH foot [10], 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 keel-shaped structural core 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 an adequate rollover shape for 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 [11], and Houdijk *et al*. reported that it improved step length symmetry [12].

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

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

Figure 3. Measuring device and WD mounting method.

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

III. DIFFERENCES IN GAIT BETWEEN HEMIPLEGIA PATIENTS AND HEALTHY PEOPLE

We analyzed the walking gait cycles of healthy individuals and those with walking disabilities using a wearable device (WD) [3]. The individuals with walking disabilities had one-side paralysis and periodically received treatment at a rehabilitation facility. We measured the output data from an acceleration sensor and a gyroscope sensor in a WD mounted on the front of a shoe to estimate the kicking power and change of angle between the foot and the floor, as shown in Figure 3. For this measurement, we used a Sony SmartWatch 3 as a WD.

We measured the angular velocity around the X-axis and the acceleration of Y-axis as shown in the Figure 3(b). The measured data were stored every 40 ms as JSON format in the memory of a smartphone. Figures 4 and 5 show examples of changes in acceleration, angular velocity, and angle for a healthy participant and one with a walking disability, respectively, over the course of two steps. Each flat period (roughly the center period) in these figures represents when the entire sole of the shoe touched the floor. Moving average lines of 120 ms are plotted in Figures 4 and 5. The maximum angular velocity at point A indicates the kicking power when

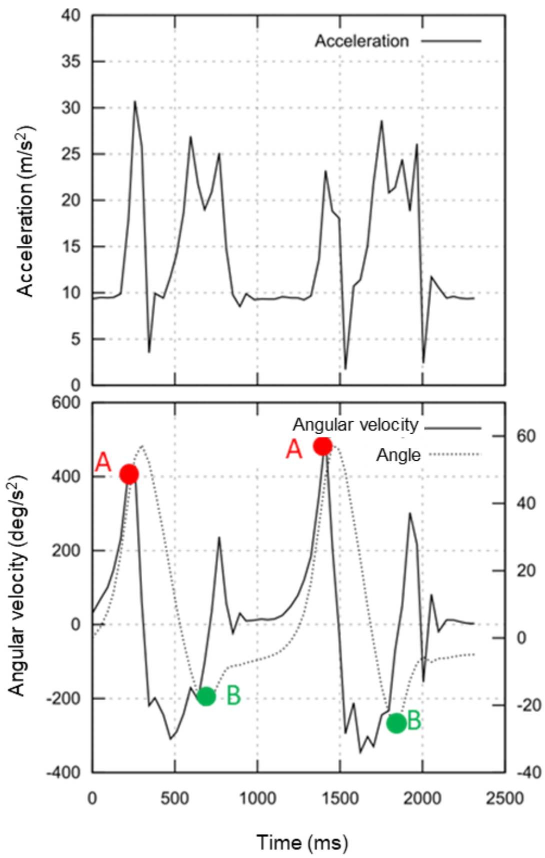


Figure 4. Angular velocity, angle, and acceleration for the unimpaired participant.

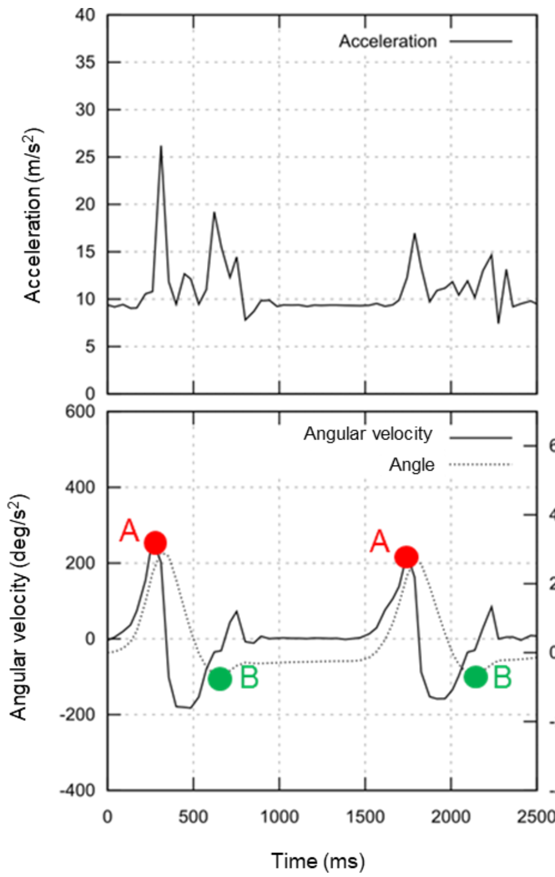


Figure 5. Angular velocity, angle, and acceleration for the participant with a walking disability.

raising the heel, and the minimum angle at point B indicates the angle to the floor at terminal swing.

The lowest angular velocity at point A of the unimpaired participant in Figure 4 was approximately 420° s^{-1}, whereas the highest angular velocity at point A of the participant with a walking disability was only 250° s^{-1}. Thus, the participant with a walking disability clearly had a weaker kicking power when raising their heel compared with that of the healthy individual, indicating a clear difference in gait.

The highest angle at point B of the unimpaired person in Figure 4 was approximately −18°, whereas the lowest angle at point B of the participant with a walking disability in Figure 5 was approximately −8°. Thus, the swinging speed of the individual with a walking disability was also slower than that of the healthy participant. Similarly, the participant with a walking disability had difficulty raising their toe during the terminal swing phase.

Tables I and II list the averages and standard deviations (SD) of the measured data for angular velocity at point A and angle at point B. The angular velocity at A is clearly different for unimpaired participants and those with walking disabilities. There is also a measurable difference between them in the angle at point B. However, the ranges of these values will sometimes overlap.

IV. PROTOTYPE SHOE TO ASSIST PEOPLE WITH WALKING DISABILITIES

As described in Section III, individuals 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 first developed a shoe, shown in Figure 6, that assists individuals with walking disabilities. This shoe has a coil spring and a leaf spring that enables the wearer to raise their heel more easily. The spring force of the coil spring is 15 kg and the shoe has a roller to avoid the toe accidentally tripping.

We compared the kicking powers (angular velocity) when the heel is raised between a normal shoe and our proposed assist shoe when worn by a stroke patient. The data are shown in Figure 7, wherein the vertical and horizontal axes show the kicking power and the number of steps taken by the individual, respectively. The kicking power with the assist shoe was lower and more stable than that with the normal shoe.

For a test trial, we then conducted an experiment with a group of eight students who were asked to walk as if they had a disability while wearing a normal shoe and the assist shoe. The measured data are shown in Figure 8 with the kicking power measured by the sensor for each of the participants (*i.e.*, eight healthy students and a disabled person). The blue and orange bars in Figure 8 show the average kicking power of participants wearing the normal and assist shoes, respectively. The highest point of the error bars represents the maximum kicking power for each individual. For every participant except one, the kicking power with the assist shoe was lower and more stable than that with the normal shoe. We also examined and observed that the shoe helped them to raise the foot slower using less power than the normal shoe and that this compensation power was stable. Measured data in Figures 7 and 8 confirm these observations.

We also took integrated electromyogram (iEMG) measurements for two individuals with walking disabilities to confirm the effect of the assist shoe. We used the wireless EMG logger from the Logical Product Corporation [13]. The wireless EMG sensors were attached along the gastrocnemius muscle at the back of the calf and thigh of the right leg, as shown in Figure 9. Measured data, taken with a sampling rate of 500 Hz, is shown in Figure 10 wherein the vertical and horizontal axes denote the peak of the EMG signal measured by the iEMG sensor for each step and the number of steps, respectively. The iEMG readings for the assist shoe were lower than those for the normal shoe for both individuals. Thus, the compensation effect of the proposed assist shoe was also confirmed using iEMG.

TABLE I. ANGULAR 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

Figure 6. Assist shoe prototype.

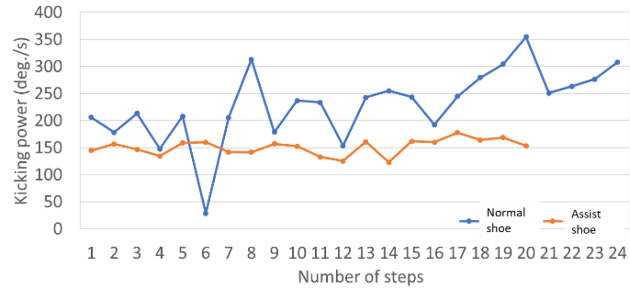


Figure 7. Kicking power when heel is raised with normal shoe (blue) and proposed assist shoes (orange) from a stroke patient.

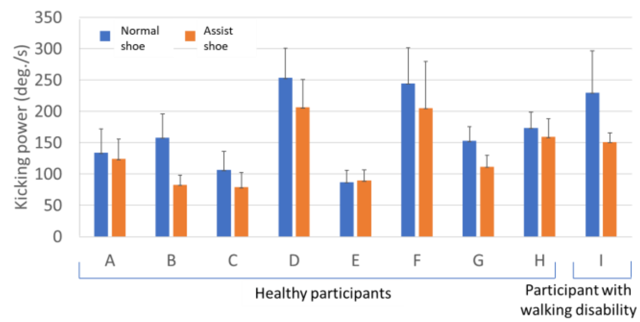
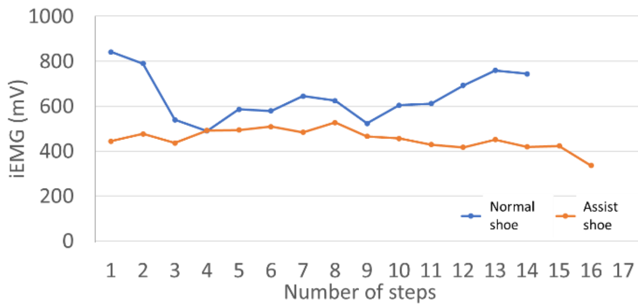


Figure 8. Kicking power when heel is raised with normal (blue) and proposed assist shoes (orange).

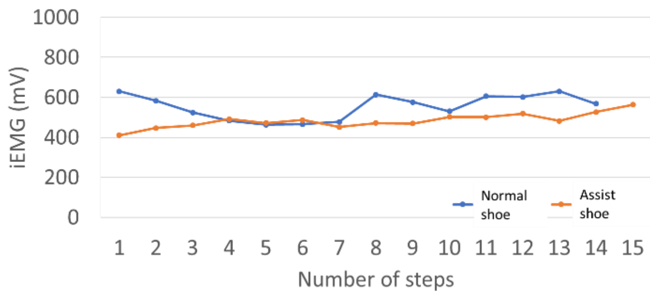
It is clear that the proposed shoe compensates for muscle weakness. However, most evaluators, including the authors, felt that the timing to generate a spring reaction force was too early for them to walk smoothly; the timing at which the knee comes out in front of the ankle is best, and participants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

The prototype shoe also features a toe roller. However, it is difficult to have individuals with walking disabilities intentionally trip over an obstacle, so we could not quantitatively evaluate this feature.

Figure 9. Placement of EMG sensors.



(1) Person with walking disability #1



(2) Person with walking disability #2

Figure 10. iEMG data when heel is raised from participants with walking disabilities.

V. SPRING ASSIST UNIT TO HELP DISABLED PEOPLE WALK

As described in Section IV, many participants felt that the timing for generating the spring reaction force was too early for them to walk smoothly. Thus, to address this, we proposed new shoes.

A. *Structure*

We focused on clarifying the correlation between body weight and optimal spring stiffness, the effect on walking posture, and developing the spring assist unit shown in Figure 11. The mechanism for this unit is very simple as it comprises only a conical coil spring and a V-shaped attachment cover made of thin stainless steel. We adopt the conical spring from the power of 3, 5, 9, or 11 Kg. The spring is selected so that it is buried in the sole of the shoe when they step.

B. *Preliminary evaluation of assistance effects*

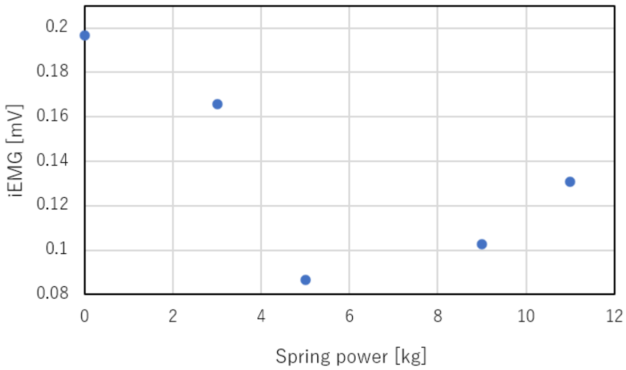
The prototype assist shoe shown in Figure 6 has a coil spring and a leaf spring and was made for the right foot. In contrast, the spring assist unit shown in Figure 11 was built into the heel part of the left and right shoes, as shown in Figure 12. To measure the degree of assistance given by these mechanisms and to evaluate the safety of these mechanisms, we ten students without walking disabilities to wear shoes with different spring stiffnesses and walk straight for 6 m while we measured the iEMG as a preliminary experiment before using participants with walking disabilities. For safety, we used people without walking disabilities as participants first. We also measured the motions of the head and mid-hip to analyze the effects on walking posture. Wireless EMG

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

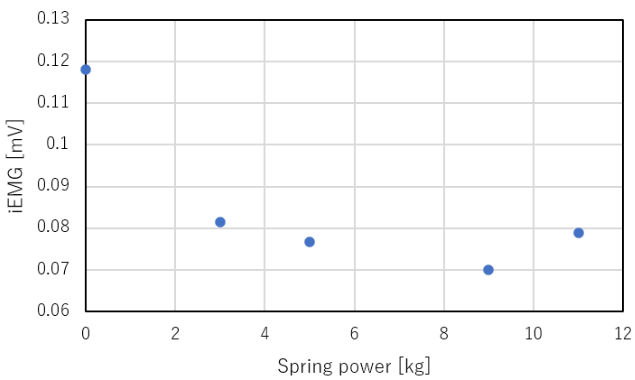
Figure 12. Pair of shoes with built-in spring assist units.

sensors were attached to the gastrocnemius of the right leg, as shown in Figure 9, to gather iEMG data. The participant's posture was measured using an MS-Kinect [14].

Examples of the measured iEMG vs. spring stiffness for two participants (A and B weighing 57 and 70 kg, respectively) are shown in Figure 13. The iEMG values were lower for each spring stiffness compared to the case without the spring assist unit. The value for Participant A was lowest at 5 kg and that for B was lowest at 9 kg. The spring stiffness resulting in the lowest iEMG signal was used to determine a relationship with the participant's body weight. The spring stiffness at the lowest iEMG vs. participant body weight is shown in Figure 14. For example, the lowest iEMG of participant A was at 3 kg of spring power; this is represented as a point at 57 kg of body weight and 3kg of spring power in Figure 14. Visualizing this data allowed us to reveal a linier correlation between participant body weight and the magnitude of spring stiffness at the lowest iEMG reading.



(a) Participant A (weight 57 kg).



(b) Participant B (weight 70 kg).
Figure 13. Examples of measured iEMG vs. spring stiffness.

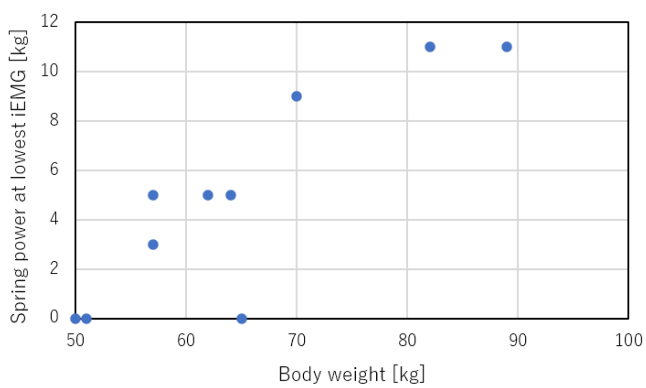
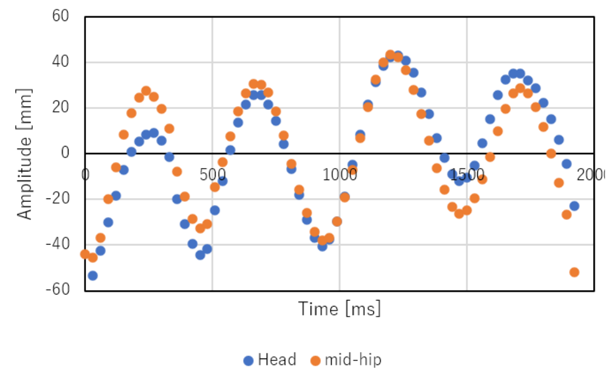
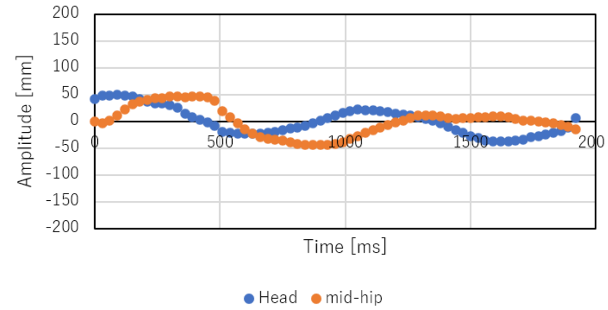


Figure 14. Spring power at lowest iEMG vs. participant body weight.

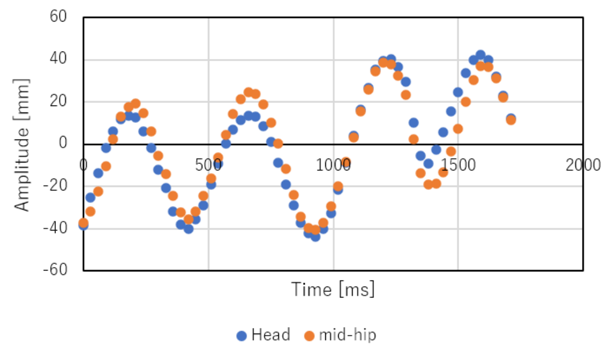
The measured positions of the head and mid-hip for Participant C are shown in Figure 15 for walking without a spring assist unit and in Figure 16 for walking with the most effective spring assist unit. For both Figures 15 and 16, the vertical and horizontal axes denote displacement and elapsed time, respectively. We expressed “amplitude” as the position



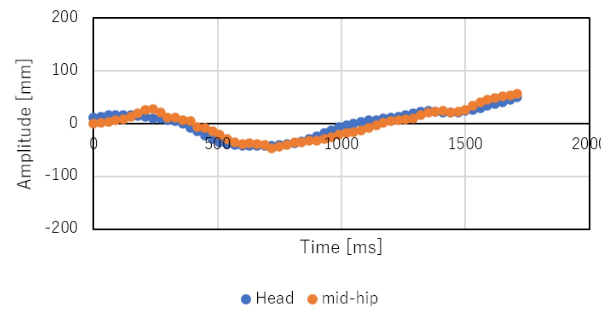
(a) Up and down direction



(b) Left and right direction
Figure 15. Motion of Participant C without spring assist unit.



(a) Up and down direction



(b) Left and right direction
Figure 16. Motion of Participant C with spring assist unit.

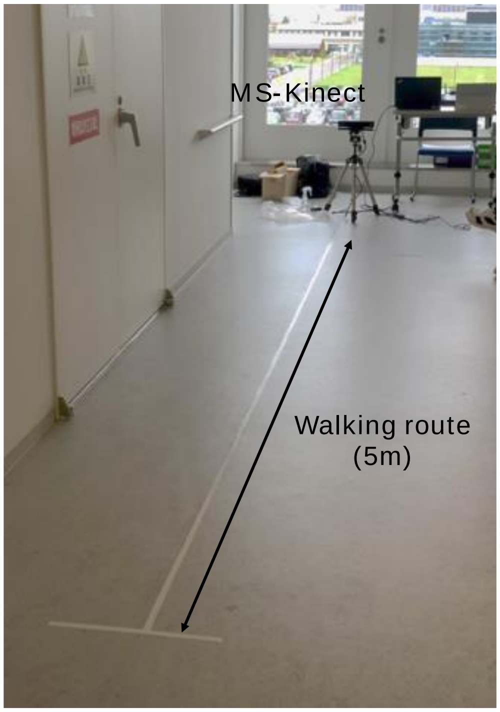


Figure 17. Experimental setting.

Figure 18. An elderly participant walking.

measured by the Kinect in these figures because the data were measured as swinging by the steps. Blue and orange points denote the amplitude of the head and mid-hip, respectively. Changes in the up and down motion (UD) are shown in Figures 15(a) and 16(a) whereas changes in the left and right (LR) motion are shown in Figures 15(b) and 16(b). Without a spring assist unit, the UD motion of the head and mid-hip clearly mimics a sine wave, as shown in Figure 15(a). The LR motion of the head and mid-hip also changes, but apparently not like a wave as in Figure 15(b); and the period for this cycle does not differ from that of the UD motion. Thus, there are no significant differences between participants walking with or without a spring assist unit.

TABLE III. AVERAGE PEAK-TO-PEAK RANGE FOR LR AND UD MOVEMENTS MEASURED OVER TWO STEPS FOR HEALTHY STUDENTS PARTICIPANTS [MM].

Spring power [kg]			0	3	5	9	11
Participant 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
Participant 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
Participant 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
Participant 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
Participant 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

From these figures, we also confirmed the safety of the mechanism in people without walking disabilities.

The average range of each step's peak during both LR and UD motions for participants A through E, who tried springs of 0 kg to 11 kg, is shown in Table III. Although there are differences between participants and spring stiffnesses, they seem to be random with no obvious patterns.

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 thus concluded that the spring assist unit does not affect walking posture.

VI. EVALUATION OF THE SYSTEM BY ELDERLY PEOPLE

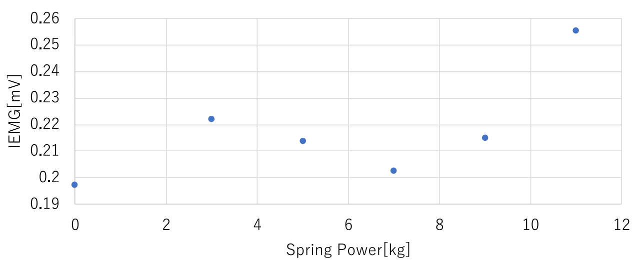
To examine the performance of the assistance device in a more practical environment, we enrolled elderly people for the evaluation because many lack sufficient power to raise their leg, a similar condition to disabled people.

A. *Experimental setting*

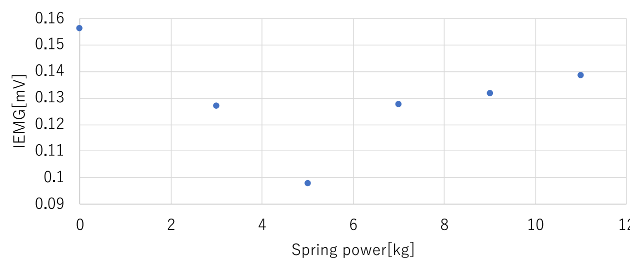
Seven participants named Participant 1 to Participant 7 (ages 79, 78, 84, 76, 76, 87 and 89 years old, respectively) were involved in the experiments. Three participants, Participant 5, 6, and 7, were female and the others were male. Figure 17 shows the experimental setting wherein participants walked straight for 5 m along the white line. Posture and EMG were measured using a Kinect and iEMG on the left leg, as shown in Figure 9. Figure 18 shows an elderly participant walking during an experiment. Participants attached the iEMG to their feet as shown in Figure 18. Each participant walked with our proposed device with and without springs of 3, 5, 7, 9, and 11 kg power, as shown in Figure 11.

B. *Experimental results*

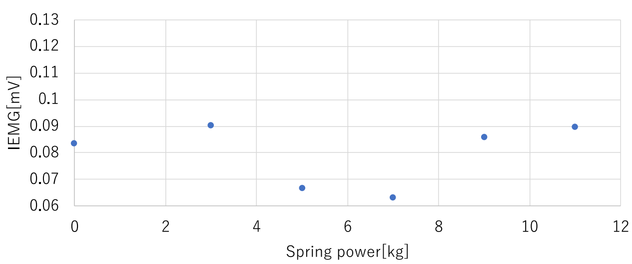
Figure 19 shows the iEMG values vs. the spring power for each participant. The vertical and horizontal axes show the voltage of the iEMG and the spring power used by each participant, respectively. Result shown in Figure 19 (e), (f) and (g) were obtained from the female participants. We observed similar trends as for the experiment involving healthy students (Figure 13). Many participants said that they could walk more easily using the springs. Participant 5 said that it was hard to walk because the shoe did not fit; however, the effectiveness of the spring was still evident. Lower iEMG values were obtained with and without the spring assistance



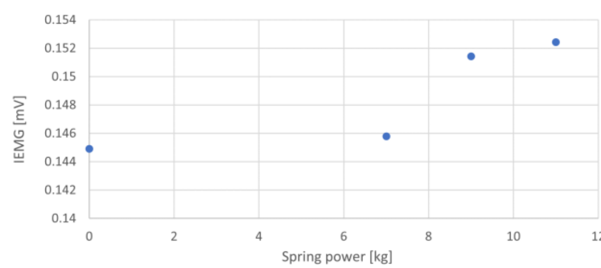
(a) Participant 1 (60 kg, 79 years old)



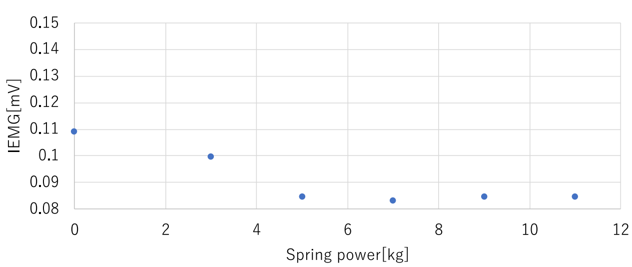
(d) Participant 4 (56 kg, 76 years old)



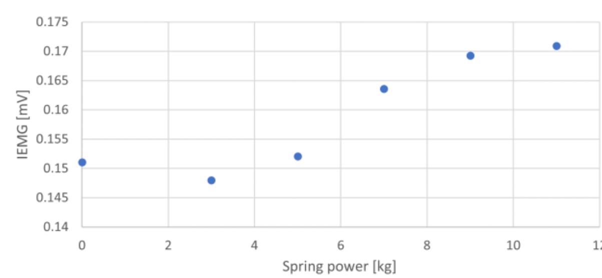
(b) Participant 2 (60 kg, 78 years old)



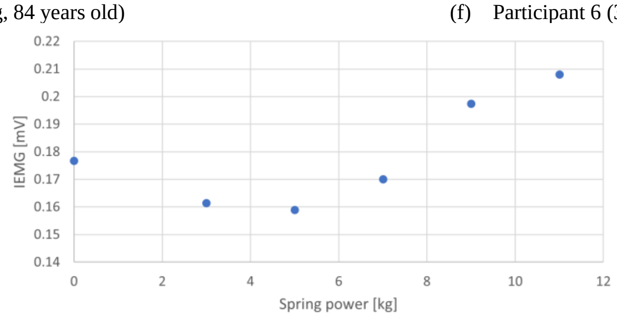
(e) Participant 5 (48 kg, 76 years old)



(c) Participant 3 (60 kg, 84 years old)



(f) Participant 6 (38 kg, 87 years old)



(g) Participant 7 (57.5 kg, 89 years old)

Figure 19. Spring power vs. IEMG for each participant.

devices for both the 80-year-old and 76-year-old participants (Figures 19(c) and (d), respectively).

Figure 20 shows the spring power at the lowest iEMG value vs. body weight for each participant. As for the younger participants (Figure 14), there was a linear correlation between the participant's body weight and the spring power at which the lowest iEMG signal was obtained.

Figures 21 to 25 present the measured positions of the head and mid-hip for Participants 1 to 5, respectively. Each figure shows the posture during walking (a) without a spring assist unit and (b) with the most effective spring assist unit (*i.e.*, 7 kg, 7 kg, 7 kg, 5 kg, 3 kg, 3 kg, and 5 kg for Figures 21 to 27, respectively). We confirmed that not only the UD motion, but also the LR motion of the head and mid-hip changes periodically and that two cycles of UD motion are likely to occur during each period of LR motion. We considered that UD motion occurred with each step whereas LR motion occurred with each right or left step. This trend is different in the younger participants shown in Figures 15 and 16 and is likely due to aging. An especially large LR motion was observed in Participant 3 and the period of Participant 5 was smaller because of the step length. However, there were no significant differences between participants walking without a spring assist unit and those with one, similar to Figures 15 and 16.

The average peak-to-peak range for each step during the LR and UD motions of each participant with varying spring stiffnesses of 0–11 kg is shown in Table IV. Although there are differences between participants and spring stiffnesses, these are again random with no obvious patterns, similar to Table III.

From, this experiment, we could not confirm the difference in the results between gender. Chumanov *et al.* showed that females walked with greater peak hip internal rotation and adduction than males, as well as gluteus maximus activity [15]. However we could not confirmed this trend from Table IV. Ko *et al.*, [16] show that women walked with higher cadence and shorter stride length than men and had less hip range of motion, greater ankle in the sagittal plane, and greater hip in the frontal plane. We could not confirm the gender-based difference from Table IV.

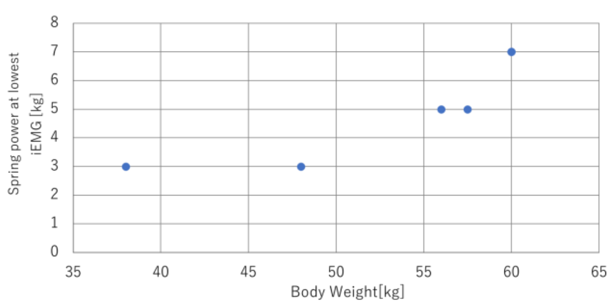
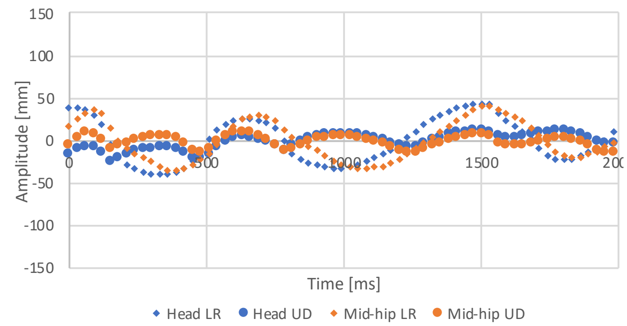
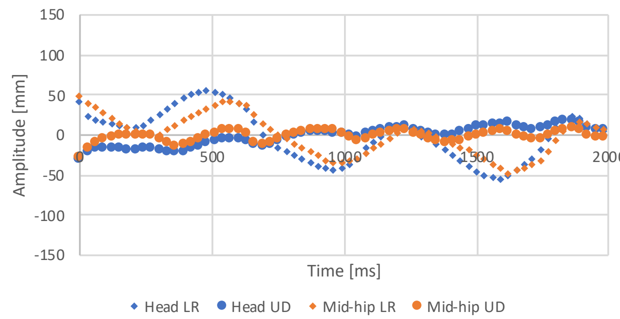


Figure 20. Spring power at lowest IEMG vs. body weight.

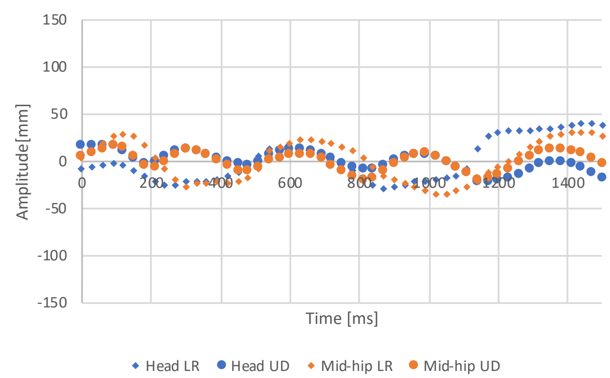


(a) Without spring

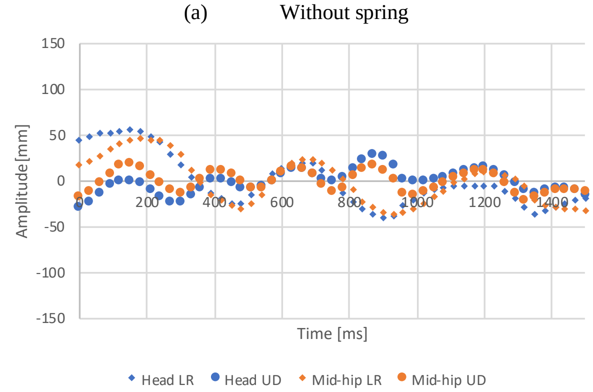


(b) With a spring assist unit (7 kg)

Figure 21. Motion of Participant 1 obtained by the Kinect.

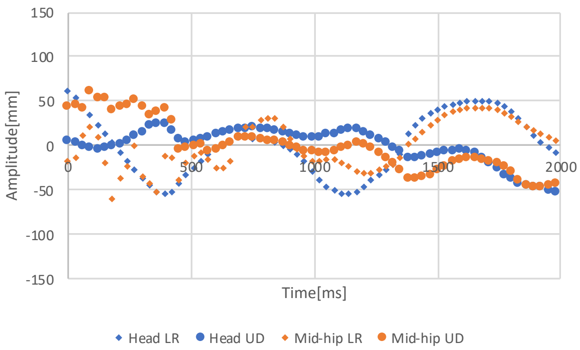


(a) Without spring

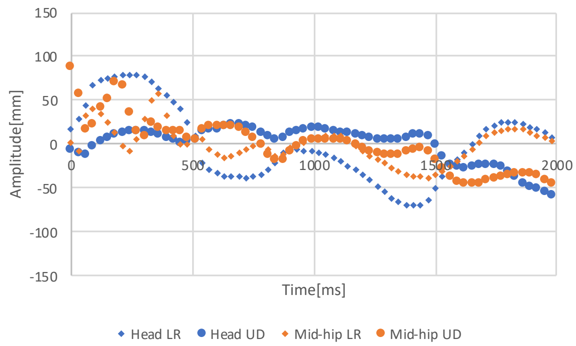


(b) With a spring assist unit (7 kg)

Figure 22. Motion of Participant 2 obtained by the Kinect.

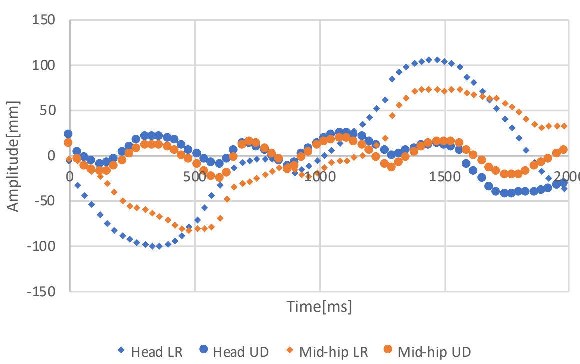


(a) Without spring

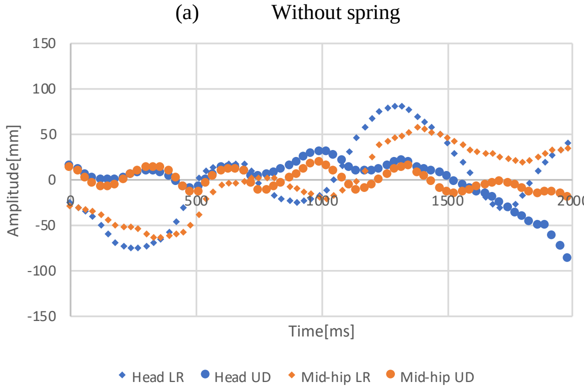


(b) With a spring assist unit (7 kg)

Figure 23. Motion of Participant 3 obtained by the Kinect.

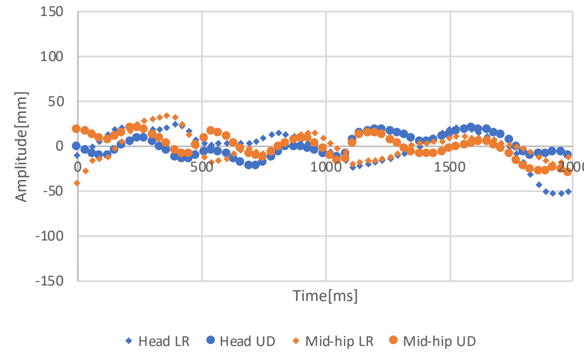


(a) Without spring

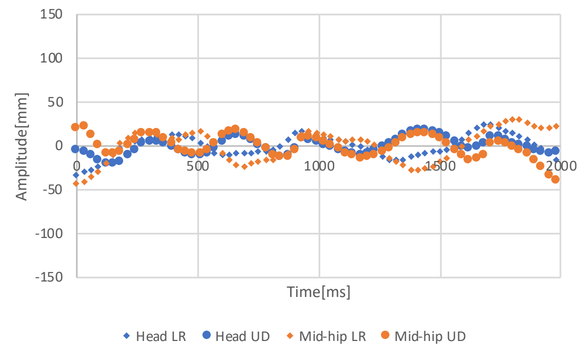


(b) With a spring assist unit (5 kg)

Figure 24. Motion of Participant 4 obtained by the Kinect.

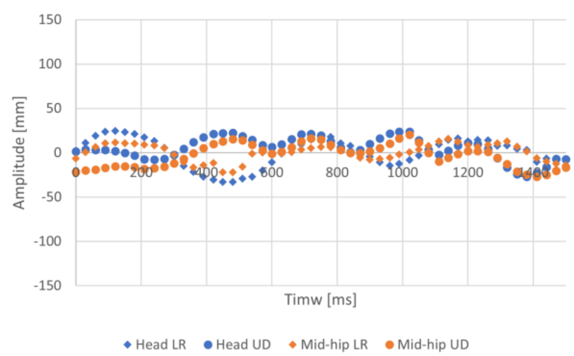


(a) Without spring

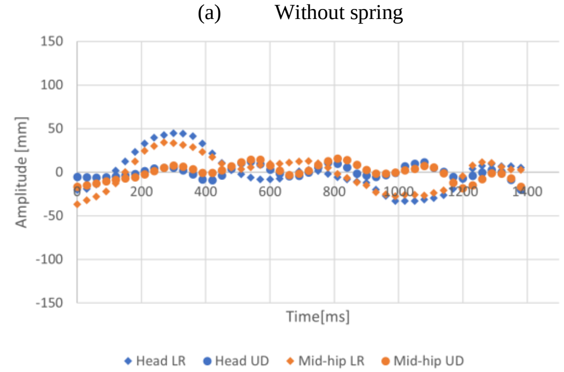


(b) With a spring assist unit (3 kg)

Figure 25. Motion of Participant 5 obtained by the Kinect.

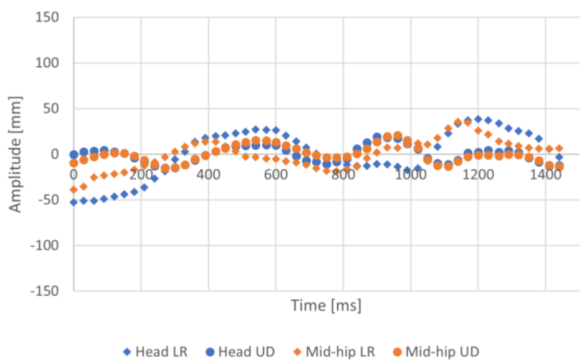


(a) Without spring

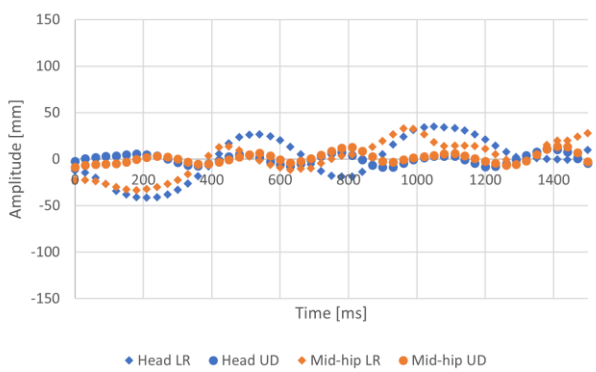


(b) With a spring assist unit (3 kg)

Figure 26. Motion of Participant 6 obtained by the Kinect.



(a) Without spring



(b) With a spring assist unit (5 kg)

Figure 27. Motion of Participant 7 obtained by the Kinect.

VII. Conclusions

It is difficult for individuals with walking disabilities to raise their heels because of their lower muscle power. Therefore, most shuffle their feet when walking and sometimes stumble or trip over something and fall down. The shoes that we developed with the spring assist unit enable people with walking disabilities to raise their heels more easily and walk more smoothly. The electromyogram (EMG) was measured to analyze the efficacy of our assistance device. The EMG values for devices incorporating various spring stiffnesses were all lower than for those walking without the spring assist unit. There was a linear relationship between the spring stiffness at which the lowest EMG signal was observed and the participant's body weight. These results demonstrate the assistive effect of the spring assist unit and that there is a linear correlation between body weight and the optimal spring stiffness of such devices. The spring in the shoes should be provided depending on user's weight.

We also measured the positions of the head and mid-hip with and without the spring assist unit for spring stiffnesses of 3 to 11 kg. Although there were differences between the participants and between the optimal spring powers, including devices with no spring, the differences were random without any obvious trends. The results also demonstrate that the spring assist unit did not affect walking posture. These trends were true for not only healthy young students, but also elderly people. Therefore, spring shoes afford wearers the chance to walk comfortably without affecting the posture in both people without walking disability and elderly people. Spring shoes are also expected to be applicable for people with disabilities when the strength of spring is adopted to the weight of users.

Our future work will include the launch of a commercial version of this spring assistance device.

TABLE IV. AVERAGE PEAK-TO-PEAK RANGE FOR LR AND UD MOVEMENT MEASURED OVER TWO STEPS FOR ELDERLY PEOPLE [MM]
.

Spring power [kg]			0	3	5	7	9	11
Participant 1 (79 years old)	Head	LR	66.88	60.40	57.97	68.29	75.61	65.40
		UD	20.68	10.13	22.21	7.59	13.89	11.83
	Mid-hip	LR	62.42	59.12	51.13	53.55	63.88	57.20
		UD	19.438	16.87	21.83	15.63	13.02	19.04
Participant 2 (78 years old)	Head	LR	41.25	59.01	42.44	56.36	54.62	50.98
		UD	15.82	25.08	19.39	15.36	14.53	18.56
	Mid-hip	LR	58.50	58.97	55.10	58.71	62.14	77.81
		UD	26.25	26.36	25.53	26.58	24.64	26.05
Participant 3 (84 years old)	Head	LR	71.67	91.23	78.08	71.68	82.36	73.21
		UD	13.32	8.30	16.74	26.26	10.41	21.85
	Mid-hip	LR	53.25	44.67	43.69	47.30	65.53	47.34
		UD	24.49	18.28	24.93	30.66	27.17	31.38
Participant 4 (76 years old)	Head	LR	60.62	69.26	91.38	79.01	75.42	76.53
		UD	19.44	23.61	13.56	22.83	20.01	13.70
	Mid-hip	LR	51.08	61.32	66.90	59.02	47.74	58.85
		UD	32.25	31.89	23.30	28.10	26.78	26.75
Participant 5 (76 years old)	Head	LR	30.45	30.99	52.27	45.35	41.29	40.08
		UD	15.28	20.30	17.85	20.30	11.63	10.23
	Mid-hip	LR	31.10	37.87	46.90	34.01	38.90	32.52
		UD	24.56	23.84	22.21	23.24	14.79	17.67
Participant 6 (87 years old)	Head	LR	41.99	36.825	36.95	37.61	38.90	22.61
		UD	22.64	20.76	25.87	19.07	16.52	16.06
	Mid-hip	LR	25.74	33.13	25.26	26.78	30.39	21.17
		UD	20.15	25.156	18.779	21.16	14.14	16.19
Participant 7 (89 years old)	Head	LR	46.55	48.94	41.169	45.0	34.58	39.54
		UD	28.181	14.1	15.61	18.09	14.29	14.01
	Mid-hip	LR	43.29	31.80	31.61	40.55	28.28	32.91
		UD	29.93	29.63	23.97	24.95	13.19	19.95

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