

Development of a String-Driven Walking Assist Device Powered by Upper Body Muscles

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Abstract—The elderly sometimes fall owing to tripping caused by decreased dorsal flexion during the swing phase. A wearable assist device that was able to increase dorsal flexion to reduce the risk of tripping was proposed in this study. This device assists in dorsal flexion during the swing phase by transferring the tension force from the upper body to the foot by connecting them using strings. To improve the efficiency of the assist, the string path was optimized to maximize the assist torque at the time of minimum foot clearance (MFC). In this study, two string-path patterns were tested. Based on the walking experiment results, we confirmed that the wearable assist device that we developed successfully increased both the dorsal flexion and minimum foot clearance during the swing phase.

Keywords— *gait motion, assistive device, minimum foot clearance*

I. INTRODUCTION

The walking ability of the elderly gradually declines owing to the decrease in muscle strength associated with aging. Recently, wearable gait assist devices (hereinafter referred to as wearable assist devices) have been proposed for supporting the walking motion of the elderly. Such wearable assist devices was able to be used to improve the working efficiency of aged labors. In addition, increasing demand of rehabilitation associated with increasing population of the elderly required such assist devices for several decades. Active leg exoskeleton (ALEX) [1], which is a rehabilitation wearable assist device, assists in walking rehabilitation for stroke patients. It modifies the gait motion of the wearer by applying torque to the hip and knee joints. Hybrid assistive limb (HAL) [2] utilizes electromyography, which can be used to estimate the intention of the muscle contraction to apply torque to joints at the appropriate time. The rhythm assist device [3] can be synchronized with the gait timing of the wearer using a central pattern generator. This light device is designed to fit the wearer.

The decrease in gait ability is determined based on the combination of a reduction in balance sensation, muscle strength, and other functions. These typically result in a reduction in stride length, range of joint motion, and foot clearance [4-6]. The reduction in foot clearance and increase in the variation of the foot clearance increases the risk of tripping. Such changes in gait motion occur in the elderly owing to the decrease in the dorsal flexion angle [7]. Thus, it is important to increase the dorsal

Table. 1 Type of walking assist device

Type	Actuator Type [9]	Supporter Type
Assist Power	○High Power	×
Assist Pattern	○Variable Assist Pattern	×
Weight	×	○
Simplicity of Attaching	×	○
Operating Time	×	○

flexion angle and restore the foot clearance in order to avoid tripping [8].

Several wearable assist devices include an actuator that provides torque to each joint to assist in the wearer's joint motion [9]. Thus, although it is possible to adjust the assist force and pattern to the wearer, the usability of such devices in daily living environments is limited because of its weight, and limited operating time owing to the battery, the attach and detach processes, and other mechanical limitations. The other type of wearable assist device—the supporter type—can also support dorsal flexion by continuously supplying elastic force. Thus, such devices are usually lighter than the actuated ones and can be easily attached/detached. However, the assist pattern should be constant, and thus, it potentially restricts the motion of the ankle joint. The characteristics of such wearable devices are presented for comparison in [Table 1].

Owing to achieve profits of actuator type and supporter type devices at the same time, we developed a muscle-force-transmission type dorsal flexion assist device using string, which utilizes a strong wearer's muscle as the power source instead of actuators [10]. The string-type wearable assist device applies torque to lower limb joints through the string fixed along the leg portion [11, 12]. The tension was generated by pulling the string using the upper body muscles. By optimizing the string path, the efficiency of the torque transfer was able to be improved, which meant that the torque applied to the ankle joint was maximized in the swing phase although the collateral torque applied to the other joints was minimized.

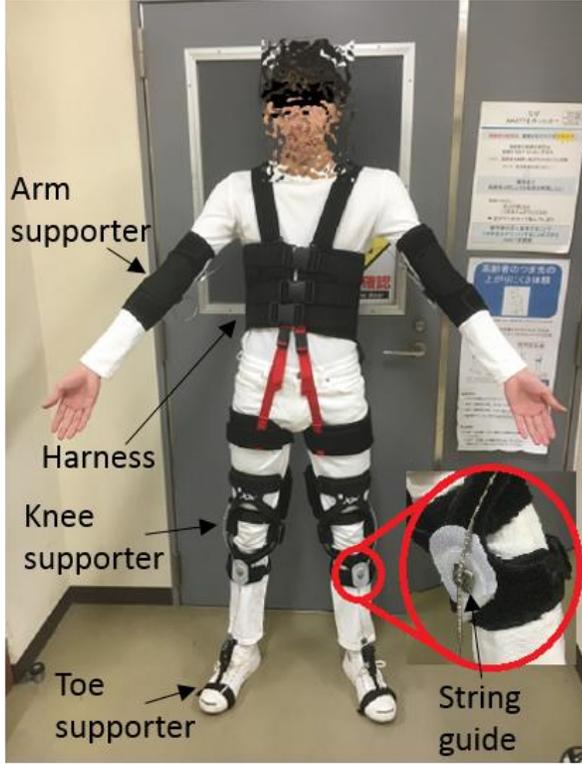


Fig. 1 Design of wearable assist device

II. DESIGN OF WEARABLE ASSIST DEVICE

A. Structure of the wearable assist device

The wearable assist device developed in our study consisted of cuffs and belts that were fixed to the subject's body and limbs. Strings were used to connect the toe to the arm or body through string guides fixed on the cuffs and belts. The string transferred tension force from the upper body, which generates tension force synchronously with the gait cycle, to the toe for assisting the dorsal flexion during the swing phase. In this study, motions of the upper body, such as the shoulder twist and arm swing, were used for assistance in walking instead of an actuator. The wearable device that was capable of realizing these functions was manufactured as shown in Fig. 1. It consisted of four components: toe supporter, knee supporter, harness, and arm supporter. String guides, which is highlighted in Fig. 1, were fixed at the appropriate position of the appliances using Velcro. Because a part of the string consisted of rubber, the tension in the string was able to be measured from the expansion of the rubber portion.

B. Calculation of torque applied by the device

First, the normal gait motion of a healthy young adult on the treadmill was recorded. Following vectors was calculated as a standard coordinate, the zero point was located on the experimental floor. The torque applied to joints during walking was then calculated. The dorsal torque T_1 acting around the ankle joint was obtained from the cross product of the pulling

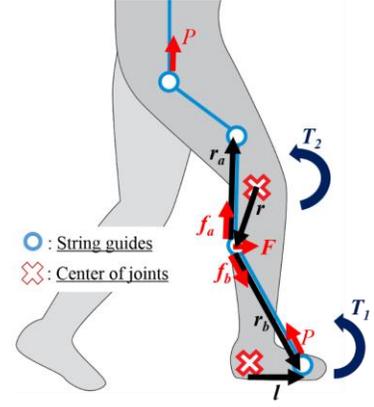


Fig. 2 Theory of generating resultant force and torque

force of the string P and the distance vector from the ankle joints to the string guide l (shown in Fig. 2).

$$T_1 = l \times P \quad (1)$$

The extension torque T_2 around the knee joints was then calculated. The tension vector acted along the folded strings f_a and f_b , and the resultant force F , vectors r_a and r_b from the string guide to adjoining guides, and the distance r from the knee joints to the string guides were defined. These definitions are shown in Fig. 2. Using these values, the extension torque T_2 is obtained from the following equation.

$$f_a = P \times \frac{r_a}{|r_a|} \quad (2)$$

$$f_b = P \times \frac{r_b}{|r_b|} \quad (3)$$

$$F = f_a + f_b \quad (4)$$

$$T_2 = r \times F \quad (5)$$

C. Design of string path

In order to avoid tripping due to decreased dorsal flexion during the swing phase, the assist of the dorsal flexion is effective in maintaining foot clearance. Usually, the foot clearance is the minimum when the swing leg passes by the stance leg. This minimum foot clearance (MFC) appeared in the range of 80% to 85% of the gait cycle (Fig. 3), as reported previously [7]. MFC and dorsal flexion are illustrated in Fig. 4. Thus, the dorsal flexion during this phase should be assisted to increase the foot clearance. In addition, the collateral force generated by the string tension should not disturb the natural gait motion. Because the string guides were placed along the links of the entire leg, the string tension possibly also applies torque to the other joints such as the knee.

Thus, the string path on the lower limb was optimized by maximizing the dorsal flexion torque at the ankle joint at the time of MFC and minimizing the maximum torque around the knee joint. The positions of the string guide were defined as the variable parameters. The evaluation function J was defined as shown in eq. (6). By minimizing J , the dorsal flexion torque at the ankle joint T_{ankle} was maximized and the torque at the knee extension torque T_{knee} was minimized. In this simulation, the

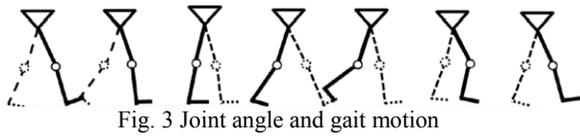
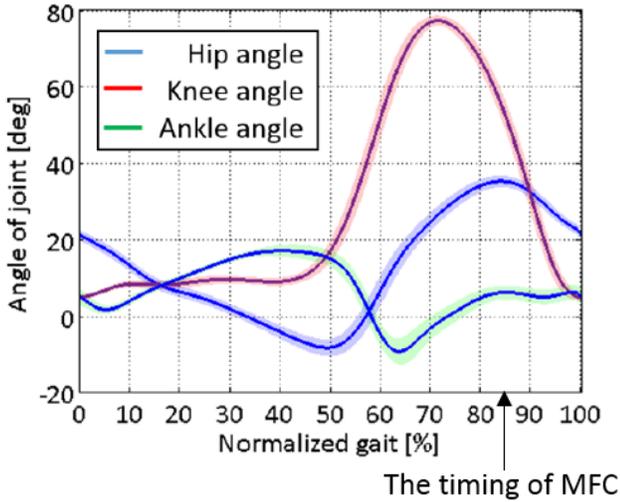


Fig. 3 Joint angle and gait motion

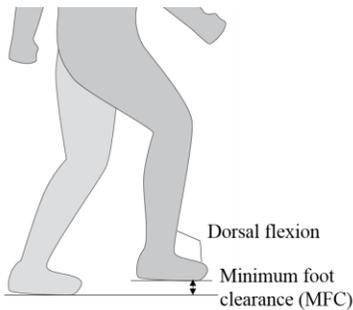


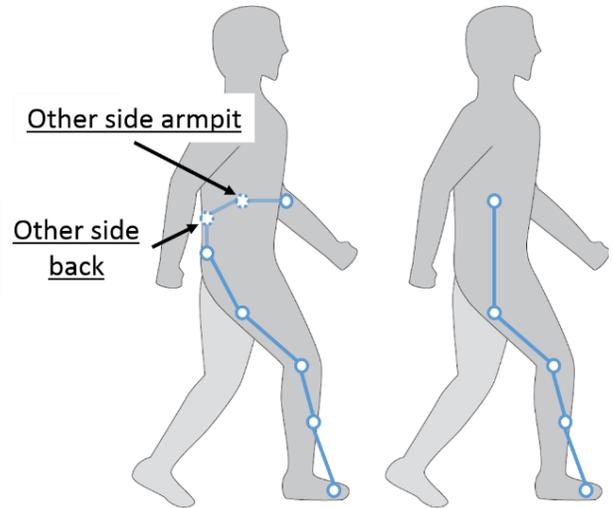
Fig. 4 MFC and dorsal flexion

tension force was maintained at a constant value throughout the gait cycle.

$$J = -T_{ankle} + T_{knee} \quad (6)$$

The string path from the upper limb, which should be designed to pull the string at the appropriate time, was determined by simulating the changing pattern of the string-path length during the gait cycle because the tension probably increased in proportion to the path length. In the preliminary experiment, markers were attached at the position where the string guide was possibly fixed in order to record the relative motion between markers during the gait cycle. By connecting the positions of such markers, the string path, along which the string was expected to stretch at the time of assist, was tentatively determined. The sum of the distances between the markers selected as paths was defined as the string-path length. The tension generated in the string was calculated from the elongation, which was defined as the extension of the string path length, and the elastic modulus of the string. The elastic coefficient of the rubber was 2.8 N/mm.

Using these assumptions, the tension occurring in the string was simulated, and the torque acting on the joints via the string path was calculated using equations (1) and (5). Through this



(a) Arm swing path (b) Body twist path

Fig. 5 String path candidates

process, the string path, along which the torque was generated at the appropriate time, was identified heuristically.

As a result of the optimization process mentioned above, two string paths were determined. The first string path was the arm swing path (Fig. 5 (a)), which generated the force from the swing of the arm. This string path was designed to increase the elongation of the string greatly and to have the ability to adjust the timing when the dorsal torque is generated by controlling the arm swing. The second string path was the body twist path (Fig. 5 (b)), which made use of the twist of the torso. This string path was designed to generate the dorsal torque stably using the twist of the body, which generates the strongest force of the upper body. Figs. 6 and 7 show the estimated torques generated by the respective string paths. It seems that the torque was generated around the ankle joints in the range of 60% to 80% of the gait cycles performing dorsal flexion. Although torque was generated around the knee joint, the magnitude of the torque was minimized.

III. WALKING EXPERIMENT

A. Apparatus

The motion of the subject was measured using a motion capture system (MAC 3D System, Motion Analysis). The position of the markers, which were fixed at the center of each joint of the subject, was projected in the sagittal plane. Each joint angle was then calculated. The angles of the hip, knee, and ankle joints were determined to be 0 deg in the upright posture. The direction of the joint flexion or dorsal flexion was determined to be in the positive direction. The elastic coefficient of the string was 2.8 N/mm. The tension applied to the string was calculated from the extension of the string in a manner similar to the design process of the string path. A treadmill was used to continue the walking for several minutes.

B. Protocol

The experiment was performed with the permission of the institutional review board of the Nagoya University. Two

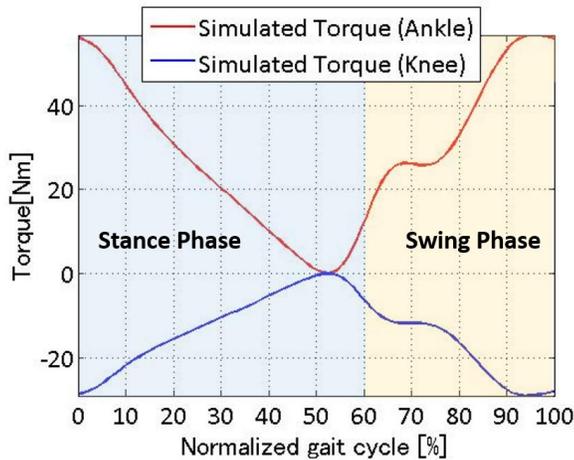


Fig. 6 Simulated joint torque (arm swing path)

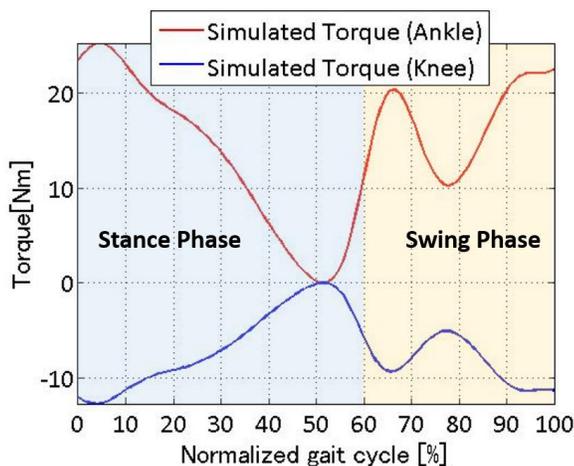


Fig. 7 Simulated joint torque (body twist path)

healthy young male students participated in the experiment. The mean ages of them was 22 years old. To evaluate the influence of the wearable assist device on the gait motion, the gait motions of the assisted trial, which consisted of arm swing path and body twist path, were compared to the normal gait. At first, normal walking was recorded to avoid the learning effect caused by assist devices. Then, assisted trials were recorded sequentially. The order of the assisted trials was randomized. To eliminate the possibility that subjects intentionally changed their gait, the electromyograms of the tibialis anterior, gastrocnemius, quadriceps femoris, and hamstrings were obtained. The walking speed was determined for each subject and fixed through the experiment. The time of the single trial was 3 min, of which the first 1 min was used for gait stabilization and the subsequent 2 min were analyzed. The measured data was normalized by the gait cycle—which was determined as the period between successive heel contacts—and statistically processed.

C. Experimental result

Because a similar trend was observed between the two subjects, the result of one subject is mentioned in the following analysis. In addition, it should be mentioned that there was no difference in the electromyogram of the tibialis anterior during

dorsal flexion, which confirmed that the change in dorsal flexion was caused by the assist device. Furthermore, the stroke of the string measured in the experiment was much smaller than that in the simulation because the tension of the string not only affected the ankle but also the upper body motion. Thus, the effective torque acting on the lower limb joints was smaller in the experiment.

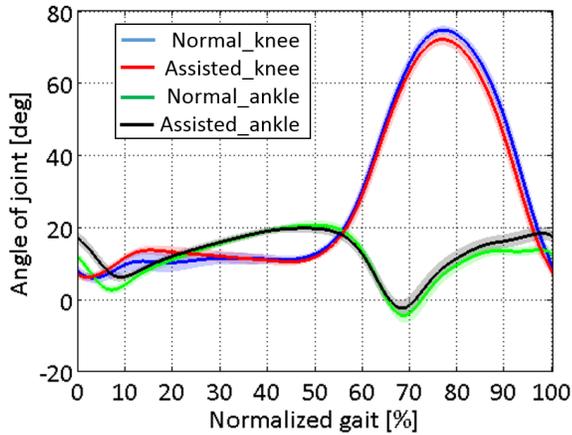
Fig. 8 (a) shows the joint angle pattern of the assisted walking with the arm swing path and normal walking. The torque acting on the knee and ankle joints was also drawn in Fig. 8(b). As compared to normal walking, the dorsal flexion angle of the assisted walking became approximately 2 deg greater in the range of 70% to 80% of the gait cycle, which was the time of the assistance.

Fig. 8 (b) is compared with the simulation result shown in Fig. 6. The torque pattern differed greatly in the range of 25% to 65% of the gait cycle; Fig. 8 (b) shows that the torque became zero in the experiment during this phase, which means that the string was sagging. Because the arm swing path generated force from the swing of the arm, the wearer was able to adjust the timing for generating the dorsal torque. Thus, the pattern of the string extension, which generated tension, was probably controlled by the subject.

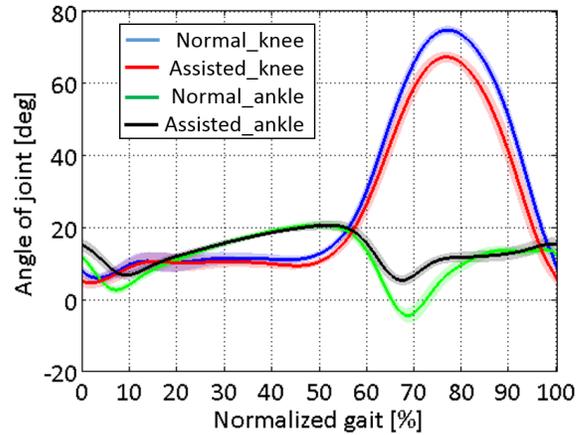
Fig. 9 (a) shows the comparison of the joint angle pattern of the assisted walking with the body twist path and normal walking. According to this graph, the assist increased the dorsal flexion angle by approximately 10 deg during the assisted time in the range of 60% to 80% of the gait cycle. This result indicated a decrease in the planter flexion, which meant a weak push-off at the timing of the toe-off. This weak push-off probably caused the approximately 8 deg of reduction in the knee flexion angle, which was larger than that of the arm swing path.

According to Fig. 9 (b), the extension torque was exerted at the knee joint, which probably affected the joint angle. However, the value of the knee extension torque did not differ from that of the arm swing path. Thus, the weak push-off probably decreased the knee flexion angle. On comparing the estimated joint torque shown in Fig. 9 (b) and the simulation result shown in Fig. 7, the pattern was found to differ in the range of 80% to 100% of the gait cycle. The change in the gait motion, which meant a weak push-off, resulted in the change in the time at which the torque was generated.

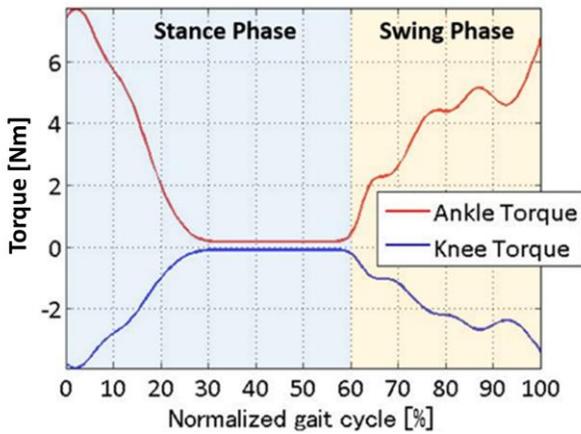
The MFC was then compared for various path in Fig. 10. The MFC of both the assisted trials became larger than that of normal walking, which meant that the assist device was able to decrease the risk of tripping. In addition, the assist of the arm swing path worked more effectively to increase the MFC than the assist of the body twist path. On the other hand, in the other subject, the MFC of the assist with the body twist path was smaller than that of normal walking as shown in Fig. 11. Because this greatly decreased the knee flexion angle, the height of the foot probably decreased despite the increased dorsal flexion in the body twist path. This result suggested that in order to maintain the knee flexion, which was probably realized by adjusting the string path not to prevent push-off, it is important to increase the MFC to avoid tripping. Thus, such individual differences have to be considered to achieve stable assist performance.



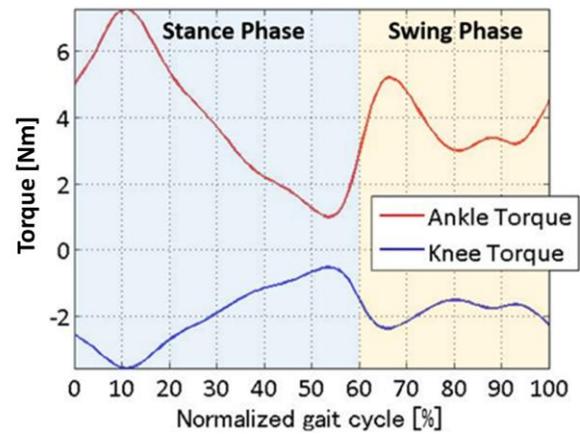
(a) Comparison of joint angles



(a) Comparison of joint angles



(b) Generated joint torque



(b) Generated joint torque

Fig. 8 Comparison of normal walking versus arm swing path

Fig. 9 Comparison of normal walking versus body twist path

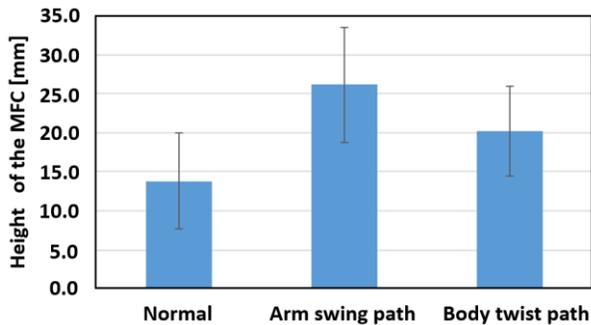


Fig. 10 Comparison of foot clearance

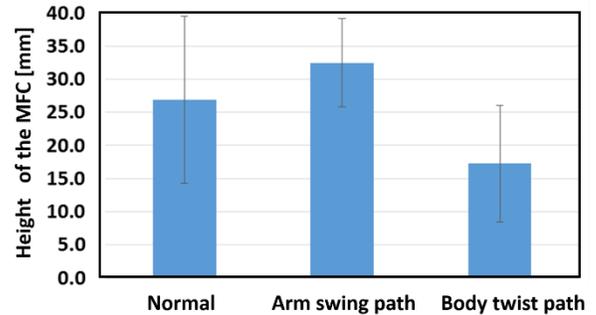


Fig. 11 Comparison of foot clearance of other subject

IV. CONCLUSIONS

In order to prevent falls induced by tripping in the elderly, we developed a non-actuated dorsal flexion assist device, which applied dorsal flexion torque to the foot from the upper body using string. Using a simulation that estimated the torque applied to the lower limb joints via the designed string path, two string-path patterns were determined. The first string path was the arm swing path, which had a long stroke during the gait cycle

and allowed the adjustment of the torque pattern by controlling the arm swing. The second string path was the body twist path, which was able to generate the dorsal torque stably using the strong muscle. The assist performance was then evaluated by comparing the gait motion between assisted walking and normal walking. Although the two string paths exhibited different characteristics, both paths successfully increased the dorsal flexion and MFC.

However, the result obtained from one of the subjects suggested that the weak push-off, which was caused by the collateral torque owing to the string path, decreased the MFC in the case of the body twist path through the reduction in the knee flexion. Thus, it is necessary to improve the string path to reduce the collateral torque and fit individual differences. Furthermore, the experiment will be conducted to many subjects, the difference of the result between subjects will be analyzed.

REFERENCES

- [1] S. K. Banala, S. H. Kim, S. K. Agrawal, and J. P. Scholz, "Robot Assisted Gait Training With Active Leg Exoskeleton (ALEX)," *IEEE Transaction on Neural System and Rehabilitation Engineering*, vol. 17, no. 1, 2009
- [2] CYBERDYNE official homepage, <http://www.cyberdyne.jp/index.html>, April, 2017
- [3] G. Taga, Y. Yamaguchi, and H. Shimizu, "Self-organized control of bipedal locomotion by neural oscillators in unpredictable environment," *Biological Cybernetics*, vol. 65, pp. 147-159, 1991
- [4] P. M. Mills, R. S. Barrett, and S. Morrison, "Toe clearance variability during walking in young and elderly men," *Gait&Posture*, vol. 28, pp. 101-107, 2008
- [5] R. Begg, R. Best, L. Dell'Orti, and S. Taylor, "Minimum foot clearance during walking: strategies for the minimisation of trip-related falls," *Gait&Posture*, vol. 25, pp. 191-198, 2007
- [6] A. H. Khandoker, S. B. Taylor, C. K. Karmakar, and R. K. Begg, "Investigating Scale Invariant Dynamics in Minimum Toe Clearance Variability of the Young and Elderly During Treadmill Walking," *IEEE Transaction on Neural System and Rehabilitation Engineering*, vol. 16, no. 4, AUGUST, 2008
- [7] R. Barrett, P. Mills, and R. Begg, "A systematic review of the effect of aging and falls history on minimum foot clearance characteristics during level walking," *Gait&Posture*, vol. 32, pp. 429-435, 2010.
- [8] Y. Honda, Y. Aso, A. Ibe, M. Katayama, T. Tamaru, "A Study of Effects of Foot Bath before Exercise on Fall Prevention," *Journal of Japan Society of Nursing Research*, vol. 35, no. 5, 2012
- [9] J. E. Pratt, B. T. Krupp, C. J. Morse, and S. H. Collins, "The roboknee an exoskeleton for enhancing strength and endurance during walking," *In Proc. of IEEE Int'l Conf. on Robotics and Automation*, pp. 2430-2435, 2004.
- [10] K. Ohashi, Y. Akiyama, S. Okamoto, and Y. Yamada, "Development of the Wire Driven Walking Assist Device with the Optimized Wire Passes," *Robomech2016*, Yokohama, Japan (in Japanese)
- [11] A. T. Asbeck, S. M. M. De Rossi, K. G. Holt, and C. J. Walsh, "A biologically inspired soft exosuit for walking assistance," *The International Journal of Robotics Research*, 2015
- [12] M. Wehner, B. Quinlivan, P. M. Aubin, E. M. Villalpando, M. Baumann, L. Stirling, K. Holt, R. Wood, and C. Walsh, "A Lightweight Soft Exosuit for Gait Assistance," *IEEE International Conference on Robotics and Automation*, Karlsruhe, Germany, May, 2013