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Design of a device with non-interacting cam mechanism that independently assists dorsiflexion and plantarflexion based on hip motion

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Abstract— **Stroke patients with muscle weakness and gait disturbances require assistance for inadequate ankle dorsiflexion and plantarflexion movements. This research developed a non-powered walking aid device that assists ankle movements using hip motion as the power source. Unlike systems that rely on active actuators, such as motors, this device employs a cam mechanism and cables to convert the energy from hip flexion and extension movements into dorsiflexion and plantarflexion torques during walking. A study conducted on able-bodied subjects demonstrated that, when assisted by the device, there were significant increases in the ankle joint angle at initial landing, plantarflexion torque in the late stance phase, and dorsiflexion angle in the swing phase. However, a decrease in ankle plantarflexion angular velocity was observed, suggesting that further improvement of the device is needed.**

I. INTRODUCTION

The global prevalence of stroke is alarmingly high, with approximately 80.1 million individuals affected worldwide, and the issue of residual physical disabilities that disrupt life post-stroke is increasingly evident [1]. Hemiplegia, a common sequel of stroke, significantly impairs daily activities and heightens the risk of falls, potentially leading to reduced motivation for exercise, muscle weakness, and joint contractures [2]-[5].

In the context of lower limb movements in the sagittal plane, key actions include dorsiflexion, where the toes are lifted, and plantarflexion, where the toes point downward. A reduction in ankle dorsiflexion can result in stumbling, while diminished plantarflexion may lead to decreased propulsive force. It is crucial, therefore, to utilize orthotics to enhance life quality by supporting these movements during gait rehabilitation [6]-[7].

The following is a list of robotic orthotics capable of assisting walking.

- Robo-ankle: A wearable device that uses soft robotics to assist with ankle motion, targeting rehabilitation for patients with lower limb impairments. [8]
- WalkAid System: This system provides functional electrical stimulation (FES) to assist with dorsiflexion, especially useful for individuals with multiple sclerosis or stroke survivors. [9]
- LOPES Exoskeleton Robot: A lightweight exoskeleton designed to support walking for people with lower limb disabilities, improving their gait and mobility. [10]
- HAL (Hybrid Assistive Limb): A robotic exoskeleton that uses sensors to detect the wearer's movement intentions, offering assistance to those with weakened lower limbs, especially useful for patients with spinal cord injuries or stroke. [11]
- Wearable Power-Assist Locomotor (WPAL): Originally designed as an assistive orthotic for paraplegic individuals, it has evolved into a wearable robot, the WPAL, to provide assistance during walking and rehabilitation phases. [12]

While robotic orthotics offer substantial support, their reliance on heavy, costly components such as motors and batteries renders them impractical for widespread and home use, particularly during the maintenance phase of rehabilitation [13]. This highlights the need for developing a non-powered, compact, and lightweight walking aid.

This research aimed to design and evaluated a small, non-powered walking aid that supports both dorsiflexion and plantarflexion simultaneously in daily activities. Unlike conventional devices, the one developed here harnesses hip motion during walking to provide targeted assistance. Specifically, the device aims to aid plantarflexion just before and after the foot pushes off, and dorsiflexion just before and after heel contact. We developed this novel device and assessed its effectiveness on healthy subjects, marking the first instance of a non-powered device capable of supporting both dorsiflexion and plantarflexion during gait.

II. DEVELOPMENT OF A NON-POWERED WALKING AID DEVICE

A. Requirement

The requirements for the non-powered walking aid device are outlined as follows:

- Non-powered: The device operates without external power sources.
- Dorsiflexion Support: It should provide dorsiflexion support to the ankle joint during 60% to 80% of the gait cycle.
- Plantarflexion Support: It should offer plantarflexion support to the ankle joint from 0% to 12% of the gait cycle.

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- Hip Joint Utilization: Given that the range of motion of the hip joint in patients at Brunnstrom stages 5 or 6 is similar to that of a healthy individual—ranging from approximately 20° of flexion to 20° of extension—the hip joint serves as the power source for the device [14]-[15].

B. Device Overview

Figures 1 and 2 present the general layout of the non-powered walking support system. The system features a frame that attaches to the body via a belt, with a disk that moves to the upper limit following the frame's path. Dorsiflexion and plantarflexion are independently facilitated by hip motion and are connected to the toes and heels, respectively, through pulleys.

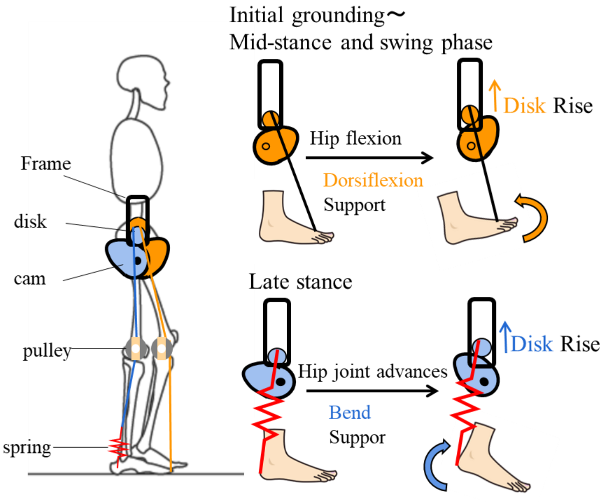

Figure 1. Overview of the non-powered device

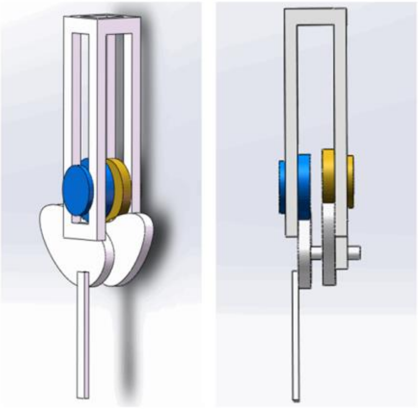

Figure 2. Non-interfering cam mechanism capable of independently assisting dorsiflexion and plantarflexion

Figure 3 presents the front and side views of the actual device. The frame dimensions are 6 cm in length, 6 cm in width, and 12.8 cm in height. Fabricated using a 3D printer, the device is constructed from Full Cure 720 material and weighs a total of 512 g.

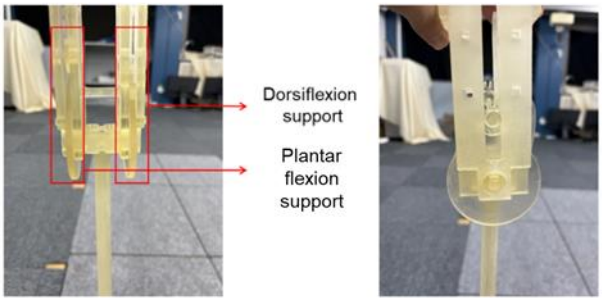

Figure 3. Device created with a 3D printer

A diagram of the frame model designed using SolidWorks is illustrated in Fig. 4. The dimensions of the frame are 6 cm x 6 cm x 12.8 cm (L x W x H). Inside the frame, two disks and a cam are installed. During walking, the center of each disk and the center of rotation of the cam align vertically within the same plane.

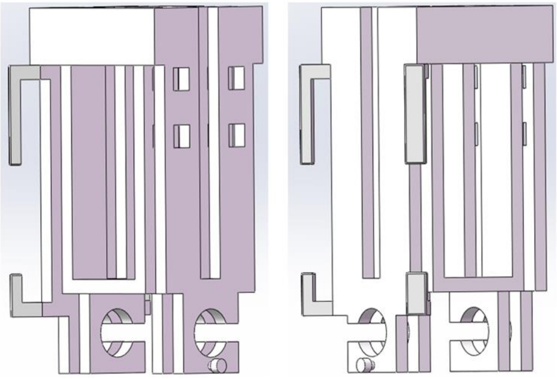

Figure 4. Frame diagram designed by Solidworks

The appearance of the developed device when mounted on a human is depicted in Fig. 5. The length of the wire can be adjusted to suit individual height differences by engaging a spring buckle. The wire, made of stainless steel, measures 130 cm in length and has a diameter of 1.5 mm. The spring constant for the heel spring was set at 1.2 N/mm.

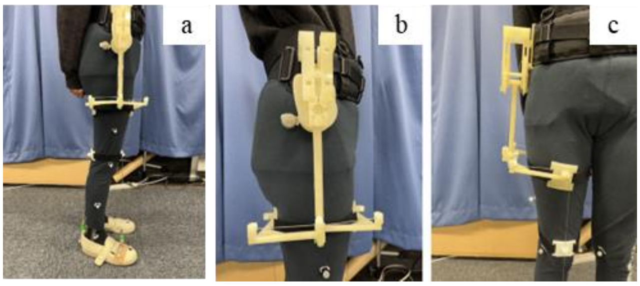

Figure 5. (a) Appearance of non-powered walking support device installed on a human, (b) fixation to the hip and thigh, (c) fixation to the thigh and knee

C. Design of Dorsiflexion Support Unit

Figure 6 illustrates the cam and disk mechanism for the dorsiflexion system. The disk is vertically displaced by the cam's rotation. A 20° rotation of the hip joint results in a 10 mm upward displacement of the disk. Figure 7 displays the cam movement during dorsiflexion.

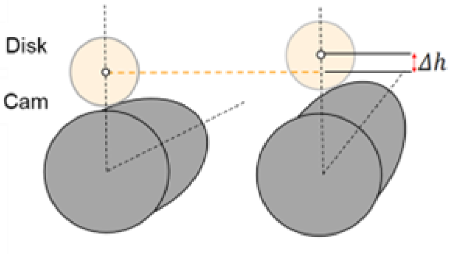

Figure 6. Design of the dorsiflexion system for a non-powered walking assist device

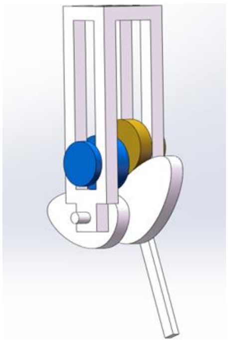

Figure 7. Movement of the cam when dorsiflexion is assisted.

D. Design of Plantarflexion Support Unit

The plantarflexion support mechanism, depicted in Figure 8, employs the same disk and cam configuration as the dorsiflexion unit. A wire runs from the disk at the hip joint through a pulley at the knee to a spring at the heel. During the transition from mid-stance to late stance, as the hip joint extends backward, the disk rises, the spring stretches, and energy accumulates to aid plantarflexion. As the heel rises, the stored energy is released to facilitate plantarflexion. Figure 9 illustrates the cam movement during plantarflexion.

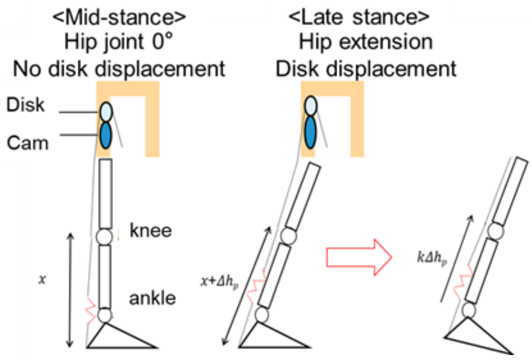

Figure 8. Plantarflexion support mechanism

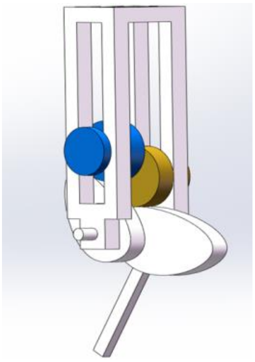

Figure 9. Movement of cam and disk during p;antarflexion support

E. Mechanism of force transmission

As depicted in Figures 10 and 11, wires connect the disk to the toes and heels through pulleys affixed to the thigh and knee, supporting both dorsiflexion and plantarflexion. Additionally, a folded structure in the disk doubles the displacement, effectively transmitting it to the ankle joint (Figure 11).

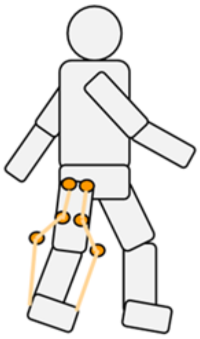

Figure 10. Transmission of force from the hip joint by wire

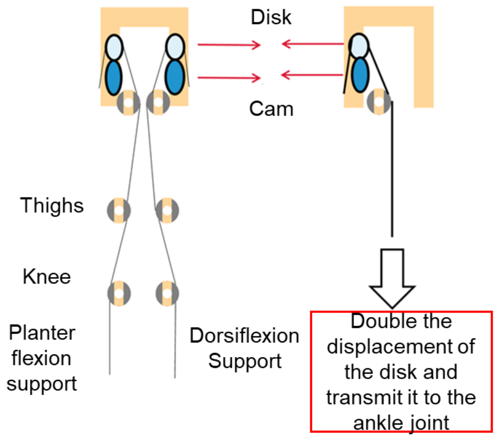

Figure 11. Disk displacement and ankle joint support

III. FEASIBILITY STUDY

A. Study Design

Eight healthy adult subjects, aged 24.8 ± 0.83 years, walked a 5-meter distance six times each, both without and with the device. We investigated whether the non-powered walking aid improved walking support during various phases of the gait cycle, including initial ground contact, mid-stance, late stance, and swing.

B. *Evaluation Method*

The primary evaluation metrics included the heel contact angle, plantarflexion torque during the late stance phase, and ankle joint angle during the swing phase. Body movements and joint angles were measured using a motion capture system (MAC3D, nac Inc., Japan), which calculates the positions of reflective markers with eight dedicated cameras. Nineteen markers were placed on the lower half of the body as depicted in Figure 12. Ankle joint angles were derived from the coordinate data collected. For assessing plantarflexion torque, eight force plates (AMTI Gen 5: Advanced six-channel signal conditioner, USA) were utilized to measure the torque, and the level of torque support was calculated from the data obtained. Statistical analysis was conducted using the Wilcoxon signed-rank test.

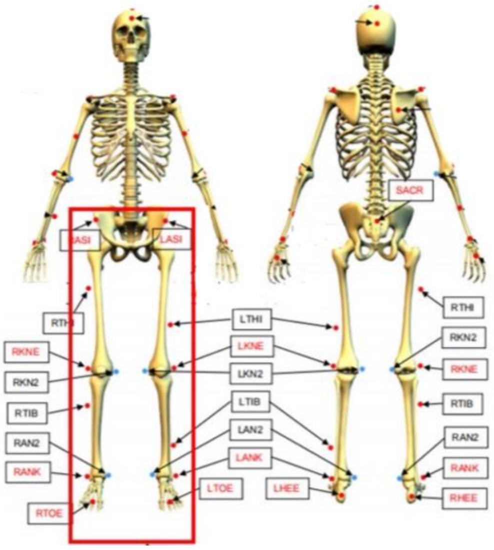

Figure 12. Marker placement

Figure 13. Force plate used to measure plantarflexion torque

IV. RESULT AND DISCUSSION

A. *Study Results*

Figure 14 presents the heel contact angle. The dorsiflexion joint angle at initial contact with the orthosis significantly increased compared to that without the orthosis, indicating effective support for dorsiflexion by the device. This enhancement allowed the toes to be raised at heel contact.

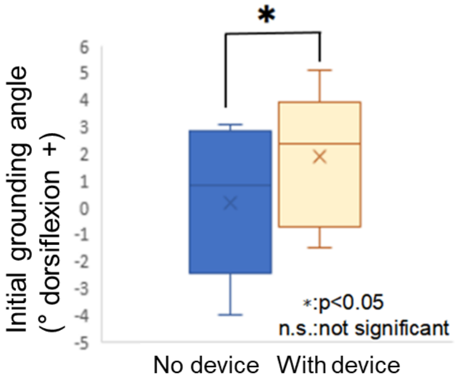

Figure 14. Result of heel contact angle

Figure 15 depicts the plantarflexion torque during the late stance phase. There was no significant difference in torque between conditions with and without the orthosis, suggesting that the spring force was insufficient to support plantarflexion in the late stance phase. This indicates that the current spring constant may need adjustment to effectively support plantarflexion.

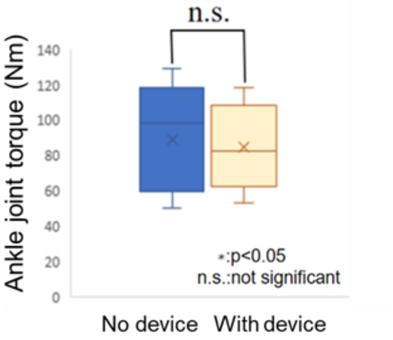

Figure 15. Result of plantarflexion torque

A comparison of dorsiflexion angles during the swing phase is shown in Figure 16. The angle increased significantly with the device compared to without it, demonstrating the device's efficacy in supporting dorsiflexion during the swing phase.

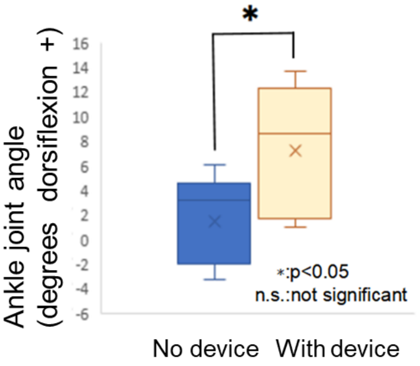

Figure 16. Result of dorsiflexion angle during the swing phase

B. Discussion

The results indicate that the use of the developed non-powered walking aid significantly increased the ankle joint angle at initial ground contact, facilitating better heel ground contact. Moreover, the significant increase in dorsiflexion angle during the swing phase suggests that the device enabled subjects to walk with their toes elevated, potentially reducing the risk of falls. This underscores the device's utility in enhancing gait safety and efficiency, though improvements in plantarflexion support are necessary.

In contrast, as depicted in Figure 17, no significant difference was observed in plantarflexion torque during the late stance phase when using the device, attributed in part to insufficient spring force for supporting plantarflexion. Consequently, the experiment was repeated with a modified spring constant, increasing from the original K1 = 1.2 N/mm to K2 = 6.5 N/mm, involving two young healthy subjects. The results, illustrated in Figure 17, indicate that a higher spring constant facilitated an increase in plantarflexion torque during the late stance phase, demonstrating the device's capacity to enhance plantarflexion torque through adjustment of the spring constant. Therefore, identifying an appropriate spring that provides adequate plantarflexion torque for each user remains a critical challenge.

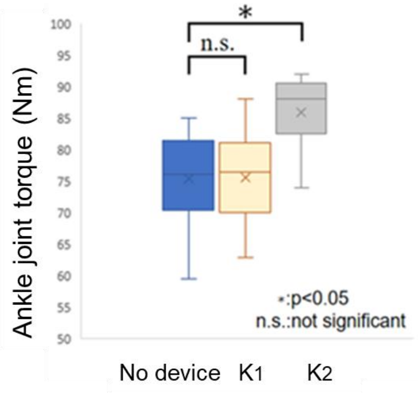

Figure 17. Comparison of plantarflexion torque at different spring constants

The device's impact on gait was also examined using 12 young healthy subjects. Figure 18 shows the angular velocity of toe drop during the loading response phase, while Figures 19 and 20 display the hip flexion and extension angles, respectively. The findings reveal significantly lower angular velocities of toe drop, hip flexion angles, and hip extension angles when using the device. The reduced angular velocity of toe drop could result from the device pulling on the toes, potentially interfering with normal walking. Additionally, the decreased hip joint angles suggest a loss of kinetic energy from the device, which harnesses the hip joint's kinetic energy to support plantarflexion and dorsiflexion. These observations underscore the need to enhance the device to preserve the kinetic energy of the hip joint during walking and to devise a mechanism that minimizes walking impediments.

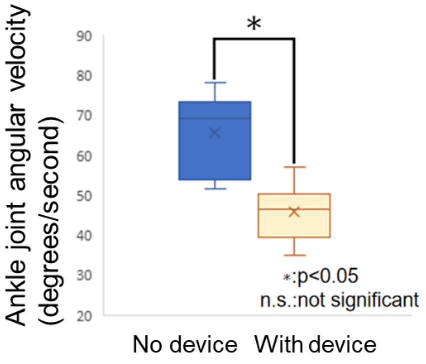

Figure 18. Comparison of toe drop angular velocities in loading response phase

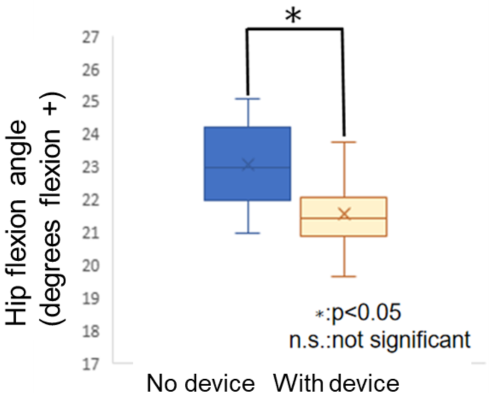

Figure 19. Result of hip flexion angles

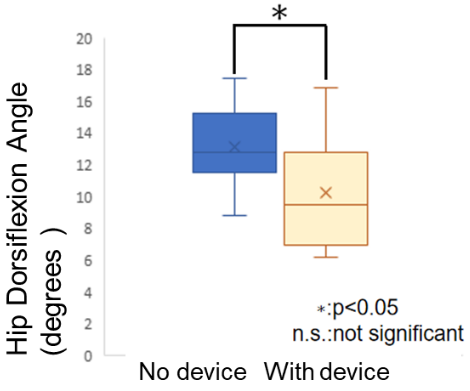

Figure 20. Result of hip extension angles

V. CONCLUISON

In this research, we developed a non-powered device that assists both dorsiflexion and plantarflexion by harnessing hip motion. The study involving this device demonstrated an increase in the heel contact angle, indicating its potential to prevent stumbling. Furthermore, an increase in the plantarflexion angle during the late stance phase suggests that effective plantarflexion support is achievable. The device also increased the dorsiflexion angle during the swing phase, further indicating its capability to prevent stumbling and promote secure heel ground contact. However, as the study was conducted with eight healthy subjects, the device's applicability to stroke patients remains to be determined. The current device weighs 512 grams, highlighting the need for further weight reduction to enhance usability. Since this study focused only on the sagittal plane and did not consider the effects on the horizontal or forehead planes, it is necessary to consider three-dimensional motion in future development. Additionally, it is imperative to design a structure that minimizes the loss of kinetic energy at the hip joint and to improve the device's wearability to reduce the duration it needs to be worn.

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