Analysis of Recovery Motion of Human to Prevent Fall in Response to Abnormality with a Physical Assistant Robot

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***Abstract*— A physical assistant robot is useful for elderly and disabled people, as well as workers in certain industries, as it enhances their locomotive capabilities. However, risk assessment has identified some unique risks such as wounding of the skin and fall because of the use of such physical assistant robots. Among these potential risks, falls can be particularly severe. This study focuses on the effect of a sudden interference to the motion of the wearer while using a physical assistant robot. Changes in the gait motion are measured by an experiment that considers the unexpected torques experienced by a healthy wearer. The results indicate that the wearer can effectively recover despite changes in the joint angles and centers of pressure that can be used to analyze the risk of fall with a physical assistant robot.**

I. INTRODUCTION

Greater longevity has increased the importance of quality of life (QOL) for elderly and disabled people. In addition, falling birth rates in some countries have led to a need for greater worker efficiency. A physical assistant robot is one solution to such problems. These are robots that are fixed to the wearer by cuffs, bands, and other means, and thereby apply torque to the wearer's joints. They support locomotion, rehabilitation, and manual tasks. Such close contact between the physical assistant robot and the wearer increases the importance of safety measures. Therefore, previous studies, such as that by Akiyama et al. [1], focused on the contact safety of the physical assistant robot, and ISO 13482 has recently been published to specify the levels of safety that service robots should satisfy.

However, the amount of consideration given to the risk of falls is insufficient despite the potential danger. Some of the strategies and tolerances of healthy humans toward certain kinds of perturbation were revealed by Yang et al. [2] and Zhou et al. [3]. However, the presence of a robot could change these motions even if the user can still apply these strategies to their motion, even if only partially. The effect of the change in physical characteristics was tested by Arellano et al. [4] and Meuleman et al. [5]. The results of these studies can help us consider the effect of the mass and inertia of the robot. However, the effect of the robot's actuation cannot be estimated.

Table I lists the results of a risk assessment for a physical assistant robot. Considering both the results of previous research and the contents of this table, we can see that the effect of the application of unexpected torques by the physical assistant robot has not yet been sufficiently considered, despite its importance. An applied torque will interfere with the motion of the wearer if it is applied at an inappropriate timing or in other than the required direction, regardless of whether or not it is caused by a failure. In the worst case, it could even lead to a fall. Both the severity of the hazard and the difficulty of its predictability increase the significance of this unexpected torque. Therefore, this study focuses on the effect of unexpected torques on the gait of a subject wearing a physical assistant robot as a basic approach to analyzing human reactions when using a physical assistant robot. An analysis of how humans fall when using a physical assistant robot would be helpful to evaluate the safety of a physical assistant robot.

II. SETUP OF TRIPPING EXPERIMENT

A. Apparatus

1) Measurement system: An overview of the experiment is shown in Fig. 1. There is a 5 m walking lane, 2 m of which is a runway. In the recording area, the attitude and position of the subject are recorded using a 3D motion capture system (MAC 3D system, Motion Analysis Corporation, USA) and a video recorder (SONY, Japan). The ground reaction force (GRF) is measured by force plates (M3D force plate, Tech Gihan Co., Ltd., Japan) attached to the soles of the subject's shoes. The recording of the subject's position and reaction force is done at 500 Hz. To monitor some features of the subject that are covered by other devices, cluster markers are used. After recording, a motion simulator (SIMM, MusculoGraphics, Inc., USA) is used to calculate the joint angles of the subject.

To ensure the safety of the subject, he wore a safety harness that was suspended from a gondola that moves along the walking lane. The gondola does not, however, support the weight of the subject, except in the event of a fall. In addition, a supporter was used to prevent sprains.

2) Physical assistant robot: A physical assistant robot was developed for use in the tripping experiment. This leg-type motor-actuated lower-limb orthosis (L-MALO), shown in Fig. 2, has two separate legs. Each leg is fixed to the wearer using cuffs and shoes. The cuffs are fixed to the wearer's thigh and lower thigh. This L-MALO has a 1-DOF foot joint and a 1-DOF knee joint. The hip joint is not restrained. Only the knee joints of the L-MALO are actuated by a motor (RE40, Maxon), and compensation for the rotational friction of the knee joint is applied. The total weight of the L-MALO is 8 kg.

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TABLE I

RISK FACTORS OF FALL

Category	Hazard source	Potential consequences
Factors related to physical assistant robot	Restraint of joint DOF	Loss of balance because it is impossible to perform the normal motion
	Decrease of joint motion range	Loss of balance because it is impossible to perform the normal motion
	Restraint and torque of joint motion	Loss of balance because there is a mismatch between purpose and motion of user
	Change of center of gravity and inertia moment	Loss of balance because mechanical characteristics are unfamiliar for user
	Mismatch of assist timing	Loss of balance because of unexpected torque
Environmental factors	Collision with obstacle	Loss of balance because there is sudden external force
	Collision with low obstacle	Loss of balance because it is impossible to swing the leg
	Instability of footing	Loss of balance because it is impossible to support body weight
	Narrow way	Loss of balance because it is impossible to keep supporting leg polygon

Note: Device failure and those factors which are not associated with contact (light, sound etc.) are omitted.

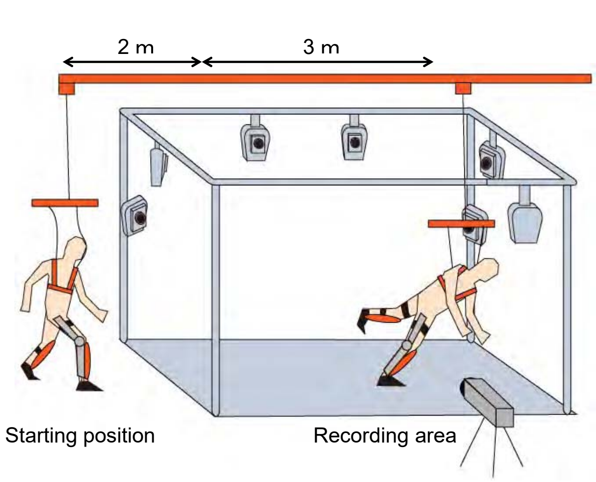

Fig. 1. Gait measurement system

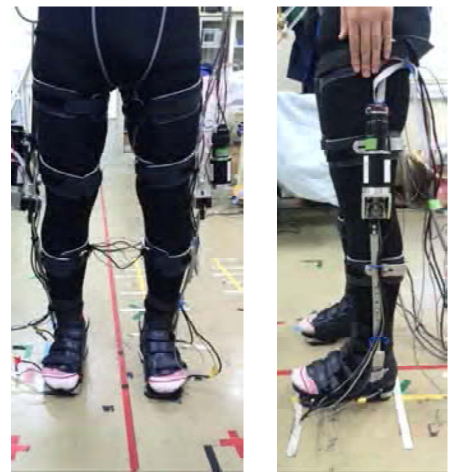

Fig. 2. L-MALO (left: front view, right: side view)

The L-MALO is designed to mimic a general power-assistance device. Therefore, gait cycle based assistance is adopted. The gait time is measured by counting the timing of the peak knee angle, and the mean gait time of the last three cycles is used to calculate the assistance timing for the next gait cycle. From the viewpoint of mechanical design, Pratt et al. [6] studied a physical assistant robot with a similar configuration. In addition, Lewis and Ferris [7] adopted a similar strategy for counting the gait timing. Although there are other types of robots such as [8], these robots are not considered in this study because of the difference in their assistance strategies.

The L-MALO applies 8.5 N·m of flexion torque in the 35%–50% range of the gait cycle. Then, in the 85%–100% range, an 8.5 N·m extension torque is applied. The gait cycle was calculated from the timing of the maximum knee flexion. Flexion assist is applied in the middle of the support phase, which helps to move the supporting leg backward. Extension assist is applied between the middle and end of the swing phase. This helps to extend the knee before the heel strikes the ground. The magnitude of the assist torque corresponds to about 30% of the torque at the knee joint, which is assumed to be the maximum assist rate that can be applied by a general gait-assist robot. An overview of the assist timing is shown in 3.

As the tripping method, an extension torque of 8.5 N·m was applied from the start of the stance phase of the left leg. Then, that torque was applied until the end of the gait. This torque represents the unique timing of a timing error in the assist algorithm because a person is not used to the application of an external extension torque in daily life. For safety, however, a mechanical limiter restricts the extension of the knee joint to no more than 180°.

B. Protocol

This experiment was conducted using a healthy male volunteer. Fig. 4 shows an overview of the experimental protocol. After putting on the L-MALO, force plate, markers, and safety harness, the subject was asked to walk so that the assistance being provided by the L-MALO could be adjusted. This was continued until the subject no longer felt uncomfortable. Then, the subject walked along the test lane at a comfortable speed while his motion was recorded. As the subject was walking, however, torques were randomly applied (the subject was not told when these torques would be applied). The number of cases was N = 8 (assisted cases) and N = 9 (tripped cases). The total time taken for the

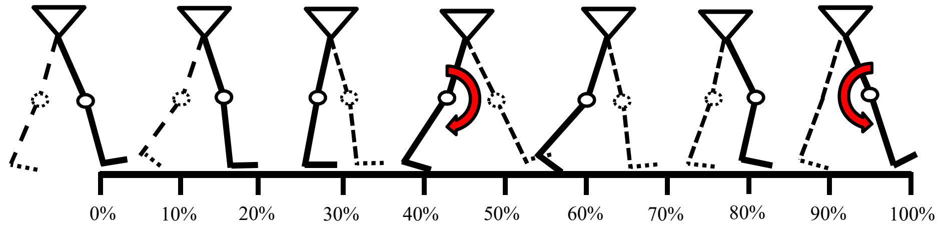

Fig. 3. Assist timing (continuous line: main leg, dashed line: opposite leg)

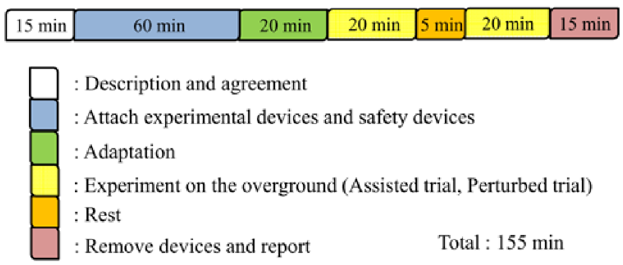

Fig. 4. Protocol of the tripping experiment

experiment was about 2.5 h.

This experiment was done with the permission of the IRB of Nagoya University.

C. Data processing

Data related to the joint angles and GRFs is filtered by a 6-Hz low-pass filter. Then, the offsets of the joint angles caused by inaccuracies in the positions of the markers are corrected using the data for a standing posture. Then, the data for the motion from one left heel contact to the next left heel contact is clipped to give the gait cycle. Each gait is normalized using the time period for the gait motion or that for the stance. Cases of normal assist are processed statistically.

The gait parameters for assisted walking were calculated from the positions of the motion capture markers. The forehead marker was used to calculate the subject's speed. The step length was assumed to be the distance between the heel markers of each leg in the direction of travel. The cadence was given by the frequency of the gait of one leg.

To compare the GRF values, the magnitude of the GRF is normalized by the weight of each subject and the horizontal axis is normalized by the stance.

Cases with an unexpected torque are normalized using the average gait time or stance time for the normal assist case and processed in the same way as for a normal case. This is because the aim of the normalization is to compare the motion after the application of an unexpected torque to that for a normal case, including the time scale. However, the subject would sometimes move out of the recording area. Therefore, the data for those cases in which an unexpected torque is applied is shorter than that for a normal case.

TABLE II

GAIT PARAMETERS OF ASSISTED WALKING

	Average	Standard deviation
Speed [km/h]	2.68	0.14
Step length (left) [m]	0.579	0.049
Step length (right) [m]	0.623	0.023
Cadence [/min]	36.9	0.93
Double support ratio [%]	28.0	0.97

III. RESULTS

A. Gait parameters

Table II lists the gait parameters for assisted walking. Gait parameters were calculated using the gait cycle for non-perturbed trials. The comparison of the gait parameters with the normal gait shown by Neumann [9] shows that the walking speed was low compared to the average speed of a healthy adult (4.98 km/h in normal gait). The main reason for this low speed was the small cadence (55 /min for a normal gait). In addition, the double support ratio is about 10% higher than that for a normal gait. A synchronous change in the speed and double support ratio is a general trend observed in a normal gait.

When using the L-MALO, the wearer walked more slowly. The reason for this change was that the weight and fixing of the L-MALO caused the subject to be more careful. Some physical assistant robots exhibit such a tendency, although it varies with the degree of usage of the physical assistant robot.

The difference in the step length of each leg suggests an asymmetry in the gait of the subject. In our experiment, the unexpected torque was applied to the left leg. Therefore, the subject probably changed his gait in preparation for the unexpected torque, even though the subject claimed that he was not doing so.

B. Joint Angle

The joint flexion angle for assisted walking is shown in Fig. 5. This result suggests that the gait of the subject resembles a general gait, except for the knee angle in the stance phase. In the stance phase (about 20–30%), the left knee angle appears to be small in comparison with that of a normal gait. This trend seems to be the result of extension assist at the end of the swing phase.

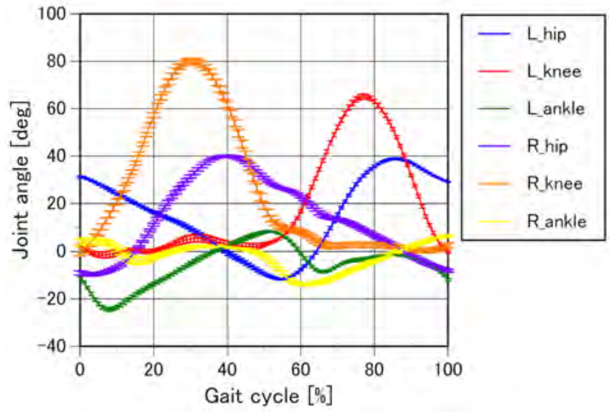

Fig. 5. Joint flexion angle on sagittal plane

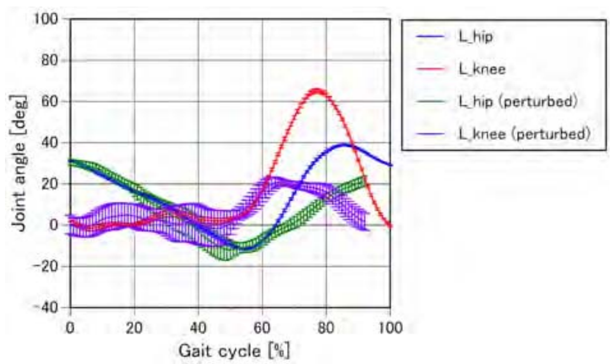

Fig. 6. Joint flexion angle of affected side

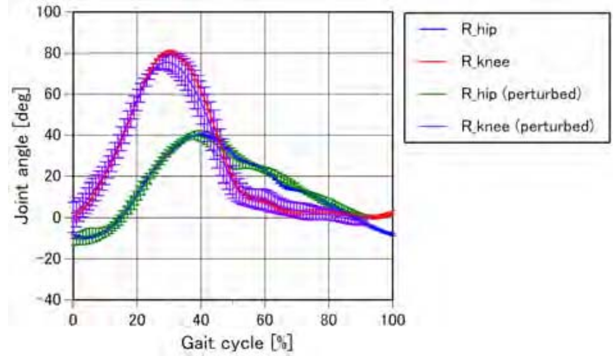

Fig. 7. Joint flexion angle of unaffected side

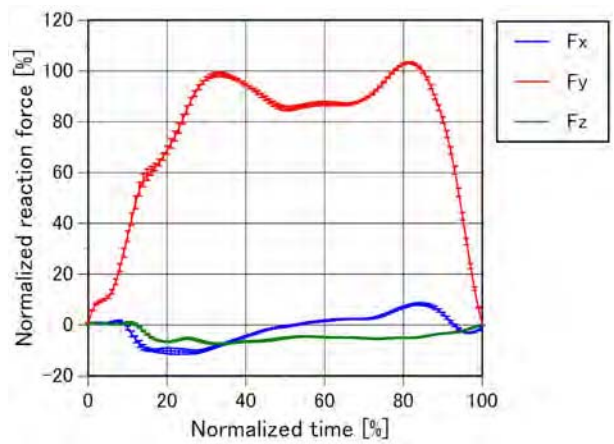

Fig. 8. Ground reaction force of assist walking

In addition, the maximum knee angle appears to be asymmetric. This trend also points to the subject changing his gait to ready himself for the unexpected torque.

Because of application of the unexpected torque, the joint angle of the perturbed leg changed after about 60% through the step, which is the early swing phase of the perturbed leg. Fig. 6 suggests that there is a decrease in the knee flexion. This is a result of the knee extension torque providing unexpected assistance. At the same time, the flexion of the hip of the perturbed leg became smaller. This is because the prevention of knee flexion decreases the clearance between the shoe and the ground, which the subject needs to swing his leg. In contrast, Fig. 7 suggests that the motion of the unaffected leg in the sagittal plane does not change drastically, even when the flexion of the other leg is disturbed.

C. Reaction forces

Fig. 8 shows the GRF for the left leg in normal assist trials. *F*x is the front-back direction and *F*y is the vertical direction. Therefore, *F*z indicates the lateral direction. Generally, *F*y immediately increases and reaches a maximum shortly after heel contact. Although *F*y of the early stance phase (0–30%) is smaller than this trend, the result does not appear unusual. The effect of the shoe thickness should be investigated to evaluate this trend. However, this trend in the GRF does not affect the motion of the next swing of the left leg, because the reaction force after the middle stance phase resembles the general gate motion.

Then, the GRF of the other leg is shown in Fig. 9. This graph is normalized by the time between the heel contact of the right leg and the heel contact of the left leg. Therefore, the GRF before the middle stance phase of the right leg is drawn in this graph. From this graph, the effect of the unexpected torque is not clear despite it being the support leg during a tripped swing. Therefore, this is further analyzed in the discussion.

IV. DISCUSSION

The L-MALO changes the gait of the subject to some degree. There is no major change in the joint angle, except for the knee angle in the stance phase. In the stance phase, flexion of the knee angle is suppressed. The reason for this phenomenon is the effect of the assist torque. Although the extension torque at the end of the swing phase can effectively prevent tripping and helps to support heel contact, it also prevents the start of flexing of the knee joint in the early stance phase. In addition, the knee flexion angle of the right leg increases compared to that of the average gait motion because the assist torque helps to flex the knee joint before the swing phase. Although this flexion angle is larger than

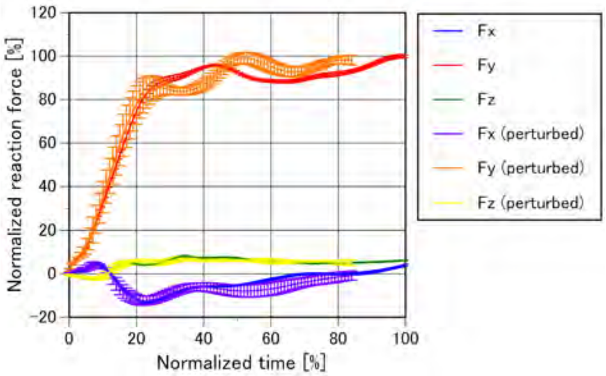

Fig. 9. Ground reaction force after tripping

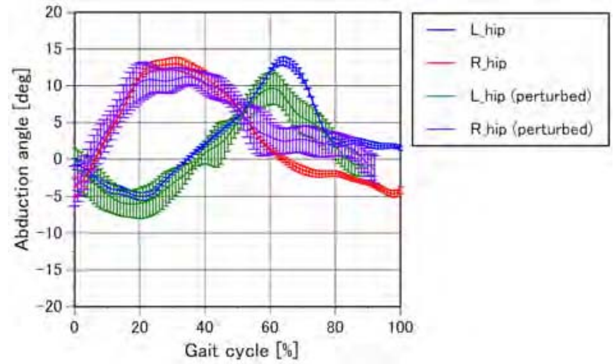

Fig. 10. Abduction angle of hip joint

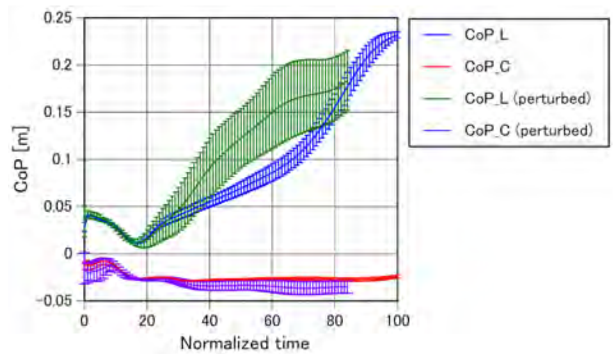

Fig. 11. CoP of support leg

that required to walk, it does not disturb the gait of a healthy person but will help a person with lower muscle power.

On the other hand, the L-MALO reduces the walking speed, because the cadence is lower. An increase in the double support ratio is also caused by a change in the speed. One reason for this change is the mass of the L-MALO. It was suggested by Browning [10] that the additional mass leads to a reduction in speed when the subject walks with a natural gait. Therefore, the subject will be affected by the mass of the L-MALO because it is not self-supporting in the swing phase, as each side of the L-MALO is independent.

Then, any unexpected torque drastically changes the joint angle of the affected leg in the sagittal plane. However, this change cannot be described only in terms of the sagittal plane because the straight leg disturbs the swing motion. The abduction angle of the hip joint is shown in Fig. 10. This graph suggests that the affected leg (left leg) moves outward in the swing phase due to the abduction of the hip joint of the unaffected leg. At the same time, the hip joint of the affected leg is adducted. This means that the pelvis inclines toward the unaffected side to move the leg being swung forward without dragging.

The same tendency can be observed in a pathological gait. A patient who has limited hip flexion exhibits symptoms that are similar to those of the subject when tripped. In this case, the patient swings his/her leg by using the rotation of the pelvis, and abduction of the support leg is also observed at the same time, which maintains sufficient clearance between the swing leg and the ground. The gait of the subject resembles this gait in spite of the differences in the limitation of the joint motion.

From the viewpoint of the risk of a fall, the center of pressure (CoP) of the support leg provides a useful means of analysis, in addition to the analysis of the reaction force. The distance between the CoP and the edge of the support polygon indicates the stability margin of the support leg. Fig. 11 shows the position of the CoP in the longitudinal direction (COP_L) and cross direction (COP_C). The larger deviation in the perturbed case indicates the randomness of the recovery motion. The graph in Fig. 11 shows that the CoP moves forward faster when the subject is tripped than in a normal assist case. In addition, the CoP changes to the inner side (negative direction on the graph) in almost every stance period.

Fig. 12 shows the trajectory of the CoP. This trajectory also suggests that the CoP moves toward the inner side of the support leg. This change can be described by the motion of the swing leg. The centrifugal force of the swing leg pulls the body toward the inside. As a result, the CoP suggests that the subject is likely to overlean toward a point in front of his trunk on the side of the leg being swung. However, the leg is swung in the direction of the subject's overlean. Therefore, the subject can easily perform this motion by stepping.

Therefore, in this experiment, although the unexpected assist changes the balance of the subject, it was found that the probability of a fall does not increase drastically. This result also suggests that the subject can change his gait strategy instantly. Although the gait changes from a normal gait, a perturbed gait seems optimal in terms of the limitation of the joint angle because it resembles the gait of a patient who has limited hip flexion. Therefore, the advanced adaptivity of the human gait is suggested by this experiment.

Then, the asymmetry of the gait of the subject should be considered. Even with normal assistance, the motion of the gait exhibits some asymmetry. One reason for this trend is the compensation by the subject. In this experiment, the unexpected torque was applied only to the left leg to avoid the effect of the dominant foot. This possibly changed

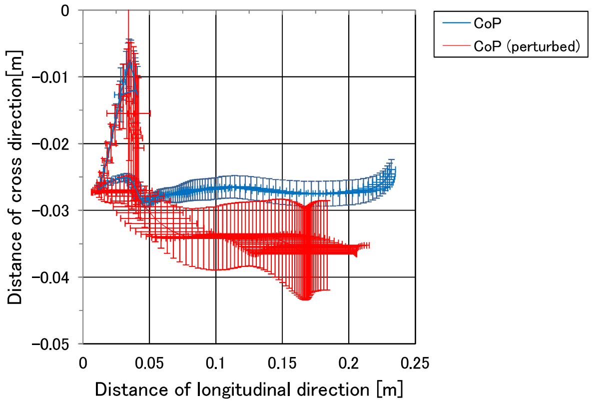

Fig. 12. Trajectory of the CoP of stance leg

the gait asymmetry. As a result, the knee flexion and hip abduction angle of the swing phase were affected. This change suggests that the subject moved his leg outward relative to an average gait. This motion seems to be effective for preventing falls in the case of tripping because it expands the support polygon. However, the effect of asymmetry is not so severe as to affect the subject's walking because he can continue with his gait. Although the results of this experiment perhaps place the risk of the subject fall at rather less than the actual value, the experiment clarifies the strategy applied by the subject to avoid a fall.

It is impossible to evaluate whether this motion is common for other people because of the limitation on the number of subjects and their individual attributes. Therefore, a more thorough inspection is required to reveal and classify the entire fall avoidance motion. However, the results of this study suggest that the recovery strategy applied by a user upon tripping differs from that applied upon falling or slipping.

V. CONCLUSIONS

An experiment was performed to analyze the reaction motion when an unexpected torque was applied by a physical assistant robot. In the experiment, an unexpected assist torque was applied to the subject, in addition to the assist torque. A physical assistant robot, the L-MALO, which was attached to each leg independently, was used in this experiment. The application of a sudden abnormal torque that prevents the flexion of the knee joint drastically changed the gait of the subject.

Because of the additional torque, the subject increased his hip abduction instead of reducing the flexion of his hip and knee joints. Although this gait also changed the trajectory of the CoP and decreased the stability margin, the risk of the subject fall did not seem to increase because it was easy for him to support himself with his trunk by stepping onto the leg being swung. In addition, a different gait strategy was adopted to reduce the effect of the limitation of joint motion when the subject was tripped. This strategy points to the instantaneous adaptivity of humans, in that it resembles the motion of a patient with limited joint motion.

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