Development of Walking Assistance Orthosis by Inducing Trunk Rotation Using Leg Movement: 1st Report on Prototype and Feasibility Experiment

Harutaka Ooki[1], Sho Yokota[1][a] and Akihiro Matumoto[1][b], Daisuke Chugo[2][c], Satoshi Muramatsu[3] and Hiroshi Hashimoto[4][d]

[1]*Dept. of Mechanical Engineering, Toyo University, Saitama, Japan*
[2]*School of Engineering, Kwansei Gakuin University, Sanda, Japan*
[3]*Dept. of Applied Computer Eng., Tokai University, Japan*
[4]*Adv. Insitute of Industrial Tech., Shinagawa, Japan*

Keywords: Walking, Assistance, Trunk, Human Powered.

Abstract: Walking is a whole-body movement including an upper body (trunk) and a lower body (pelvis and lower limbs), which is called the Spinal Engine Theory. Furthermore, there is a finding that the stride length and walking speed increases with the amount of trunk rotation. Based on this insight, there have been existing study that promotes rotation of the upper body to assist in walking. However, motors are used to assist the rotation of the upper body, which requires maintenance of the power supply and complicates the system. Therefore, this research aims to develop an orthosis that promotes gait by increasing the amount of trunk rotation without any active actuators. In particular, this paper reports on the basic experiment to confirm that the prototype can apply assistive torque to the trunk when the leg is raised.

1 INTRODUCTION

In contemporary society with a low birth rate and an aging population, the independence of the elderly has become a critical challenge. Extending healthy life expectancy is considered one solution to address this issue (Ministry of Health, Labour and Welfare, n.d.). According to a survey by the Ministry of Health, Labour and Welfare Japan, it has been shown that the physical functions that decline relatively early among the daily activities of the elderly is the abilities to move, such as walking and getting up from a seated position (Ministry of Health, Labour and Welfare, n.d.). Therefore, actively engaging in walking exercises in daily life is effective as an early preventive measure against activities of daily living disabilities among the elderly. Furthermore, it is beneficial for physical health and contributes to improvements in subjective well-being, life satisfaction, and sense of purpose (Murata, 2009). Due to these reasons, various walking assistive devices have been researched and developed. Examples include: the "Power Suit HAL," (CYBERDYNE, n.d.), which measures muscle activity and assists lower limb movement during walking. The "aLQ," (Imasen, n.d.) incorporates passive walking and uses springs to assist lower limb movement. "TPMAD" (Hashimoto, 2018), which utilizes motors to assist trunk rotation and walking. These devices aim to support walking and range from practical applications to those with proven efficacy in rehabilitation.

However, devices such as the "Smart Suit HAL" and "TPMAD" require power sources, such as batteries, due to the use of motors and sensors. In contrast, the "aLQ" utilizes passive materials attached to the legs without power sources, leveraging their elastic force to assist leg movements. However, this device does not consider upper body movements. Since walking involves full-body activities, including

[a] https://orcid.org/0000-0002-8507-5620
[b] https://orcid.org/0000-0002-3004-7235
[c] https://orcid.org/0000-0002-3884-3746
[d] https://orcid.org/0000-0003-2416-8038

Ooki, H., Yokota, S., Matumoto, A., Chugo, D., Muramatsu, S. and Hashimoto, H.
Development of Walking Assistance Orthosis by Inducing Trunk Rotation Using Leg Movement: 1 st Report on Prototype and Feasibility Experiment.
DOI: 10.5220/0012235200003543
In *Proceedings of the 20th International Conference on Informatics in Control, Automation and Robotics (ICINCO 2023) - Volume 1*, pages 740-746
ISBN: 978-989-758-670-5; ISSN: 2184-2809

Copyright © 2023 by SCITEPRESS – Science and Technology Publications, Lda. Under CC license (CC BY-NC-ND 4.0)

the upper body, it is also necessary to consider the upper body's activities.

Therefore, this research aims to develop a walking assistance orthosis that can be easily used daily and induces trunk rotation by using leg movements without motors. In this paper, we model the trunk movement and conduct simulations to examine the maximum torque the proposed orthosis assisting the trunk rotation. Additionally, we prototype the orthosis and verify its capability to provide assistive torque to the trunk by leg movements.

2 THE CONCEPT OF THE PROPOSED ORTHOSIS

Generally, walking is a whole-body movement involving the lower and upper limbs. This movement is governed by the Spinal Engine theory (Gracovetsky, 1987), in which energy exchange occurs between the trunk and the legs through the spinal column. During this process, the trunk rotates opposite to the pelvis rotation, thereby controlling balance during walking. It is also considered that increased trunk rotation leads to longer stride lengths and higher walking speeds (Nishimori, 2006). Therefore, based on these insights, this study aims to indirectly assist walking by using lower limb movements to assist trunk rotation. Furthermore, we propose to simplify the system and eliminate battery management by assisting trunk rotation without the use of motors or other actuators.

Therefore, we consider using leg force as a source of force to induce the trunk rotation. In other words, the proposed orthosis converts the power generated by leg movements during walking into trunk rotation torque and utilize it to assist trunk rotation. Thus, the proposed walking assistance orthosis uses leg movements to assist trunk rotation. The specifications of the proposed orthosis can be summarized as follows.

1) Assisting trunk rotation using leg movement
2) Targeting users capable of independent walking, aiming to maintain and improve walking.
3) Lightweight, providing a comfortable fit to minimize user burden.

The primary feature of this orthosis is 1): Assisting trunk rotation by leg movements enhance stride length and improves walking speed. Existing walking assistive orthosis typically rely on external forces such as actuators or springs to supplement muscle strength during walking. In contrast, this orthosis aims to promote walking only using the user's own leg muscles without any actuators. Regarding feature 2), the orthosis is designed for users who want to maintain their health. The feature of 3) is an essential specification to ensure practical usability.

The conceptional design of the orthosis is depicted in Figure 1. The user wears the orthosis, including attachments for securing thigh wires. These wires are connected to a bobbin located at the waist via pulleys. The tension in the wires rotates the bobbin, and the resulting torque is transmitted through shoulder plates to induce trunk rotation. The motion of the orthosis can be summarized in the following four steps.

(i) The legs' swing motion pulls the wire.
(ii) The pulled wire rotates the wire bobbin via the pulley.
(iii) The torque of the bobbin is transmitted to the plates.
(iv) The torque of the plate is transmitted to the shoulders, inducing trunk rotation.

It is important to note that excessive assistance torque can increase the risk of injury. Therefore, to ensure safety, a torque limiter is inserted between the bobbin and plates to prevent the transmission of torque exceeding the estimated maximum required for assistance.

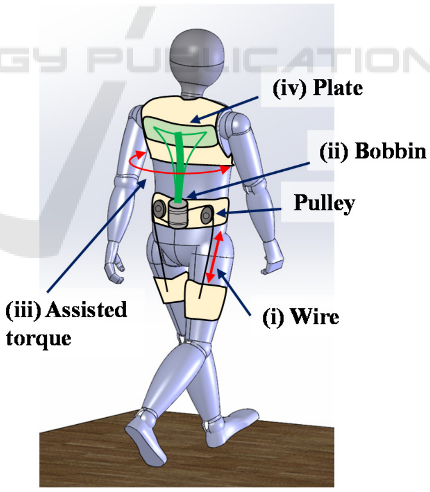

Figure 1: Conceptual image of the proposed orthosis.

3 ESTIMATION OF THE TORQUE OF THE TRUNK

The leg movements generate the assistive torque for the trunk rotation through wire, pully, and bobbin. We simulate the trunk rotation to study the torque that

SCITEPRESS SCIENCE AND TECHNOLOGY PUBLICATIONS

amplifies the effective trunk rotation angle for walking.

Figure 2 is a top view of a user. The *x*-axis indicates the walking direction of user, the grey circle represents the head, and the yellow oval represents the trunk. The trunk and head are simply regarded one rigid body as shown in Figure 3, and the torque around *z*-axis is considered. The torque around the *z*-axis is composed of the $\tau_h(t)$ meaning the rotational torque of the human muscle, and $\tau_a(t)$ being the torque around the center of rotation of the bobbin of this orthosis, which induces trunk rotation. This trunk rotation is formulated based on a (Irving, H. P., 2007) as (1).

$$J\ddot{\theta}(t) + D\dot{\theta}(t) + K_1 e^{K_2(\theta(t)-K_3)} - K_4 e^{K_5(K_6-\theta(t))} = \tau_h(t) + \tau_a(t) \quad (1)$$

Here J is the moment of inertia when considering the trunk and head as unified one rigid body, D represents the viscous friction resistance, and K_1 to K_5 are coefficients related to elasticity. Hereafter, $\tau_a(t)$ is referred to as assist torque. The values of each coefficient in the equation were taken from the research (Aoki, 1998) (Yamazaki, 2006). The moment of inertia J was calculated using the trunk mass (Irving, 2007) with a body weight of 69 kg. These values are shown in Table 1.

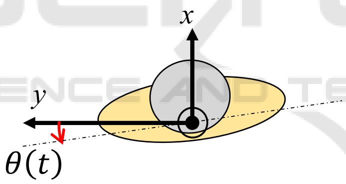

Figure 2: Coordinate setting (top view).

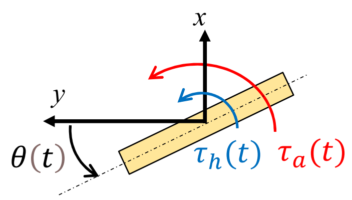

Figure 3: Trunk as a rigid body and torque around the axis of rotation (top view).

We investigate the maximum value of $\tau_a(t)$ the assist torque required to induce trunk rotation. For this purpose, it is necessary to identify the rotation torque $\tau_h(t)$ by the user's own muscles during walking in $\tau_a(t) = 0$ in (1). For this purpose, we referred to a Japanese typical trunk rotation angle during walking (Hashimoto, 2018), and exploratively determined the torque $\tau_h(t)$ so that the trunk rotation angle $\theta(t)$ in (1) would be the same as this reference angle. Figure 4 shows a reproduction and drawing of the approximate trajectory of the trunk angle (Hashimoto, 2018). $\tau_h(t)$ was identified exploratively so that the trunk rotation angle $\theta(t)$ in (1) to be same as Figure 4.

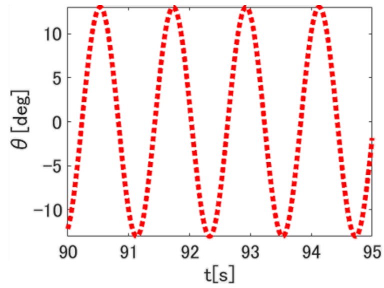

Figure 4: Trunk rotation angle during walking for $\tau_h(t)$ plotting the data from the study (Hashimoto, 2018)).

Table 1: Various coefficients.

J [kg · m²]	0.221		K_3 [rad]	30.0
D [kg · m²/s]	0.020		K_4 [Nm]	94.75
K_1 [Nm]	0.679		K_5 [1/rad]	0.181
K_2 [1/rad]	0.181		K_6 [rad]	-30.0

3.1 User's Own Torque of Trunk

Since the angular frequency of the reference trajectory in Figure 4 was 5.23 rad/s, the angular frequency of $\tau_h(t)$ is also 5.23 rad/s. Therefore, $\tau_h(t)$is expressed as

$$\tau_h(t) = A\sin(5.23t + \phi_1). \quad (2)$$

The amplitude A and phase ϕ_1 in (2) were obtained exploratively, and the reference trajectory in Figure. 4 coincides with $\theta(t)$ in case of (3) without assist torque, i.e., $\tau_a(t) = 0$, which is the only torque by user's own muscles.

$$\tau_h(t) = 6\cos(5.23t) \quad (3)$$

Figure 5 shows the trajectory of the rotation angle $\theta(t)$ in the state (without the orthosis) without assist torque and only rotation torque by the user's muscles, superimposed on the reference trajectory in Figure 4 . The red dashed line in Figure 5 is the reference trajectory, and the blue solid line is the numerical solution of (1) with (3) and $\tau_a(t) = 0$. As this Figure shows, the two lines are fitted, thus (3) represents the torque by the human muscles for trunk rotation during walking in the model of (1).

SCITEPRESS SCIENCE AND TECHNOLOGY PUBLICATIONS

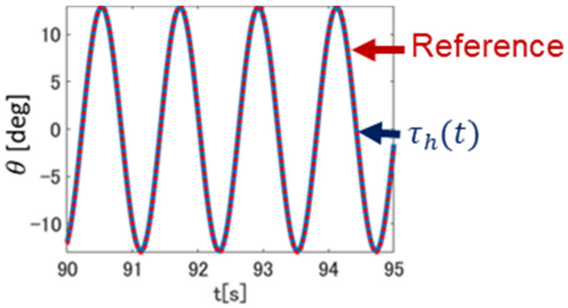

Figure 5: Reference trajectory (red dashed line) of trunk rotation angle and trunk rotation angle with $\tau_h(t)$ = 6 cos (5.23t), $\tau_a(t)$ =0 (blue solid line).

3.2 Required Assist Torque

we find the approximate maximum assist torque $\tau_a(t)$ required for assisting trunk rotation. When determining the assist torque, there is a question as to what degree of rotation angle is sufficient to facilitate walking safely. In this paper, we focus on gait speed among the gait parameters and find the maximum value of assist torque $\tau_a(t)$ to satisfy the trunk rotation angle required to increase the gait speed.

According to the research (Nishimori, 2006), the walking speed is maximized when the trunk rotation angle is 20 deg. The angular frequency of the assist torque, $\tau_a(t)$, is the same as the frequency of user's own toque, $\tau_h(t)$, shown in (3). Thus, the assist torque, $\tau_a(t)$, is expressed as (4) with the angular frequency as in (3).

$$\tau_a(t) = B\cos(5.23t) \tag{4}$$

The coefficients of B was obtained in an exploratory manner so that the amplitude of $\theta(t)$ is 20 deg when $\tau_h(t)$ is in (3) (during walking). As a result, in case of

$$\tau_a(t) = 3.25\cos(5.23t), \tag{5}$$

the amplitude of $\theta(t)$ became 20 deg. Figure 6 shows the trajectory of $\theta(t)$ when inputting $\tau_a(t)$ (5) and $\tau_h(t)$ (3) to (1). As Figure 6 shows, the amplitude of the rotational angle of trunk, $\theta(t)$, reaches 20 deg in the steady state. Compared to the rotation angle of 13 deg without assist torque (blue line in Figure 6), the rotation angle increased by 7 deg with the addition of assist torque. Based on these results, the guideline for the maximum assist torque of this orthosis is 3.25 Nm.

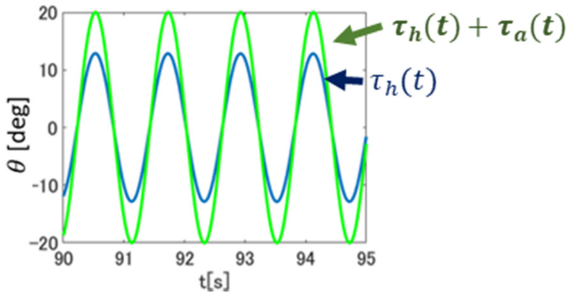

Figure 6: Comparison of trunk rotation angle without assisted torque(blue) and trunk rotation angle with assisted torque $\tau_a(t)$(green).

4 PROTOTYPE

Figure 7 shows an overall view of the prototype. The prototyped orthosis consists of the Top Part and the Bottom Part connected by the flexible shaft; the Bottom part extracts torque from the leg movements and the Top Part transmits that torque to the trunk. At the Bottom Part, both ends of a single stainless-steel wire ϕ1.25 mm are attached to the back of the each thigh via nylon band. The centre of the wire is wound around a bobbin. When one leg is raised, the wire is pulled, and the bobbin rotates. This rotational torque becomes the assist torque and is transmitted to the Top Part via the flexible shaft. In other words, the leg movement assists the trunk rotation by using the orthosis.

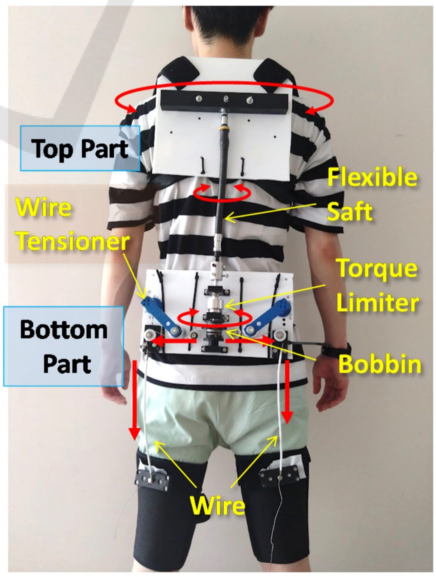

Figure 7: Overview of orthosis.

Figure 8 shows the mechanism of the Bottom Part where the bobbin is installed. To ensure that the torque transmitted from the bobbin to the Top Part via the flexible shaft does not exceed the maximum assist torque, 3.25 Nm shown in (5), a torque limiter is inserted between the bobbin and the flexible shaft. The cut off torque of the torque limiter is adjustable to individual physical characteristics in the range of X to Z Nm. The radius of the bobbin was set to 0.04 m, in order to exceed the maximum torque of the torque limiter. The thigh flexion moment during walking is about 30 Nm (Yamamoto, 2003). The distance from the root of thigh to the nylon band is about 0.3 m. Therefore, the bobbin rotates a maximum of approximate 4 Nm. Since there is always slack in the wire, the tensioners are placed along the wire's path. In Figure 8, the wire paths are drawn with red line to easily identify it.

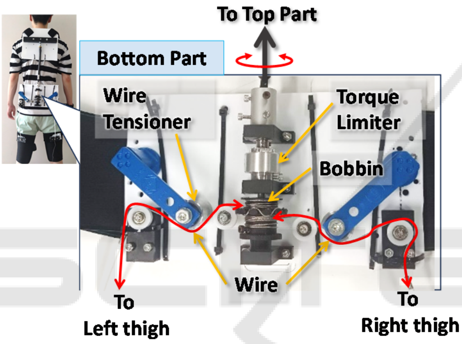

Figure 8: Bottom part of the prototype.

5 FEASIBLITY EXPERIMENT OF FORCE TRANSMISSION

This experiment aimed to confirm the conversion of leg movement to trunk rotation torque, which is the primary function of the proposed orthosis. Although it is naturally essential to verify the walking promotion and safety, the first step is to verify the generation of rotation torque of the trunk by the wire drive. The reason is that without the generation of rotational torque, the evaluation of gait promotion and safety cannot be performed.

5.1 Method and Preparation

In the experiment, subjects were asked to wear the prototype and flex their thigh to an arbitrary angle from a standing position. Then, the generation of the rotation torque is examined. In other words, the relationship between the leg flexion angle and the rotation torque is measured. A motion capture system, PERCEPTION NEURON PRO by NOITOM, was used to measure the leg flexion angle. The flexion angle, θ_{leg}, was determined with the hip joint as the origin and the direction of leg flexion as positive shown in Figure 9.

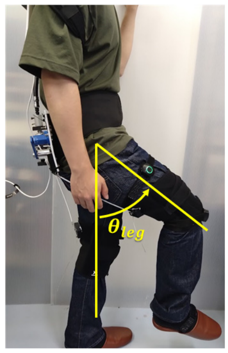

Figure 9: Definition of flexion angle θ_{leg}.

The rotation torque is calculated based on the measured force on the shoulder pressing against the subject's upper body using the FSR-406 pressure sensor. Figure 10 shows the sensor layout and jig. The pressure sensor was placed at a distance of 0.1 m from the centre of rotation of the gyration. In addition, a hemispherical jig with a diameter of 45 mm was attached so that the sensor's sensing area always touched the subject's shoulder. A sensor was calibrated in advance to convert sensor output (voltage) and physical quantity (force).

The subject was asked to flex the right leg at any time and any angle while supporting their posture by touching the wall with one hand, as shown in Figure 9. The flexion angle and torque were measured by performing this operation three times. There were two subjects shown as Table 2.

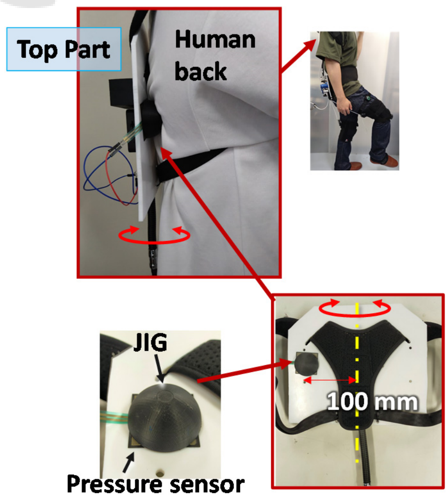

Figure 10: Pressure sensor and its jig.

Table 2: Information of subjects.

	Subject A	Subject B
Age	23	22
Weight [kg]	62.6	60.0
Height [mm]	1653	1662

5.2 Result

The results are shown in Figure 11, Figure 12. Figure 11 shows an example of the results for subject A. The orange line is the leg flexion angle, and the blue line is the rotation torque calculated from the force of the orthosis pressing on the shoulder. This figure shows that the rotational torque increases with increasing leg flexion. Figure. 12 shows an example of the results for Subject B. As in Figure 11, the rotation torque increases with leg flexion. Similar trends were observed in other measurements. The results show that the wire is pulled, the bobbin rotates, and the rotation is transmitted to the shoulder, generating a rotational torque that pushes the upper body.

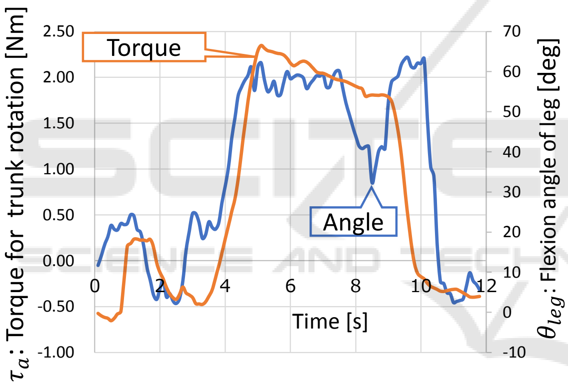

Figure 11: Example of subject A's rotational torque and leg flexion.

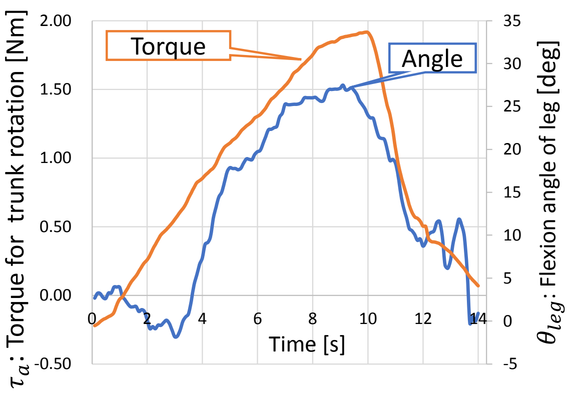

Figure 12: Example of subject B's rotational torque and leg flexion.

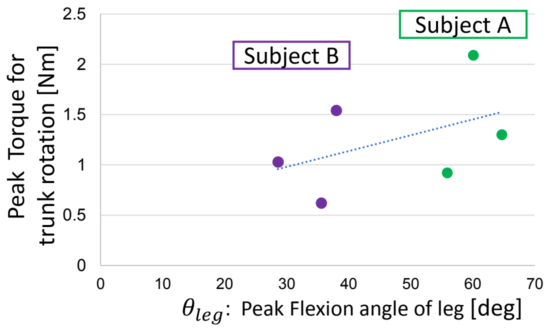

Figure 13: Peak of flexion angles and torque of both subjects.

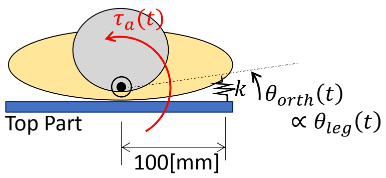

Figure 14: Orthotic and upper body mechanics model.

Figure 13 plots the maximum torque versus leg flexion angle for Subjects A and B. The green and purple dots represent the results for subject A and subject B, respectively. The dotted lines represent the linear approximations of these data. Although this is insufficient data for meaningful analysis, but indicative trend. The trend is that the torque is proportional to the flexion angle of the leg. Namely, the slope of this approximate line can be regarded as the equivalent spring coefficient, k, between the Top Part of the orthosis and the subject's shoulder, as shown in Figure 14.

Let $\theta_{orth}(t)$ be the rotation angle of the orthosis, and $\theta_{orth}(t) \propto \theta_{leg}(t)$, because the angle $\theta_{orth}(t)$ varies with the flexion of the leg $\theta_{leg}(t)$ via wire, pully, bobbin and flexible shaft. If the elasticity between the orthosis and the upper body of subject is expressed as an equivalent spring coefficient k, the torque also increases in proportion to the angle $\theta_{orth}(t) \propto \theta_{leg}(t)$, based on the torsion spring principle. This trend can be roughly seen in Figure 13.

Therefore, it was confirmed that the basic function of the orthosis, i.e., to convert leg movement into trunk assist torque, is feasible.

6 CONCLUSION

This research aims to develop an orthosis that promotes gait by increasing the amount of trunk rotation without any active actuators. In particular, this paper reported on the prototype and basic experiment to confirm that leg movement can be converted into an assist torque to the trunk.

The experimental results confirm the feasibility of the proposed orthosis to apply assist torque to the trunk. Limitations of this paper include the limited number of subjects and that torque was measured in a standing leg flexion rather than in a walking position.

The next phase of the study will involve a comprehensive experiment with a larger number of subjects. We will employ a treadmill to quantify the assist torque during walking, assessing the validity of the torque thresholds. Furthermore, leveraging the capabilities of both the treadmill and motion capture systems, we will determine the rotation angle of the trunk, the flexion angles of the legs, and the arms during gait. This experiment will highlight the effectiveness of our proposed orthotic.

REFERENCES

Aoki, K., & Yamazaki, N. (1998). The Role of Joint Resistance by Passive Tissues in Human Bipedal Walking. Society of Biomechanism Japan, 14, 59–68.

CYBERDYNE, (n.d.). What's HAL?. https://www.cyberdyne.jp/products/HAL/index.html

Gracovetsky, S. A., & Lacono, S. (1987). Energy Transfers in the Spinal Engine. Journal of Biomedical Engineering, 9(2), 99–114.

Hashimoto, K., Tanaka, T., & Kusaka, T. (2018). Trunk and Pelvic Rotation Assist for Walking Assist and Running Augmentation. Transactions of the JSME, 84(864), 18–64. https://www.jstage.jst.go.jp/browse/transjsme/-char/ja/

Imasen electric industrial Co., LtdCO, (n.d.). alQ. http://www.imasen.co.jp/alq.html

Irving, H. P. (2007). Physics of the Human Body: Biological and Medical Physics, Biomedical Engineering. Springer-Verlag GmbH & CO.KG.

Nishimori, T., & Ito, A. (2006). The Cross-Correlation Analysis of Trunk Rotation with Increased Gait Speed during Walking and Running. Rigakuryoho Kagaku, 33(6), 318–323.

Murata, S., Murata, J., Ota, H., Matsunaga, H., Oyama, M., & Toyota, K. (2009). The Effect of Walking on Physical, Cognitive and Psychological Functions of Community-Living Elderly: A Randomized Controlled Trial. Rigakuryoho Kagaku, 24(4), 509–515.

Ministry of Health, Labour and Welfare, Japan. (n.d.). Physical Activity or Exercise. https://www.mhlw.go.jp/www1/topics/kenko21_11/b2.html

Yamamoto, S. (2003). Biomechanics of Human Movement. Rigakuryoho Kagaku, 18(3), 109–114.

Yamazaki, N., & Tanaka, T. (2006). Development of a Human Joint Imitated Dummy. Biomechanism, 18, 175–185.