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Effect of different gait phase-based assist patterns of a wearable robot on gait motion

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***Abstract*— Wearable walking assist robots are expected to improve the quality of life of the elderly, and a few such robots have been developed recently. However, the human response to such robots depending on the assist pattern must be evaluated. We aimed to compare the gait under multiple assist patterns to determine what assist pattern is effective in improving gait. We developed two assist patterns related to human motion or torque exerted on the gait cycle. We conducted an experiment using our exoskeleton MALO and measured the joint angle, muscle activity, and interaction force. As a result, we found that hip and knee flexion is assisted by our robot. The maximum flexion angle of the knee joint was high in motion-based assist, and a decrease in biceps femoris and gluteus maximus muscle activity was observed in torque-based assistance.**

I. INTRODUCTION

Presently, many developed countries have a high elderly population. A wearable walking robot is expected to improve the quality of life of the elderly. Wearable walking assist robots such as HAL [1] and HONDA Rhythm Assist [2] have been developed. The HAL detects a person's motion intention through myoelectricity and assists the wearer in moving their joint toward the intended direction. In contrast, the HONDA Rhythm Assist assists the user by estimating the gait phase and gait ratio to maintain a constant gait ratio. In this way, all assistive robots have a different assist strategy; they have been shown to be effective in assisting.

The relationship between the assist pattern and the change in gait motion has not been compared or evaluated on a single device. The human response to such robots must be evaluated depending on the timing of the application of the assist torque or the differences in joints to design future assist robots [3].

The aim of this study was to compare the gait under multiple assist patterns to determine which assist pattern is effective in improving the gait. We examined whether assistance decreases the energy required for gait and facilitates walking motion.

II. METHODS

A. Apparatus

The experimental scene is shown in Fig. 1. Subjects walked on a treadmill for 90 seconds per trial. The walking motion of the subjects was measured using a motion capture system (Motion Analysis, MAC 3D System). The markers were attached to the subjects, and the hip and knee joint angles were calculated. Muscle activity was measured via surface electromyography (TeleMyo DTS EM-801, Noraxon) during walking. The surface electromyograph (EMG) was attached to the subject's right leg. To measure the activity of the hip and knee flexors, surface EMG electrodes were attached to the sartorial muscles, gluteus maximus, biceps femoris, and rectus femoris. To measure the interaction force between the body and the robot, a six-axis force sensor (WEF-6A200-4-RCD, Wako-tech Inc.) was attached to the robot. This force sensor was attached to the thigh and lower leg cuffs.

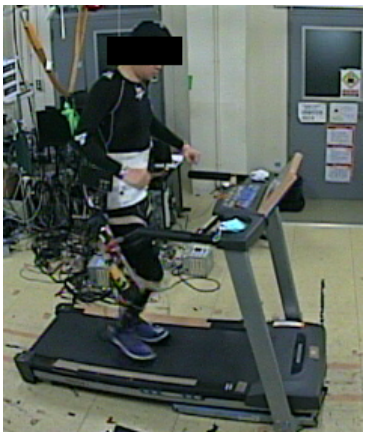

Fig. 1. Experimental conditions

The physical assistance robot, MALO, which includes a corset and long leg braces, is shown in Fig. 2. It has one degree of freedom in the hip, knee, and ankle joints. The robot is fixed to the wearer via the corset that is fixed to the torso, belts attached to the thigh and shank, and shoes. The hip and knee joints of the MALO are equipped with DC motors (Maxon RE40), which can apply sagittal torque to the respective joints. To detect the gait phase and the gait cycle, a portable force plate (M3D-FP Tec-Gihan) was attached to the soles of the MALO.

B. Assist pattern

The assist pattern was designed based on the periodicity of the gait cycle. The gait phase was determined by detecting the heel contact using a force plate. The maximum torque applied by the assist was 14 Nm at the hip joint and 11 Nm at the knee joint.

Furthermore, few previous studies have focused on the effect on gait by changing the assist timing of each joint parametrically [4][5]. In this study, two patterns of assistance were developed based on the following hypothesis.

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Fig. 2. Lower limb exoskeleton MALO

The first is assist for motion control, and the outline of the assist is shown in Fig. 3. The solid line shows the angular velocity of the joint during steady-state walking, and the dotted line shows assisted torque. The gait cycle is defined as the time period in which one foot contact the ground (0%) to when the same foot again contact the ground (100%). This assist is designed based on the motion of the joints of a healthy person. In this assist pattern, an extension torque is applied to the hip joint in the range of 0–40% and flexion torque in 45–85% of the gait cycle, and the flexion torque is applied to the knee joint in the range of 40–70% and extension torque in 70-100% of the gait cycle. These torques act in a direction that assists and facilitates the motion of the joint angle. Therefore, it is assumed that the push-off and swinging motion of the leg is promoted because the torque rises when the leg starts to move. On the other hand, the direction of assist torque and internal torque of human are opposite when human decelerates joint motion.

The second one is assist for torque control, and the outline of the assist is shown in Fig. 4. The solid line shows the internal torque of a person during steady-state walking [6], and the dotted line shows the assisted torque. In this assist pattern, extension torque is applied to the hip joint by 0–30% of the gait cycle, flexion torque by 35–75%, and extension torque by 80-100% of the gait cycle. Extension torque to the knee joint is applied by 5-30% of the gait cycle and flexion torque by 70–100%. These torques act in a timing when the joint torque is required. Joint torque is exerted before joint motion. A torque that supports the body weight and cancels the inertia of the body is required. By reducing the joint torque required by the muscles, it is expected that metabolic cost of walking may decreased[7].

C. Protocol

In this study, the experiment was conducted on two subjects who were 24-25 years old. The experiment was conducted with the approval of the Nagoya University ethics committee (approval number 20-10). In the experiments, the steady-state gait with the two aforementioned assist patterns and with only friction compensation such that the rotational resistance of the motor was zero were measured. In this study, the friction compensation assist is described as no assist. Walking was controlled at a speed of 3.0 km/h with a cadence of 95 stride/min for each assisted pattern. The comfortable speed and cadence were adjusted for subjects before recording the trial.

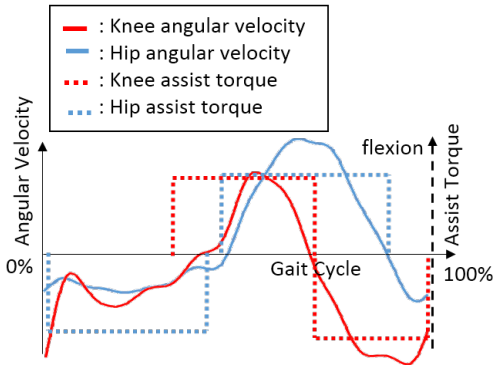

Fig. 3. Assist for motion control

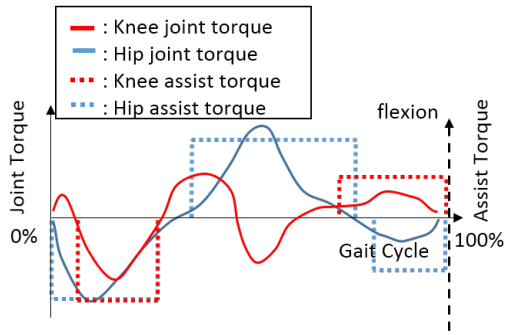

Fig. 4. Assist for torque control

D. Data analysis

The hip and knee joint angles were calculated from the feature points measured by the motion capture system. The joint angles on the sagittal plane were calculated from the trochanter, knee, and ankle markers. These were defined as 0 degrees in the upright position at the beginning of the experiment. These data, including the force sensors, were analyzed for 60 s of a 90-s walk. A force plate attached to the heel of the robot was used to determine heel contact, and data were separated for each walking cycle. The mean and standard deviation (SD) of all the cycles were then calculated.

III. Results

A. Controlled gait

The joint angles of subject 1 under controlled walking conditions are shown in Fig. 5 and Fig. 6. The data were divided based on the heel contact of the right leg. The lines represent the means of all cycles, and colored areas

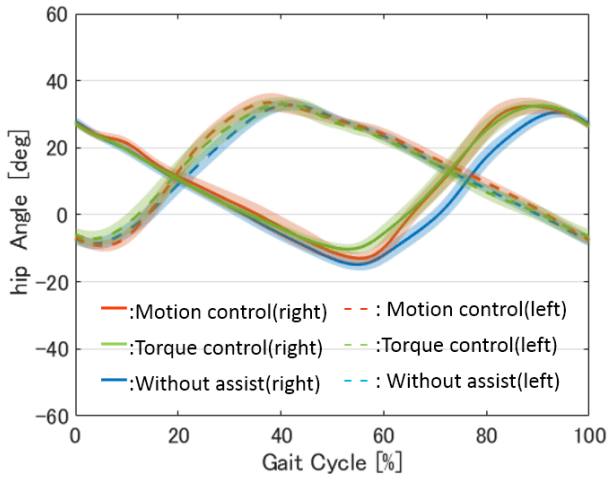

Fig. 5. Hip angle of Sub.1 at controlled gait. Data were averaged across gaits, and shaded regions represent SD.

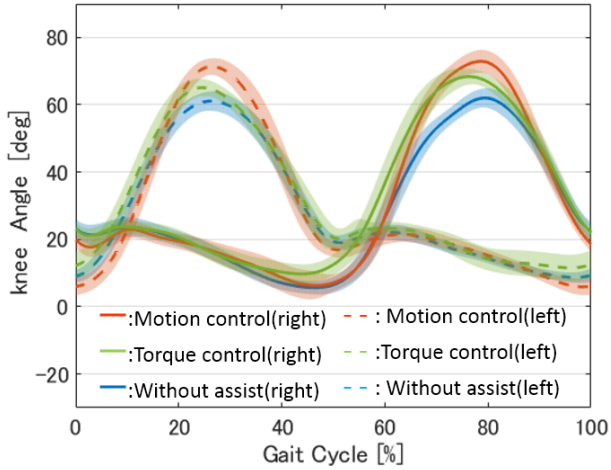

Fig. 6. Knee angle of Sub.1 at controlled gait. Data were averaged across gaits, and shaded regions represent SD.

are in the ± SD. These figures show an increase in the maximum flexion angle with assistance in both the hip and knee joints. Table 1 and Table 2 show the maximum flexion angle and maximum walking cycle of the right leg under controlled walking conditions. The maximum hip and knee flexion angles were larger, and the timing of the maximum flexion angle was earlier in the assisted condition than in the unassisted condition. In addition, the timing when the knee flexion angle reached its maximum was earlier in the assist for torque control than in the assist for motion control, which is faster in hip extension and flexion timing.

The compression forces of the right leg cuff sensor under controlled gait conditions of subject 1 are shown in Fig. 7 and Fig. 8. The lines represent the means of all cycles, and the colored areas are in the ± SD. In the assisted condition, lower leg compression was increased in the 50–70% gait cycle compared with the unassisted condition.

TABLE I

MAXIMUM FLEXION ANGLE OF THE RIGHT LEG AND THE GAIT CYCLE AT CONTROLLED GAIT(SUB.1)

	Without assist	Motion control	Torque control
Max. hip angle[deg]	30.6±1.58*	32.5±2.46	32.6±1.54
Gait cycle [%]	93.9	88.8	90.6
Max. knee angle[deg]	62.4±2.85	73.3±3.40	68.7±2.34
Gait cycle [%]	79.5	78.7	76.5

* mean angle ± SD

TABLE II

MAXIMUM FLEXION ANGLE OF THE RIGHT LEG AND THE GAIT CYCLE AT CONTROLLED GAIT(SUB.2)

	Without assist	Motion control	Torque control
Max. hip angle[deg]	22.8±1.63*	28.8±3.69	26.7±2.07
Gait cycle [%]	90.1	82.7	85.7
Max. knee angle[deg]	52.7±3.04	66.6±3.54	59.9±3.51
Gait cycle [%]	73.3	72.5	72.3

* mean angle ± SD

B. Comfortable gait

Comfortable speed and cadence of each assist condition were determined through gait practice. Comfortable gait speed and cadence are shown in Table 3. For both subjects, a faster speed was chosen when the assist was applied.

The maximum flexion angle and the maximum walking cycle of the right leg under comfortable walking conditions are shown in Table 4 and Table 5. An increase in the hip and knee angle was also observed during comfortable gait. In addition, the lower leg compression force of the assisted condition was increased in the 50–70% gait cycle compared to the unassisted condition.

IV. DISCUSSIONS

The lower leg compression force was increased in the 50–70% gait cycle compared to the unassisted condition. This is when the hip flexion torque is applied. It is considered that the hip flexion torque pushed the lower leg in the forward direction and increased the flexion of the knee and hip joint. Pushing the support leg immediately before toe-off can be an

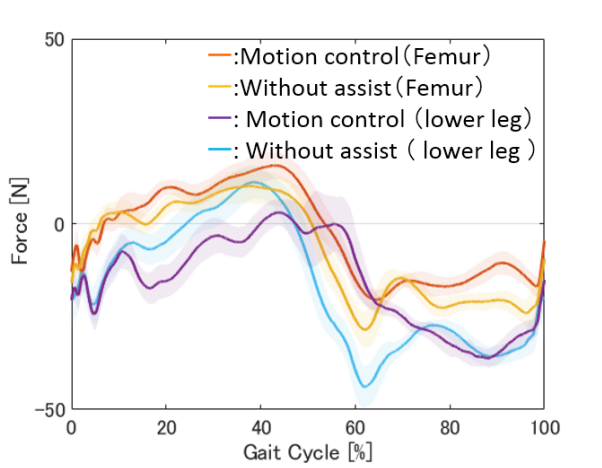

Fig. 7. Compressive force when motion control is applied. Data were averaged across gaits, and shaded regions represent SD.

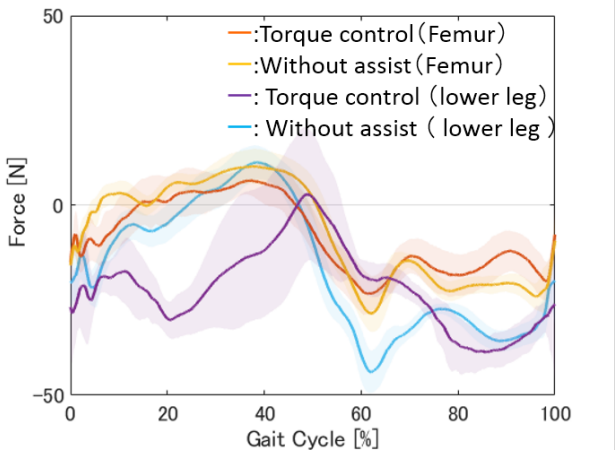

Fig. 8. Compressive force when torque control is applied. Data were averaged across gaits, and shaded regions represent SD.

TABLE III

COMFORTABLE GAIT SPEED AND CADENCE OF SUBJECTS

Assist pattern	No assist	Motion control	Torque control
Walking speed of Sub.1	2.5	3.1	3.5
Cadence of Sub.1	75	95	100
Walking speed of Sub.2	2.6	3.0	3.0
Cadence of Sub.2	80	85	85

Walking speed is expressed in km/h, and cadence in steps/min

effective assist [8]. Furthermore, the knee flexion angle for motion-controlled assistance is greater than that for torque-controlled assistance. This may be due to the fact that the knee flexion assistance was applied to the initial swing leg only when motion control assistance was applied.

In addition, in the torque control assist, an increase in tensile force on the lower leg at the early stance phase was observed. At this time, the knee joint extension torque was applied. Therefore, the knee extension torque may have increased. In the initial stages of the stance, the activity of the hip extensors was reduced in the assisted condition compared to that in the unassisted condition. This may be because the hip extension torque applied by the MALO altered the torque of the subject. In addition, the activity of the biceps femoris during the late swing phase was smaller in the torque base than in the motion base of subject 1. This trend can be understood as an effect of the hip extension and knee flexion torque applied by MALO.

TABLE IV

MAXIMUM FLEXION ANGLE OF THE RIGHT LEG AND THE GAIT CYCLE AT COMFORTABLE GAIT(SUB1)

	Without assist	Motion control	Torque control
Max. hip angle [deg]	28.1±1.55*	32.3±2.43	33.4±1.37
Gait cycle [%]	93.2	88.8	90.3
Max. knee angle [deg]	57.1±2.41	75.8±3.51	69.3±1.44
Gait cycle [%]	76.7	77.5	75.7

* mean angle ± SD

TABLE V

MAXIMUM FLEXION ANGLE OF THE RIGHT LEG AND THE GAIT CYCLE AT COMFORTABLE GAIT(SUB2)

	Without assist	Motion control	Torque control
Max. hip angle [deg]	25.7±2.44*	28.7±4.55	28.7±2.30
Gait cycle [%]	90.3	83.1	85.7
Max. knee angle [deg]	52.1±3.67	66.2±6.55	62.0±2.92
Gait cycle [%]	74.8	72.4	74.7

* mean angle ± SD

V. CONCLUSION

In this study, we aimed to compare the gait under two assist patterns to determine which assist pattern is effective in improving the gait. In the experiment, subjects walked on a treadmill using a walking assist robot. Gait motion, muscle activity, and interaction force were measured.

Our results showed that hip and knee flexion was assisted by our robot, as shown by the changes in joint angles and interaction force. The changes in muscle activity indicated that hip extension may be assisted. It is also possible that torque control assistance reduced biceps femoris activity.

In this study, we were unable to show the difference in the effect of assistance of different patterns. However, it was shown that the magnitude of the maximum flexion angle of the knee joint and some muscle activities may differ for each assist type. We intend to show the difference in the assisted effect by increasing the number of subjects in future studies.

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