The 2009 IEEE/RSJ International Conference on
Intelligent Robots and Systems
October 11-15, 2009 St. Louis, USA

Walking Assist Device with Bodyweight Support System

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***Abstract* — We aim at realization of a "user-friendly walking assist device" to support the life of the elderly, the work operation of production, and others. Based on a biomedical engineering analysis, we developed a new walking assist device with a concept to reduce the floor reaction force of the user of the device. The device comprises an axis layout in which expansion and contraction mechanisms disposed on the right and left are connected to a seat at a single point, a device layout in which the device is disposed along the inner side of the user's legs, and a mechanism that utilizes a circular arc-shaped rail. With these original layouts and mechanisms, we achieved a walking assist device that can be put on and taken off easily without the need to fasten the device to a user's body, and that can always maintain the assist force vector in the direction from the center of pressure of floor reaction force to the center of gravity of the user's body by using only two actuators. The device can reduce the load on leg muscles and joints in various movements and postures. It was confirmed that the device has an assist effect of reducing the user's leg muscle activities as well as the user's total body energy consumption when the user climbs stairs or performs a squat exercise.**

I. Introduction

AGING of society is expected to continue to progress around the world. The elderly tend to cut back on social activities when they lose confidence in their health and physical capability [1]. Meanwhile, a reduction of load has been desired for the work operations in standing or semi-crouching positions frequently found in the manufacturing industry.

In such a social background, Honda has been engaged in research on walking assist devices that support the leg capabilities of the elderly and workers since 1999. The goal of the research is to achieve a "user-friendly walking assist device." Aiming at a device that can assist a user at the will of the user, a joint moment assist method that uses the inverse dynamics analysis was proposed in 2001 [2]. In addition, a system weight compensation method was proposed in 2007 so as to prevent the weight of the assist device from being applied to the user [3]. Furthermore, with an examination from the perspective of biomedical engineering, a prototype of a walking assist device with bodyweight support system that is easily wearable with little restraint on the user's body and that can reduce the load on the user's leg muscles and joints while in various movements and postures was made public in November 2008 (Fig. 1) [4]. This paper deals with this walking assist device with bodyweight support system.

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II. Concept and Design Guideline of Assist Method

A. Concept of Walking Assist Method

In order to realize a user-friendly walking assist device, we analyzed walking movement from the perspective of biomedical engineering. A list of main forces involved in walking includes in vivo forces such as muscular force, joint moment and bone-on-bone force, as well as foot force, floor reaction force (FRF), gravity, and inertia (Fig. 2). Joint moment is an in vivo force in a rotational direction that is generated in opposite directions at the end face of the bones that face each other with a joint in between. The joint moment is generated mainly by muscular force. Bone-on-bone force is an in vivo force in a translational direction that is generated in

Fig. 1. Walking assist device with bodyweight support system.

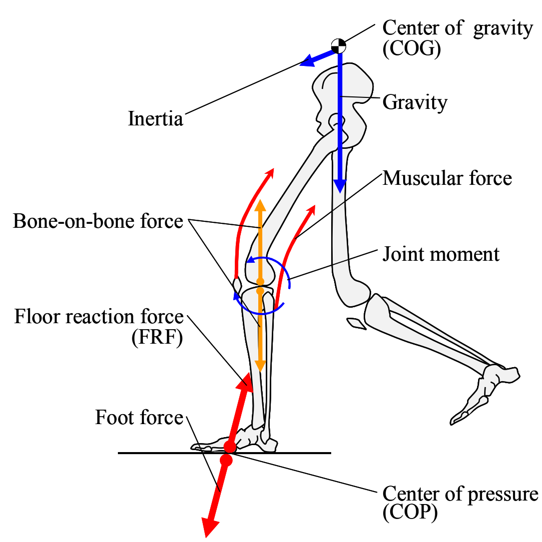

Fig. 2. Forces involved in walking.

opposite directions at the end face of the bones that face each other with a joint in between. The force is generated so as to receive all muscular forces that are present across a joint as well as the body weight. Foot force and FRF have an action-and-reaction relationship. Fig. 3 shows the relation of the aforementioned forces involved in walking. The relation of the forces is characterized in that a number of forces concentrate to a single point where the foot force meets the FRF. For this reason, we noticed that, if the FRF, which is a concentration point of the forces, is reduced, all the in vivo forces related to the foot force, which is a reaction force, can be reduced at the same time.

Based on the results of the aforementioned biomedical engineering analysis, we defined "reduction of the FRF of the user" as the concept of the assist method to aim at reduction of all in vivo forces. Specifically, the in vivo forces to be reduced are bone-on-bone force generated at the following three joints: hip joint, knee joint, and ankle, and the muscular force that generate joint moments.

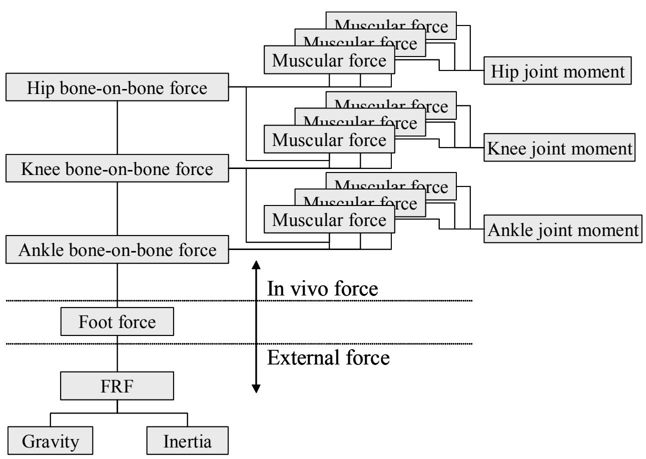

Fig. 3. Relation of the forces involved in walking.

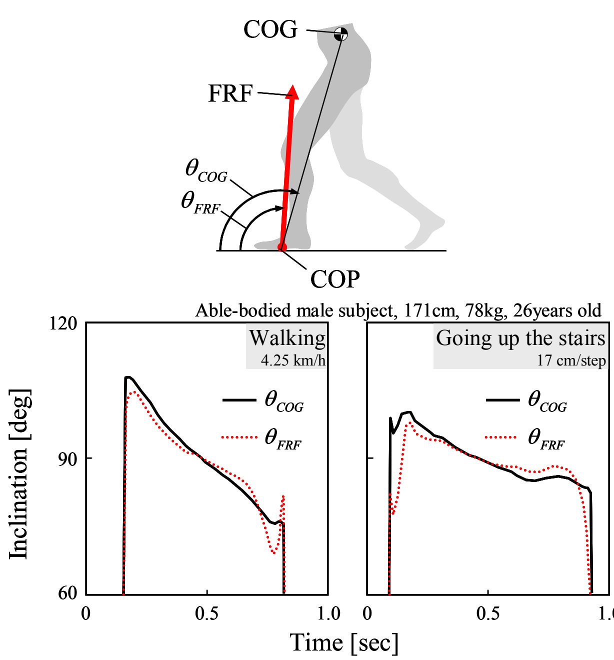

Fig. 4. Relationship between the vector of the floor reaction force and the center of gravity of the body.

B. Design Guideline

The assist force vector must be in the same direction as the vector of the FRF of the user in order to achieve a "reduction of the FRF of the user." When a person walks, translational motion is predominant with a slight involvement of rotary motion around the center of gravity (COG) of the person's body. Based on this, we framed a hypothesis that the vector of the FRF is mostly in the direction of the COG of a person's body, and confirmed the actual situations. Fig. 4 shows the location relationship between the FRF and the COG when a person walks and climbs stairs. The graph compares the slope of the vector of the FRF of a person and the slope of a line segment from the location of the center of pressure (COP) of a FRF toward the COG. From the graph, it can be confirmed that the values of the two slopes are close to each other, and that the vector of the FRF of a person extends more or less from the COP to the COG. Thus, we defined attainment of the assist force vector that extends from the COP of the FRF to the COG of the user as our design guideline (Fig. 5).

If a walking assist device that satisfies the design guideline

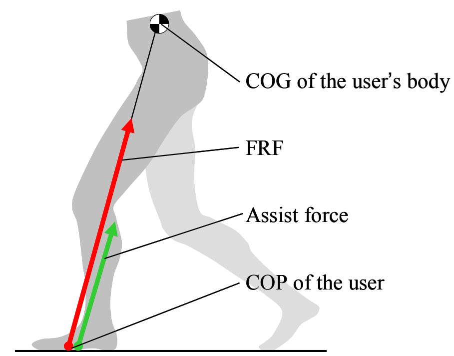

Fig. 5. Design guideline on assist force vector.

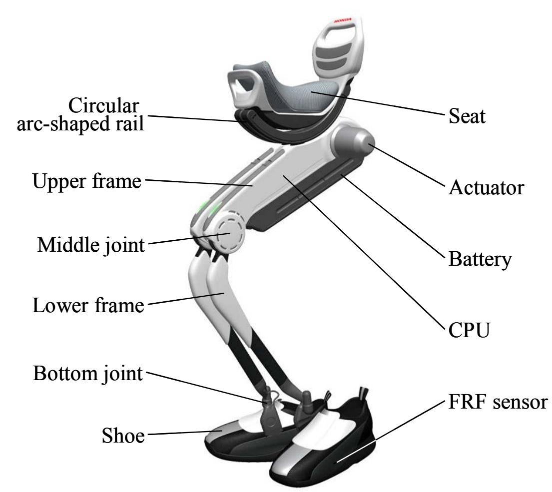

Fig. 6. Configuration of the walking assist device with bodyweight support system.

is achieved, the vector of the FRF that is applied to the sole of the user's foot will be reduced in amount without changing the direction compared to when the device is not used. In other words, the user is expected to have a feeling of weight loss.

III. Design of Device

Fig. 6 shows a prototype of a walking assist device with bodyweight support system with which we aimed at providing the assist force vector that extends from the COP to the COG of its user. The device comprises a seat, rails, upper frames, middle joints, lower frames, bottom joints, and shoes. The seat and the right and left rails are connected with a joint located behind the lower back of the user. The rails are respectively linked to one of the upper frames in a slidable way with a roller. An actuator that has a weight is disposed at the top rear portion of the respective upper frames considering the ease of leg swinging motion. The rotational force of each actuator generates a torque at the middle joint which links the upper frame to the lower frame, through a transmission mechanism. The lower frames are respectively linked to one of the shoes with a spherical joint. The respective upper frames are equipped with a built-in electrical system, such as actuator and CPU, and a battery. The total weight of the device is approximately 6.5 kg. Details of the design of the device are as follows:

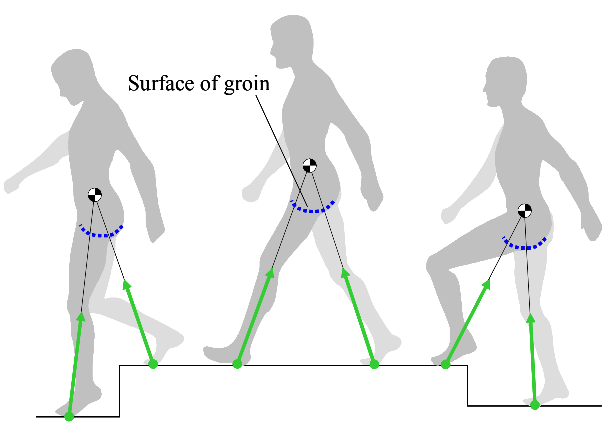

Fig. 7. Relationship between the assist force vector and the surface of groin.

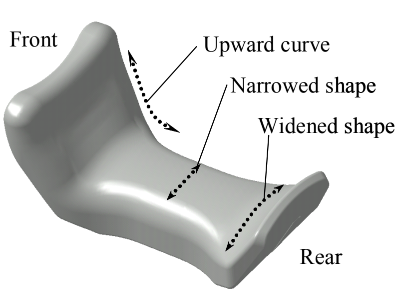

Fig. 8. Shape of seat.

A. Seat that Transmits the Assist Force to the User

As shown in fig. 7, the assist force vector that extends from the COP to the COG of the user passes through the surface of the user's groin region in various postures. For this reason, a seat serving as a member for providing the assist force vector to the user is disposed to receive the face of the user's groin.

Fig. 8 shows the shape of the seat. In order to transfer the assist force vector extending from the front to the groin region of the user in a stable manner, the front portion of the seat is shaped with an upward curve so as to follow the surface of the user's body. In addition, the center portion of the seat is narrowed in width to facilitate the anteroposterior movements of the user's legs. Furthermore, the rear portion of the seat is widened so that the assist force can be transferred to the user's hip-bone.

B. Layout of Axes with a Single-point Linkage

The expansion and contraction mechanism, which has a freely rotatable axis disposed on each of its ends, always generates a force in the direction of a line connecting the two axes on the two ends. This fundamental principle stays the same irrespective of the type of expansion and contraction mechanism. By utilizing this principle, a freely rotatable axis is disposed both at the starting point of and at the point located in the acting direction of the assist force vector, and the two axes are linked to each other with an expansion and contraction mechanism (Fig. 9). Of the two axes, the lower end axis is disposed near the COP and the upper end axis is disposed near the COG of the user. The upper end axis is configured to have two degrees of freedom, and consists of an axis for swinging the expansion and contraction mechanism anteroposterior and an axis for swinging the mechanism from side to side, both relative to the seat. In addition, the upper

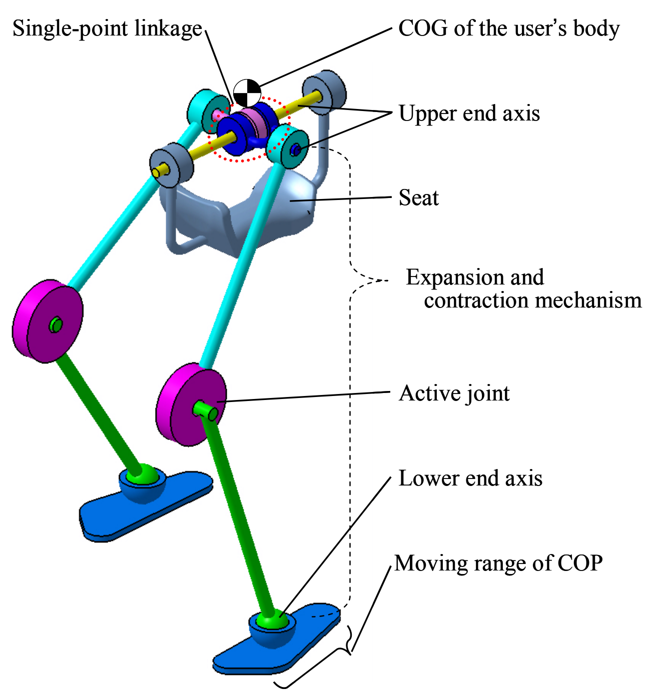

Fig. 9. Layout of rotation axis.

end axis of the expansion and contraction mechanism on the right and that of the mechanism on the left are linked at a single point near the COG of the user with a mechanism to be described later. The above-explained axis layout makes it possible to maintain the assist force vector to extend from the COP to the COG of the user in any postures by providing each leg with an actuator.

The axis layout with a single point linkage has an advantage also in compensating for the weight of an expansion and contraction mechanism in the swing phase. Suppose a layout in which the upper end axes of the right and left expansion and contraction mechanisms of the device are not linked at a single point like the one shown in fig. 10(a). In this case, the upper end axis of the expansion and contraction mechanism in the swing phase goes down due to the weight of the device. As a result, the weight of the expansion and contraction mechanism in the swing phase is not compensated, causing the load to be applied to the user's swinging leg. Meanwhile, with the layout linked at a single point, the weight of the expansion and contraction mechanism in the swing phase is supported by the expansion and contraction mechanism in the stance phase as shown in fig. 10(b). Thus, we achieved a layout that inhibits application of load to the user in which the weight of the entire device including the weight of the expansion and contraction mechanism in the swing phase is compensated in a stepping motion and other.

C. *Layout along the Inner Sides*

Since, as shown in fig. 11, the assist force vector from the COP to the COG of the user is approximately along the inner side of the right and left legs, the structural members of the device are disposed along the inner side of the user's legs. With this layout of the members along the inner side, the bending moment that acts on the structural members can be reduced, making it possible to reduce the rigidity and strength required for the member. As a result, the weight and the size of the device can also be reduced.

The layout on the inner side has various benefits beside this. Since there are no structural members on both right- and left-side of the user's body, natural swinging of arms is not harmed when the user is walking. In addition, the device does not easily contact another object when the user goes through a narrow space. Furthermore, since the mass of the device is near the center of the body axis of the user, the user hardly feels the inertia of the device as the user turns around.

D. *Realization of Freely Rotating Axis near the Center of Gravity of the User's Body*

In order to achieve both the axis layout with a single point linkage and the disposal along the inner side, a circular arc-shaped rail mechanism shown in fig. 12 is used. The mechanism comprises a circular arc-shaped rail with its center near the COG of the user and a roller for sliding the expansion and contraction mechanism along the rail.

There are a total of two circular arc-shaped rails with one on the right and the other on the left, and they form the axes for swinging the expansion and contraction mechanism anteroposterior. The circular arc-shaped rails on the right and on the left are linked to the seat with a joint in the rear, and an axis for swinging the expansion and contraction mechanism side to side is formed with this joint. Since the force transferred in the direction of the roller movement is small

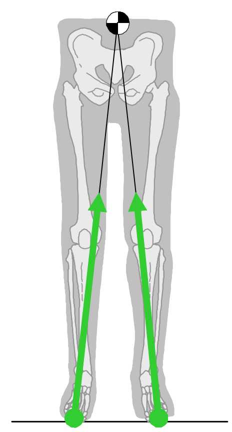

Fig. 11. Assist force vector that passes along the inner side of the legs.

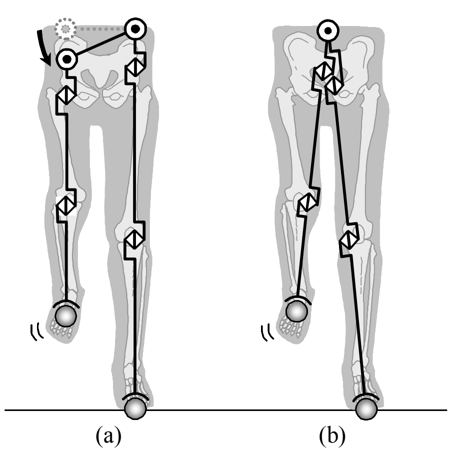

Fig. 10. System weight compensation in the swing phase.

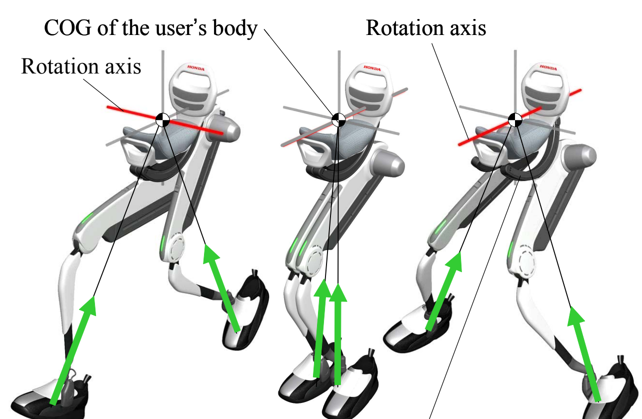

Fig. 12. Assist force vector that always stays in the direction of the COG of the user's body.

enough to be disregarded, the force vector transferred between the roller and the circular arc-shaped rail is always in the direction of the center of the circular arc.

With the above-mentioned mechanism, the upper end axis of the expansion and contraction mechanism can be disposed near the COG of the user.

E. Mechanism for Stabilization of Seat

In a configuration in which the rotational center of pitching and rolling of the seat in relation to the expansion and contraction mechanism is located at a point higher than the seat, the seat and the user are suspended from the upper end axis of the expansion and contraction mechanism as shown in fig. 13(a), which ensures a stable location relationship between the upper end axis and the user. On the contrary, when the rotational center of the seat is lower than the seat, the location relationship between the upper end axis and the user is unstable as shown in fig. 13(b), and this makes it easier for the seat to be displaced from the user. Accordingly, the vertical location of the rotational center of the seat relative to the seat is found to have a large effect on stability. Based on the above-mentioned examination, the characteristic of stable location relationship between the upper end axis and the user when the assist force is applied was accomplished by using a circular arc-shaped rail and others so as to dispose the rotational center of rolling and pitching of the seat higher than the seat.

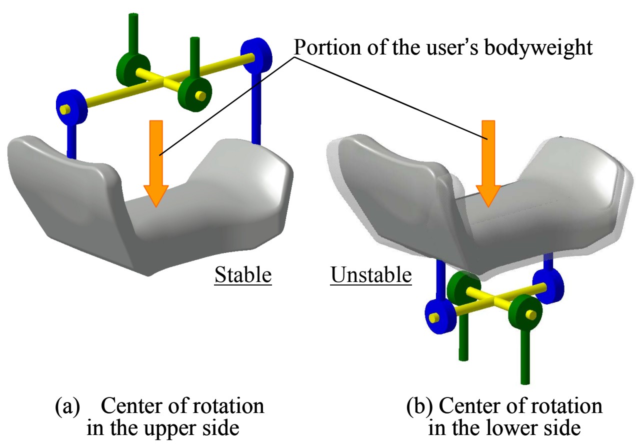

Fig. 13. Stability of Seat.

With the aforementioned shape of the seat and the location relationship between the upper end axis and the seat, the assist force can be provided to the user in a stable manner in various postures. As a result, the device does not have to be fastened to the user with a belt or the like, facilitating putting on and taking off of the device.

F. Flow of Force in Shoe Area

Since the COP of a person always stays within the projection plane of the person's foot on the floor, the assist force vector must be configured so as to pass within the projection plane of the foot. As shown in fig. 14, the point on the floor through which the assist force vector passes can be obtained by drawing a line connecting the upper end and the lower end of the expansion and contraction mechanism and finding an intersection point with the floor. Fig. 14 indicates that the lower end axis of the expansion and contraction mechanism should be disposed close not only to the longitudinal center of the shoe but also to the floor in order to limit the point on the floor through which the assist force vector passes to stay within the projection plane in all the conceivable postures. For this reason, the axis at the lower end of the expansion and contraction mechanism is disposed near the longitudinal and lateral center of the shoe immediately above the shoe.

Fig. 15 shows the flow of the assist force from the lower end axis of the expansion and contraction mechanism to the floor. The assist force passes through the lower end axis, inner frame, shoe sole, and floor in this order. The inner frame is shaped so as to avoid the leg and foot of the user, and transfers the force from the lower end axis to the floor without going through the foot of the user. Thus, the user does not feel any pressure from the assist force on the user's legs and feet. In addition, the foot force of the user that is reduced with the assist device and the assist force are added together and

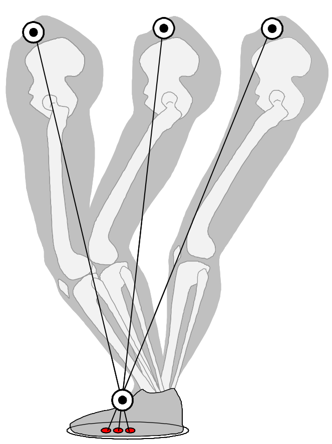

Fig. 14. Points on the floor through which the assist force vector passes.

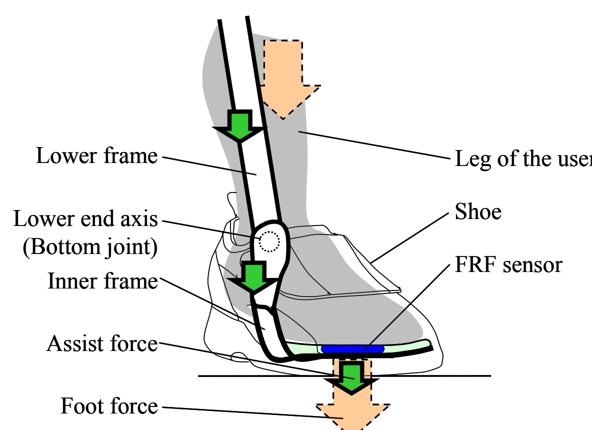

Fig. 15. Flow of the force from the lower end axis of the bend and stretch structure to the floor.

transferred to the ground through the shoe sole as shown in fig. 15. Accordingly, the tangential force and the normal force that are transferred to the ground through the shoe sole do not change even when the assist force is increased or decreased, making it possible to achieve movement such as walking with a feeling close to when the device is not used.

G. *Control*

The assist force vector includes two elements: direction and size of vector. Of these two, the direction of the assist vector is configured to always stay from the COP to the COG with the use of the aforementioned mechanism, regardless of the posture of the user. Thus, only the size of the assist force vector needs to be controlled.

It is important to determine the size of the assist force based on the FRF of the user in order to achieve the concept of the assist method, "reduction of the FRF of the user," as intended by the user. For this reason, we decided to measure the FRF of the user. In measuring the FRF of the user, a single-axis force sensor was used by disposing it immediately below the foot of the user to prevent interference with the assist force output by the device (Fig. 15).

Fig. 16 shows the processing procedures. First, a target assist force desired to be provided to the user is determined, and an overall target of the generating force is obtained by adding the weight of the device to this value. Then, the force is distributed to right and left according to the FRF measured with the right and left FRF sensor so as to obtain the target of the generating force of the respective expansion and contraction mechanisms (Fig. 17). The target of the generating force in the swing phase is determined to achieve a pull up force that compensates for the weight of the expansion and contraction mechanism in the swing phase. The generating force is feedback-controlled by a generating force sensor and a servo motor so that the respective distributed targets of the generating force are achieved.

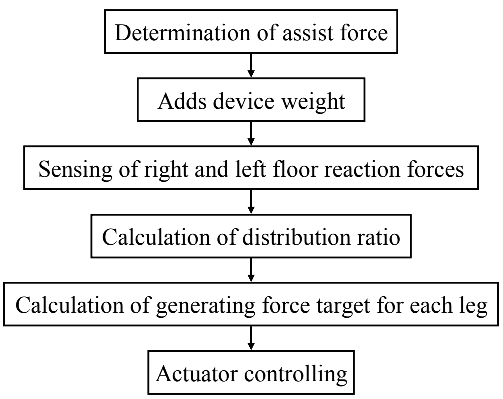

Fig. 16. Control flowchart.

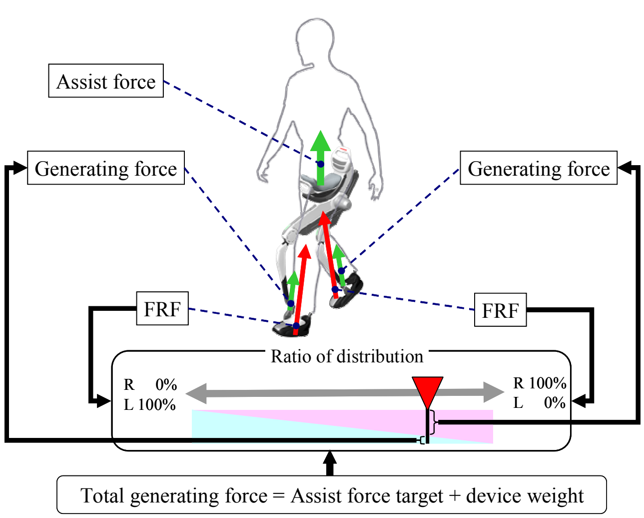

Fig. 17. Distribution method of the generating force.

The generating forces from the respective expansion and contraction mechanism on the right and left are added up due to the layout with a single point linkage so that the own weight of the device is supported and so that the remaining force serves as an assist force and is provided to the user through a seat.

IV. Testing with the Use of Device

A. *Change in the FRF of the User*

In order to confirm a reduction of the FRF of the user achieved by the assist device, the FRF was measured both with and without the use of the assist device. The measurements were taken on an able-bodied male subject while he was walking slowly on a level ground and with the assist force set at 100 N. The FRF was measured with a floor reaction force plate. The force generated by the expansion and contraction mechanism at the lower end axis was measured with a three-axis force sensor for verification having been incorporated between the lower frame and the shoe of the device. The FRF of the user was calculated by subtracting the measurement value of the three-axis force sensor from the measurement value of the floor reaction force plate. The spatial position and posture of the three-axis force sensor were measured with a 3D motion capture system.

Fig. 18 shows transition of the FRF of the user when using the assist device and when not using the assist device. It is confirmed that the FRF of the user decreases when the assist device is used.

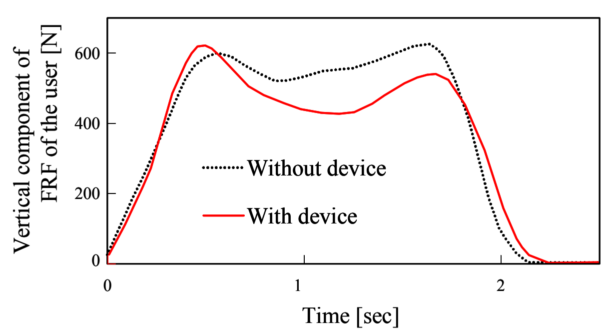

Fig. 18. Change in the floor reaction force of the user.

TABLE I
CHANGE IN MUSCLE ACTIVITIES

Symbol		Without assist [microV]	With assist [microV]	Reduction Rate [%]
Squatting exercise	Gastrocnemius muscle	21.3	17.7	17.1
	Vastus medialis muscle	79.3	64.5	18.6
Climbing stairs	Gastrocnemius muscle	76.7	52.6	31.4
	Vastus medialis muscle	69.9	64.6	7.5

TABLE II
CHANGE IN ENERGY CONSUMPTION

Symbol	Without assist [kcal/min]	With assist [kcal/min]	Reduction Rate [%]
Squatting exercise	5.9	5.3	11
Climbing stairs	3.6	3.3	7

B. Change in Local Load on the User's Body

Local load on the user's body was measured with and without the use of the assist device. The muscle activities were used as the indicator of the degree of the load. The muscle activities were measured with a surface EMG signal measuring device by measuring the myoelectric potential of the inner vastus medialis muscle, which is extensor muscle of knee joint, and of the gastrocnemius muscle, which is extensor muscle of ankle joint.

The measurements were taken on an able-bodied male subject while he did knee bend and climbed stairs. For the knee bend, movement from a standing position to a bent position with the knee angle of 90 deg was repeated at the cycle of 22.5 sec. Climbing stairs was achieved using an exercise machine with the step height of 17 cm at the rate of 60 steps/min.

Table 1 shows the measurement results. With the reduction of the muscle activities of two muscles, the reduction of the load on the leg muscles targeted to assist was confirmed.

C. Change in the Load on the User's Entire Body

The load on the entire body of the user was measured with and without the use of the assist device. The total body energy consumption was used as the indicator of the degree of this load. The energy consumption was measured by measuring the oxygen consumption of breathing with a portable system for pulmonary gas exchange analysis, and the total body energy consumption was calculated. In addition, the energy consumption in a sitting position was measured to be used as the resting metabolic energy, and the increase in the energy consumption caused by the movement was calculated and compared by subtracting the resting metabolic energy from the total body energy consumption while in motion.

The testing conditions used were the same as the measurement of muscle activities.

Table 2 shows the measurement results. By comparing the increase in energy consumption due to the movement, a reduction of load on the entire body was confirmed. In addition, these results are consistent with the subjective evaluation of the subject that the load on the body was mitigated by using the device.

Based on the results of the above-mentioned three tests, it is expected that many of the in vivo forces of legs, such as bone-on-bone force generated at three joints: hip joints, knee joints, and ankles and muscular force that generates joint moments, will be reduced.

V. CONCLUSION

In this research, based on biomedical engineering analysis results, the assist method concept of “reduction of floor reaction force of the user” was established. We then defined attainment of the assist force vector that extends from the COP of the FRF to the COG of the user as our design guideline. To realize this we devised an original system comprised of an axis layout in which expansion and contraction mechanisms are connected to a seat at a single point, and a device layout which is disposed along the inner side of the user's legs, with a mechanism that utilizes a circular arc-shaped rail. Thanks to this mechanism, constantly maintaining the assist vector from the users COP to their COG with just two actuators has been made possible. In addition, the device does not need to be fastened to the user, providing for ease of putting on and taking off of the device. Testing of the device has confirmed a reduction in floor reaction force of the user, muscle activity and total body energy consumption. From these results, we can expect a reduction in the majority of in vivo forces of the legs, such as muscular force and bone-on-bone force.

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