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Evaluation of biomimetic assist suits on cross-slope walking *

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***Abstract*— In cross-slope walking, gait differs between uphill and downhill slopes. Furthermore, because ankle sprains and other injuries are more likely to occur during cross-slope walking, assistance is required to prevent falls, ankle joint sprains, and the removal of load from the muscles involved in knee joint motion. This assistance differs between uphill and downhill slope walking. We developed biomimetic assist suits (BAS) by combining a semi-active knee joint, which assists knee joint motion, and dual axial ankle supporter, which assists ankle motion. BAS was developed to provide knee and ankle support during cross-slope walking. A walking experiment with 14 participants was conducted to verify the effectiveness of BAS. The experimental results showed that BAS could reduce the difference between the maximum and minimum angles in inversion/eversion and plantar dorsiflexion during the swing phase, protecting the ankle joint, limiting the ankle joint angle in the elastic connector, and adjusting the assisting direction mechanism.**

I. INTRODUCTION

Hiking enjoys global popularity, attracting numerous tourists each year [1]. However, it poses potential risks, particularly when hiking on uneven terrain, leading to severe injuries. Between 2009 and 2018, Switzerland recorded 11,220 mountain accidents, of which 45% were caused by falls [2]. Additionally, the incidence of falls increases with age [3-4], posing risks of adverse physical and physiological effects, particularly in the elderly [5]. Numerous ankle-assistive devices have emerged for walking, including, among others, TALOS [6], RE-Gait® [7], and active ankle-foot orthoses [8]. Previously, we developed a dorsiflexion support unit (DSU) using an elastomer-embedded flexible joint (EEFJ) [9]. In a previous study [10], we developed biomimetic knee joints (BKJ), which are polycentric orthoses that assist knee motion during crouching and standing to reduce the misalignment of the center of rotation between the biological knee joint and the supporter. However, these conventional devices primarily target level-ground walking for disabled or frail individuals and are unsuitable for navigating uneven terrains.

In addition, there is less information on the joint kinematic adaptation when walking on cross slopes than on level ground, uphill, or downhill [11-13]. To understand the motion of the lower limb joints, we measured data during cross-slope walking [14-15]. In this study, we analyzed the muscle activation patterns during cross-slope walking using the measured data. The experimental results reveal some differences between the uphill and downhill motions of the knees and ankles. For the 16° and 30° slopes, the uphill knees maintained flexion without returning to the neutral position, and the knee flexion angle on the downhill was smaller than that on the uphill. Furthermore, the ankles in the downhill position maintained inversion without returning to the neutral position, and plantar flexion was predominant. Our proposed supporter combines a newly developed knee-joint supporter and an ankle-joint supporter to provide walking assistance that takes advantage of the gait characteristics of walking on a cross slope.

II. BIOMIMETIC ASSIST SUITS (BAS)

A. Configuration

In this study, we focused on the knees and ankles as the target joints. To provide the requisite support, we developed a biomimetic assist suit (BAS) (Fig. 1). The BAS was designed to support gait on a cross slope to independently vary the supportive torque on the left and right joints. To reduce the weight of the supporter, a passive mechanism was introduced; this mechanism does not require much power and generates support torque by the deformation of the material and springs. It was also made semi-active through a mechanism that varies the mechanical properties of the supporter using a small actuator. The BAS comprises two major components: (i) the dual axial ankle supporter (DAAS), which assists ankle motion, protects the ankle joint, and assists plantar flexion/dorsiflexion and (ii) the semi-active knee joint (SKJ), which assists knee motion.

Figure 1. Biomimetic Assist Suit (BAS)

B. Function

The operation of these devices is linked to the provision of optimal assistance during cross-slope walking. The slope condition is estimated from the ankle joint motion, and the

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support torque of the SKJ is switched to an appropriate value. To cope with the slope condition, a preliminary action is assumed to occur during the swing phase of the ankle. First, inversion and plantar flexion of the ankle joints occur simultaneously downhill on a cross slope. At this instance, the angle of flexion of the knee joint is smaller than that while walking on level ground. Therefore, once ankle joint inversion is confirmed, the knee joint switches to output a large support torque at a small flexion angle, and the knee joint can support the body weight at a small flexion angle. Dorsiflexion and eversion of the ankle occur simultaneously on the upslope and cross slope. At this time, the knee flexion angle does not return to the reference angle, and the knee angle is maintained at a large value; therefore, supportive torque must be exerted at a large flexion angle. The knee support and support-switching mechanisms are explained in Section IV. The ankle also provides appropriate support for the uphill and downhill areas of cross-slope walking. This feature is further described in Section III.

III. DAAS

A. Basic structure

To assist walking not only on level ground but also on a cross slope, we proposed a DAAS, which features two axial movements in the sagittal and frontal planes. Figure 2 shows the fundamental structure of the DAAS, which comprises a control box for the bending sensor, elastic connectors, a main frame, and an adjustable assisting direction mechanism (AADM). The mass of the DAAS is approximately 328 g per foot.

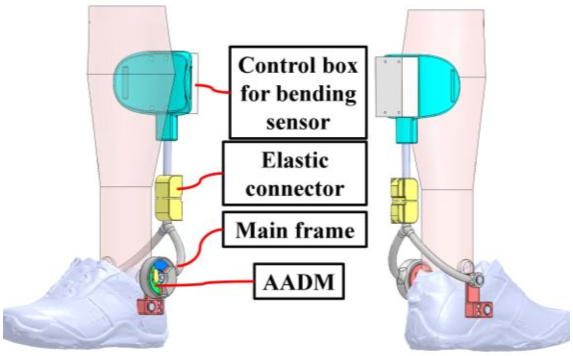

Figure 2. Dual Axial Ankle Supporter (DAAS)

Figure 3 illustrates the basic structure of AADM, which consists of adjustable stoppers and pins. The mass of AADM was approximately 15 g. Its motion is detailed in Section Ⅲ-B.

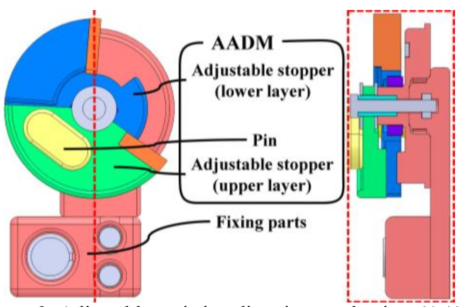

Figure 3. Adjustable assisting direction mechanism (AADM)

B. Mechanism in sagittal plane

The relative motion of the main frame and adjustable stopper caused the motion of the DAAS in the sagittal plane. The main frame moves with the user's leg, whereas an adjustable stopper is fixed to the shoe. As shown in Fig. 4, the supportive torque of the DAAS is generated when the main frame contacts the adjustable stopper. In addition, the two layers of adjustable stoppers determine the angle of movement of the main frame. The upper layer determines the end of the main frame's angle of motion in plantar flexion, whereas the lower layer determines the end of the main frame's angle of motion in dorsiflexion. The pins secure these adjustable stoppers together. The three fixing holes allow the main frame angle of motion to vary depending on the fixing method (Fig. 5). We designed three assistive modes (adjustable positions of the stopper): (1) one-way support for dorsiflexion (dorsiflexion mode), (2) two-way support (two-way mode), and (3) one-way support for plantarflexion (plantarflexion mode).

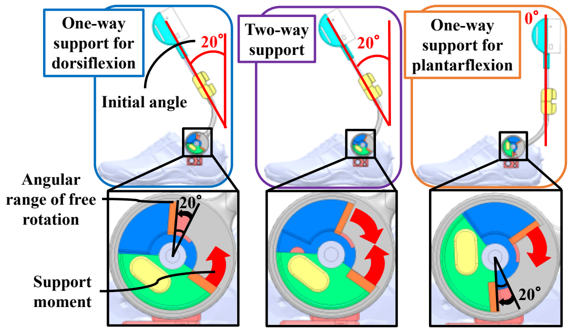

Figure 4. Supporting mechanisms in the sagittal plane

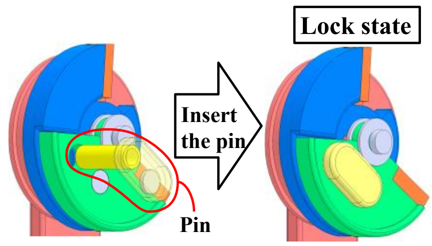

Figure 5. Mechanism of adjustable stopper

In the dorsiflexion mode, the DAAS generates dorsiflexion torque during the swing phase. This function helps the user drop their foot [9] and is used for level-ground and cross-slope walking. In addition, it operates on an uphill ankle during cross-slope walking [14]. The initial support angle was 20° for dorsiflexion. The actual supportive torque of the DAAS in dorsiflexion was measured; when the main frame was rotated 25° in plantar flexion from the initial position, the supportive torque in dorsiflexion was 2.5 Nm.

In the plantar flexion mode, the DAAS generates plantar flexion torque in its stance phase. According to the literature [9], downhill ankles in swing cause inversion or plantarflexion on cross slopes, and the downhill ankles maintain inversion without returning to their neutral positions. Plantarflexion is predominant during the stance phase of downhill cross-slope walking. Therefore, this mode was used for the downhill of cross-slope walking and downhill walking. The baseline angle was defined as an upright ankle posture. The actual supportive torque of the DAAS in plantar flexion was also measured;

when the main frame was rotated 25° in dorsiflexion from the initial position, the supportive torque in plantar plantarflexion was 1.5 Nm.

In the two-way mode, the DAAS generates both plantar flexion and dorsiflexion torques, similar to mountaineering boots. This mode is useful for rough terrain and uphill ankles during cross-slope walking.

The edge surfaces of the adjustable stopper were covered with an elastomer (TPU90, PolyMaker). The hardness could be adjusted by varying the filling density during 3D printing. A full-filling setting was selected for this study.

C. *Mechanism in frontal plane*

The motion of the DAAS in the frontal plane is caused by the elastic deformation of the elastic connector. This part connects the cuff and main frame of the DAAS. Based on the findings presented in Section III, the following design goals were defined to assist in inversion or eversion:

1. The supportive devices should not inhibit 10° inversion or 5° eversion during the swing phase.
2. The supportive devices should generate approximately 5 Nm at 20° inversion and 2.5 Nm at 10° eversion.

To achieve these goals, a nonlinear structural elasticity was applied to an elastic connector (Fig. 6). A three-dimensional printer (Mega X, Anycubic Corp.) with an elastomer filament (TPU95, Polymaker) was used to fabricate the connectors. TPU was selected because of its high allowable strain. The asymmetric shape with slits of different depths generated asymmetric and nonlinear elastic reaction moments for ankle inversion and eversion. The actual supportive torques of the elastic connector were measured as approximately 2 Nm at 15° eversion and approximately 2 Nm at 25° inversion. This experiment confirmed the nonlinear structural elasticity of the connector during the bending action in eversion.

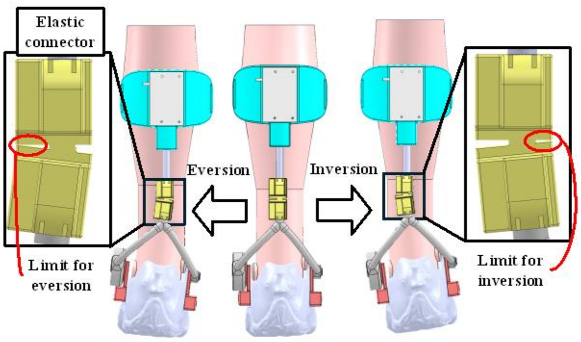

Figure 6. Elastic connector of DAAS

A bending sensor (BS-65, Sensia Technology Corporation) was inserted into the elastic connector, and a circuit box was attached to the cuff (Fig. 7). The bending sensor consists of a polyethylene terephthalate sheet and a resistor. Figure 8 shows the control circuit. The left side of the microcontroller (Arduino Nano, Arduino LLC.) is a bridge circuit that detects the internal bending of a DAAS internal bending sensor, and the resistance of the bridge circuit includes the bending sensor inserted into the DAAS.

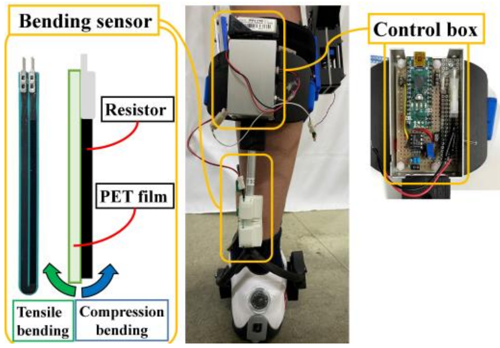

Figure 7. Control system for bending sensor

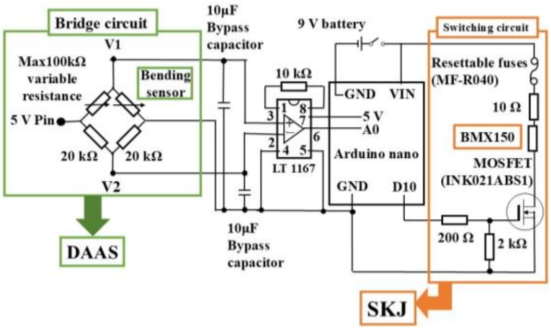

Figure 8. Control circuit for bending sensor and BKJ

The right side of the microcontroller is a switching circuit that operates BMX150 using the adjustable reaction force mechanism (ARFM) of the SKJ. When the sensor is bent by more than 10°, the voltage difference is amplified with the bridge circuit, and the microcontroller turns on the switching circuit when a differential voltage above a set threshold is input.

IV. SEMI-ACTIVE KNEE JOINT (SKJ)

A. *Basic structure of semi-active knee joint*

Figure 9 shows the SKJ developed in this study, and Figure 10 shows a schematic of its spring mechanism. The springs are arranged in parallel, and in the configuration used in this study, a weak spring is inserted into the hollow center of the strong spring. The left side of the figure shows how the SKJ type-P is attached to the upper and lower legs with the respective cuffs. The weight of the joint was approximately 300 g, and those of the cuff for the uphill and downhill legs were 155 and 128 g, respectively. SKJ type-P can match deep flexion movements > 90°.

The SKJ mimics the motion of knee joints using femoral and tibial gears. It can also switch the force required for assistance using ARFM. When the spring system in the ARFM is in the off-state of the stopper, the weak spring is pressed in the designed initial range, and the strong and weak springs are pressed simultaneously after the initial deformation (Fig. 10). However, in the on-state of the stopper, the strong and weak springs are pressed simultaneously from the beginning of the contact.

Figure 9. Semi-active knee joint (SKJ)

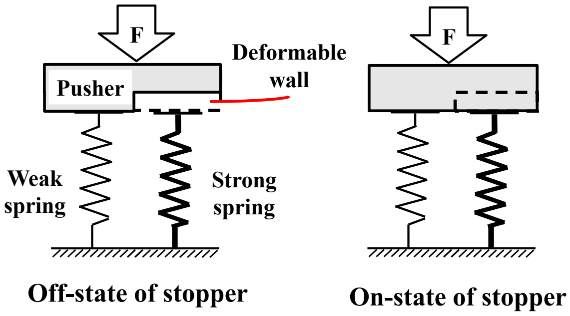

Figure 10. Spring mechanism of SKJ

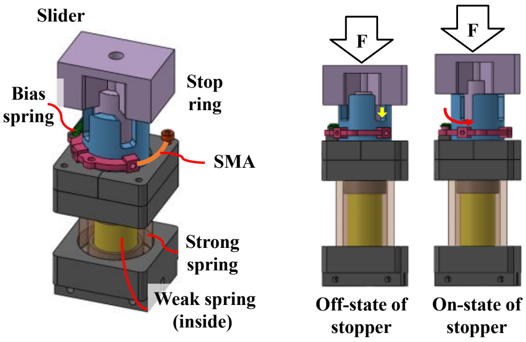

Figure 11. Adjustable reaction force mechanism (ARFM)

B. *Mechanism of ARFM in SKJ*

Figure 11 shows the basic structure of ARFM. This mechanism is composed of a strong spring with a weak spring inside and a rotating mechanism consisting of a shape memory alloy (SMA), stop ring, and slider on top of the strong spring. When the stopper is in the off-state, the slider enters the hollow stop ring and pushes only the weak spring. However, when the stopper is in the on-state, the stop ring rotates, the pusher contacts the bosses of the stop ring, and the strong and weak springs are simultaneously pushed via the stop ring. BMX150® of Toki Corp. was used for the SMA. As described in Section III, the SMA is deformed by the current input via the control box. A kinematic model of the SKJ was used to estimate supportive torque. Details are provided in [16]. In this design, the strong spring constant is 15.8 N/mm and the weak spring constant is 6.3 N/mm; in the type-P mechanism, the springs are arranged in parallel, resulting in a composite spring constant of 22.1 N/mm.

V. EVALUATION TESTS FOR SKJ IN GAIT EXPERIMENTS

A. *Method*

Figure 12 shows the experimental environment of the gait experiment. Fourteen healthy males (19–22 years old, 1.62–1.80 m in height) were recruited for the gait measurements during cross-slope walking. They walked at least three steps on level ground and a 20° cross slope at comfortable paces and strides with/without the BAS. One foot was higher on the cross slope (uphill), and one foot was lower on the cross slope (downhill). The participants wore the same shoes with and without the BAS. In the condition with the BAS, the participants wore the BAS in the off-state and on-state for their uphill and downhill movements, respectively. The radii of the femur and tibia gears of the SKJ were 30 mm for all the participants. In the DAAS, a one-way support for dorsiflexion was used on both ankles for level-ground walking. During cross-slope walking, one-way support for dorsiflexion and one-way support for plantarflexion were used on the uphill and downhill ankles, respectively.

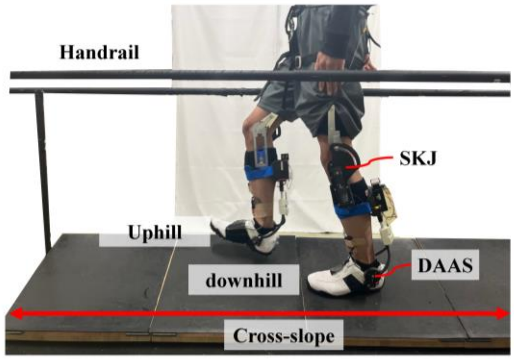

Figure 12. Experimental environment

Figure 13. Position of EMG sensors and Goniometers

In this experiment, a surface electromyogram (EMG) sensor (Trigno Avanti Sensor, DELSYS) was used to measure muscle activity, and a goniometer (SG-150, Biometrics Ltd.) was used to measure the joint angles (Fig. 13). The EMG sensor has a variable bandpass filter. The filter bandwidth was set to 20–450 Hz. The four EMG sensors were attached to the anterior tibialis (AT), inner gastrocnemius (iGS), outer gastrocnemius (oGS), and rectus femoris (RF) muscles of the right foot. Electrical goniometers were attached to both ankles. In this experiment, we obtained EMG data at 3 kHz. The EMG data were normalized using the maximum voluntary isometric contraction (MVC) method. The right foot was the dominant foot among participants. Therefore, we focused on the gait cycle of the right foot starting from the initial contact. We used the root mean square (RMS) for 100 ms. This experiment was approved by the Ethical Board of the Faculty of Science and Technology of Oita University.

B. Results

We compared the effects of support on several items under different ground conditions (level, downhill, and uphill). Figures 14–20 show the comparison results of the gait cycle, the difference between the maximum and minimum angles during the swing phase for ankle inversion/eversion (in the frontal plane) and plantarflexion/dorsiflexion (in the sagittal plane) and muscle activations, respectively. Two-way ANOVA were conducted for the supportive condition (with/without BAS) and ground condition (level ground, downhill, and uphill on a cross slope). A significant difference was observed in the gait cycle for the supportive conditions ($P < 0.05$), with no interaction effect with the ground condition (Fig. 14). Significant differences were also found in the range of motion for inversion/eversion and plantar flexion/dorsiflexion for supportive conditions ($P < 0.05$), with no interaction with the ground conditions (Figs. 15 and 16). Due to large variance, there is no significant difference between support /no support conditions for all EMG results. Muscle activations on the downhill were significantly higher than the other conditions.

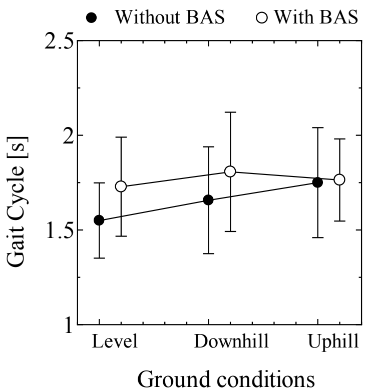

Figure 14. Gait cycle with and without the BAS

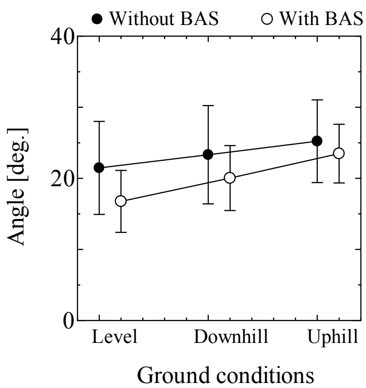

Figure 15. Ankle joint in frontal plane with and without the BAS

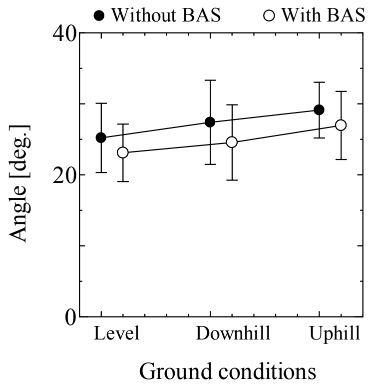

Figure 16. Ankle joint in sagittal plane with and without the BAS

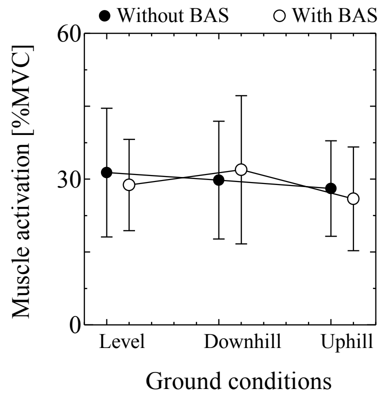

Figure 17. Muscle activation of AT with and without the BAS

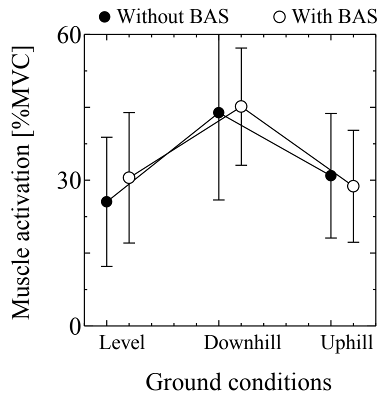

Figure 18. Muscle activation of oGs with and without the BAS

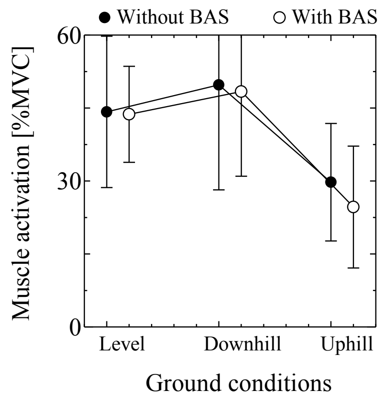

Figure 19. Muscle activation of iGs with and without the BAS

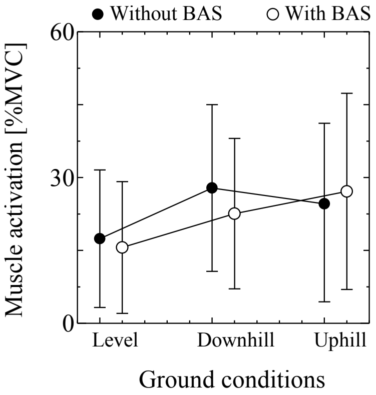

Figure 20. Muscle activation of RF with and without the BAS

C. Discussion

The comparison of the angles in the swing phases revealed significant differences with and without the supporter, and the average angular difference with the supporter tended to be smaller than that without it. These results imply that the BAS worn on the ankle joints successfully supported the users' ankles with the elastic connector and AADM functions of the DAAS. A comparison of one gait cycle revealed a longer gait cycle with the supporter. In addition, there are no significant differences in muscle activations. These facts could be attributed to the use of unfamiliar supporters. Users must individually learn the best way to use the device. A limitation of this study is that the switching of the support mode of the AADM was performed manually, and an optimal design for the automatic switching mechanism should be considered in future studies.

VI. Conclusion

This paper describes the development of a biocompatible BAS for assisting walking on a cross slope, the design and development of an SKJ that assists knee joint motion, and a DAAS mechanism that assists ankle joint motion. These comprise the BAS, designed as a supporter for cross-slope walking, and the DAAS and SKJ are linked to provide different assistance uphill and downhill of the cross-slope. The ARFM is activated when the foot is estimated to be downhill of the cross slope from the flexion sensor attached to the elastic connector and assists in weight bearing even if the knee joint flexion angle is small. The flexion sensor was not activated on the uphill side of the cross slope, and a supportive torque was exerted at a large flexion angle. To evaluate the effectiveness of the BAS, the gait cycle and ankle joint angles were recorded while walking on a cross slope and level ground. The results showed that DAAS reduced the angular change in the anterior and sagittal planes of the ankle joint during walking, protecting the ankle joint and limiting its range of motion.

Acknowledgment

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