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*Research Article*

# Bioinspired Knee Joint for a Power-Assist Suit

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Movement of the knee joint of a human includes rolling and sliding. There also exist rotations in the frontal and horizontal planes. To assist the standing movement of a human, we developed a bioinspired knee joint and torque adjustment mechanism. We evaluated the motion, torque characteristics, and stress of the developed mechanism. This joint allows deep flexion of the knee with small resistance for both the user and the device. In addition, in spite of 33% error in deep flexion, the measured torque over less than 120 degrees fits the designed torque curve. We conducted evaluation tests for a human subject. The electromyogram (EMG) of musculus rectus femoris was measured during standing with or without the assistance. The result shows 30% and 63% reduction with the assistance from 100-degree and 80-degree knee angles, respectively. In addition, the proposed device reduced up to 80% of stress in the frontal plane during standing.

## 1. Introduction

The aging of society is rapidly progressing all over the world. According to the report of Department of Economic and Social Affairs, USA [1], the number of persons aged 60 years or more was 841 million in 2013 worldwide, and the number is expected to be 2 billion in 2050. Aging affects the overall physical functions. In particular, arthralgia (joint pain) is one of the most severe factors that prevent active lives of the aged. In Japan, the number of patients suffering from knee osteoarthritis is approximately 7 million, and the proportion continues to increase [2]. Knee joints are very important for many activities of daily living, for example, standing, walking, and climbing stairs; more than 0.5 Nm per body weight in torque is generated in the knee joint during walking [3]. The torque required in the standing action exceeds this value. The pain of knee osteoarthritis causes the losses of the range of motions, motivations of activities, and range of their activities in life.

The power-assist technology was originally developed as supporting technology for military persons [4]. However, the technology is now expected to promote the active life of the aged. Commercially available power-assist wearable devices already exist, for example, MuscleSuits (INNOPHYS Co., Ltd.) [5] and HAL (CYBERDYNE Inc.) [6]; these devices are used to reduce the loads of standing or the carrying burdens of persons. Such devices use several types of actuators to generate the assistive torque and can reduce the force that directly reacts on the user's body. However, due to the limitation of the mechanism, almost all of the devices do not support motions that require deep flexion of the knee. In addition, some devices that utilized electromagnetic motors tend to be quite heavy.

In recent years, pneumatic artificial muscles have been the focus for use in power-assist system because of their high force/weight ratio [7]. A pneumatic actuator contracts with the injection of air. The core element of a pneumatic actuator is a flexible membrane closed at both ends; this rubber bag is wrapped with the sleeves of hard plastic fibers. The rubber bag inflates with the injection of air, but the deformation toward radial direction is limited by the hard sleeves. As a result, the contraction toward axial direction occurs with the injection of air [8]. The above-mentioned MuscleSuits also utilize pneumatic artificial muscles.

The use of the pneumatic muscles in a power-assist system depends on the application and is mainly categorized into two types. In some devices, the pneumatic muscles are directly attached onto the surfaces of the human body, similar to clothes, for example, in the front and back of the target joint to assist its flexion and extension [9]. The other devices are combinations of the pneumatic actuators and mechanical joints, with a single-axis that is located in the outer side of the human joint [10]. The former types have an advantage that they do not require accurate alignment to the axis of the human joint. However, the inflated actuators give users the feeling of pressure. Alternatively, although the latter types of devices do not give such feelings of pressure, they must be accurately adjusted to the human joints. In addition, the above studies have not addressed the deep flexion of knee, although there are many such motions in daily life.

The complicated motion of knee joint results from the restraints of motion with the related ligaments and tendons. For example, the extension of the knee includes the roll-back motion, which is a combination of rotation and sliding [11]. At the same time, the rotational axis moves, depending on the knee angle. It is difficult to adequately fit the natural motion of human knee joint only using a mechanical joint with a single-axis. Some studies that utilize a bioinspired knee mechanism can be found in the field of humanoids [12, 13] and prosthetic knees [14]. Terada et al. [15] developed a knee-motion-assist mechanism for a wearable robot with a noncircular gear and grooved cams. In this paper, we also focused on the bioinspired knee joint mechanism but utilized a different mechanism.

Furthermore, during flexion of the knee joint, the femur normally has a 5–10 degrees' abduction against the tibia, reaching up to 30 degrees during extension in some cases [16]. From the 30-degree flexion to the terminal extension, the femur rotates 5–10 degrees in inner rotation. This motion is important to lock the knee motion in the terminal extension of the knee and prevent hyperextension [11]. There are few studies of the mechanical knee joint that considered the motions in the frontal and the horizontal planes.

A new mechanical knee joint that solves the above-mentioned problems is necessary for the pneumatic actuator-based power-assist suit to improve the usefulness and comfort of the suit. In this study, we propose a new bioinspired mechanism for mechanical knee joint and flexible elements that can reduce the stress from the motion with the misaligned joint.

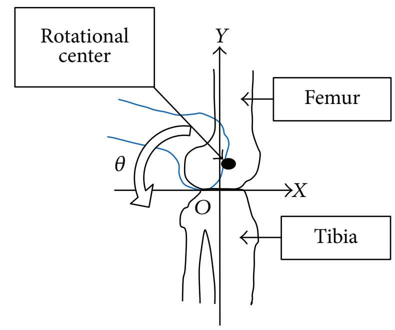


Figure 1: Coordination system of the tibia and femur [17].

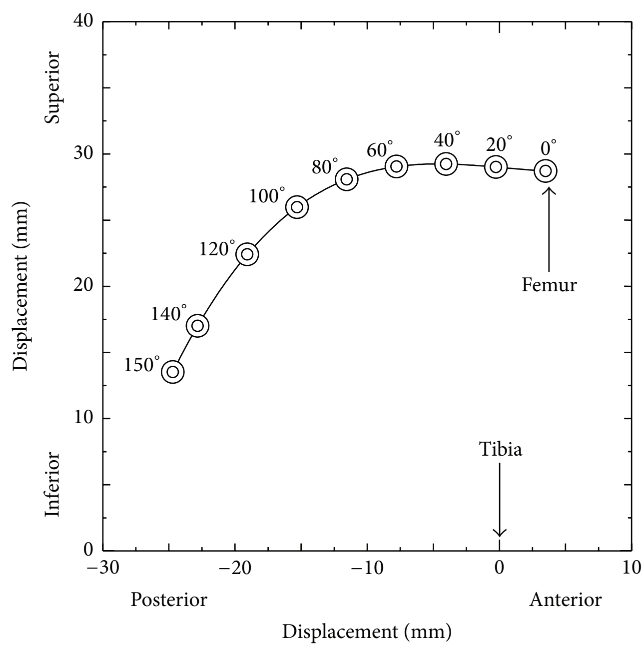


Figure 2: Trajectory of the center of the femur in the sagittal plane for a Japanese male (original data are obtained from [16]).

## 2. Target Setting

*2.1. Trajectory of the Axis of the Knee Joint.* The motion of the axis of knee joints is a combination of rotation and sliding [11]. Therefore, the rotational axis is not fixed in a point. According to the definition of the coordination system in [17] (Figure 1), the origin of the femur's rotation in the sagittal plane with respect to the tibia's origin moves as shown in Figure 2 during the standing motion [16]. The vertical axis of this figure shows the displacement of the superior-inferior motion of the rotational axis (positive values denote the superior positions). The horizontal axis shows the anteroposterior motion (positive values denote anterior positions). The number indicated at each plot denotes the knee angles, and the maximum extension and flexion are 0 and 150 degrees, respectively.

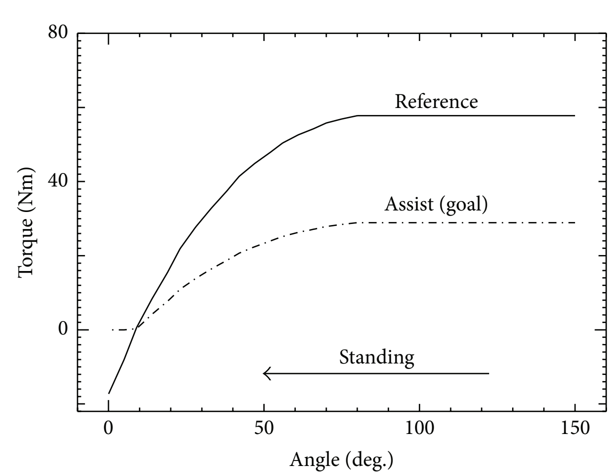


Figure 3: Natural standing torque and assisted torque around the knee joint (original data of solid line are obtained from [18]).

*2.2. Goal of Assistive Torque.* The solid line in Figure 3 shows the average torque profile during standing for a Japanese male

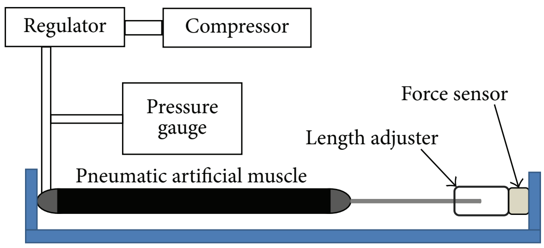


Figure 4: Measurement method for determining the force characteristics of the pneumatic actuator.

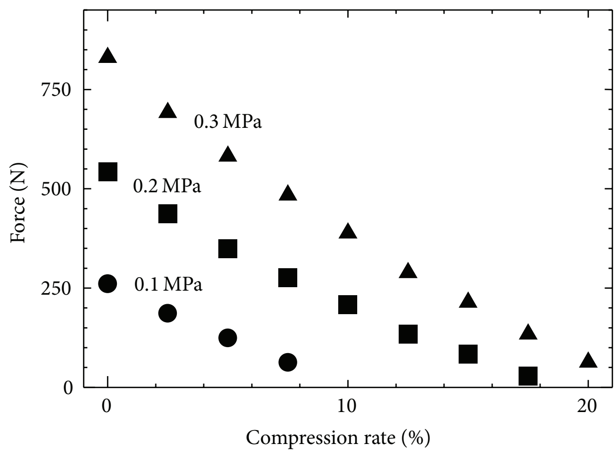


Figure 5: Compression rate versus compression force of the pneumatic actuator.

(40 years old, weight: 65.3 kg, and height: 171 cm) obtained from [18]. The maximum torque is 57.8 Nm at the moment of standing. The maximum torque remains flat within 80–150 degrees' knee angle. This line is defined as a reference in this paper. In this study, we set a half of the reference as the goal of the assistive torque (Figure 5). For simplification of the structure, only extension of the knee was assisted with the device. As a result, the negative torque in the reference was translated to zero in the goal.

*2.3. Force Characteristics of the Pneumatic Actuator.* In this study, we used AIR-MUSCLE (Kanda Tsushin Kogyo Co., Ltd., A300B20C20D, nominal length: 800 mm, and max. pressure: 0.6 MPa) as a power source. In addition, the maximum air pressure was defined as 0.3 MPa to allow for use of a compact compressor. This actuator is a McKibben-type artificial muscle, which can generate a compression force with air input. The force characteristics of the actuator were measured with the setup shown in Figure 4 because the actuator has individual difference in the force characteristics.

Figure 5 shows the experimental results of the static force characteristics of this actuator. The horizontal axis shows the contraction ratio with respect to the nominal length. The vertical axis shows the compression force with each value of inner pressure and length. This actuator generates more than 750 N with 0.3 MPa at the nominal length. The force decreases with the compression rate, and it becomes zero at approximately 20% compression.

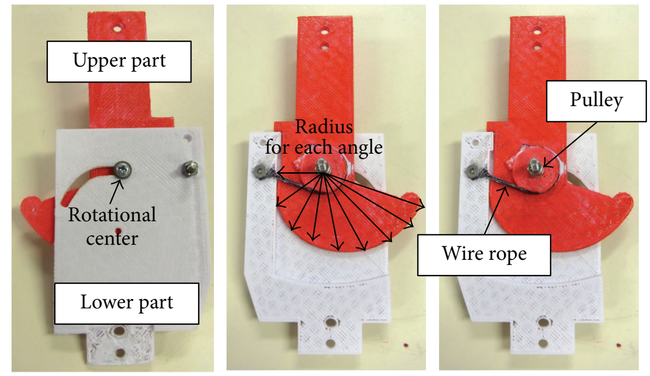


Figure 6: Knee mechanism of the device.

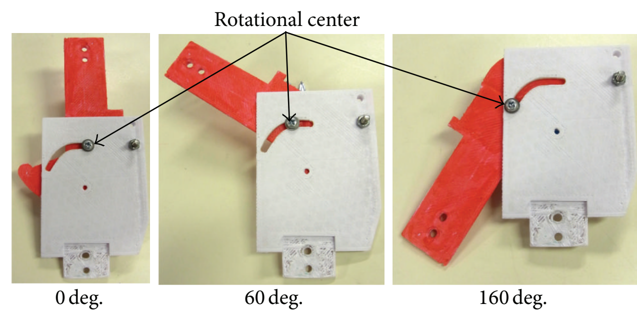


Figure 7: Motion of the knee mechanism.

## 3. Materials

*3.1. Bioinspired Knee Joint.* We developed a prototype of the joint mechanism by referring to the trajectory of the rotational center [19], as shown in Figure 2. Figure 6 shows the structure of the prototype, which consists of an upper part, a lower part, a wire rope, and a pulley. The upper part and the lower part move relatively with sliding and rolling. The rotational center of the upper part has a shaft that can move in a curved hole. A circular pulley is fixed with the upper part and is joined with a wire rope, one edge of which is fixed on a point of the lower part. The anteroposterior motion of the rotational axis depends on the knee angle because the shape of the sliding surface is noncircular (see the middle figure of Figure 6). The wire-pulley mechanism is also used; this mechanism defines the length between the rotational center and the fixed point (see the right figure of Figure 6). The actual motion of the mechanism is shown in Figure 7. This joint allows for deep flexion of knee with small resistance for both the user and the device (Figure 8).

*3.2. Lever Arm for the Pneumatic Actuator.* Figure 9 shows the basic structure of the proposed power-assist suit. The joint mechanism mentioned above is activated with a McKibben-type artificial muscle. For simplification of the mechanism, there are no reduction gears. In this structure, 40 mm of

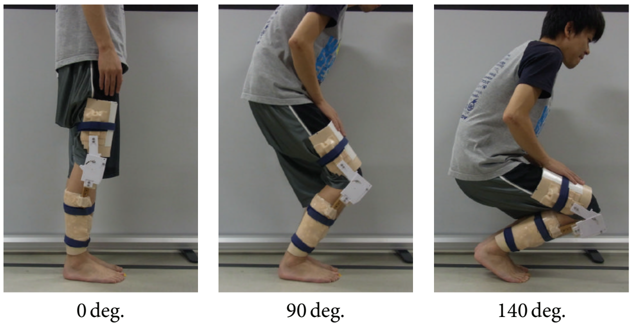


FIGURE 8: Squat with the knee mechanism.

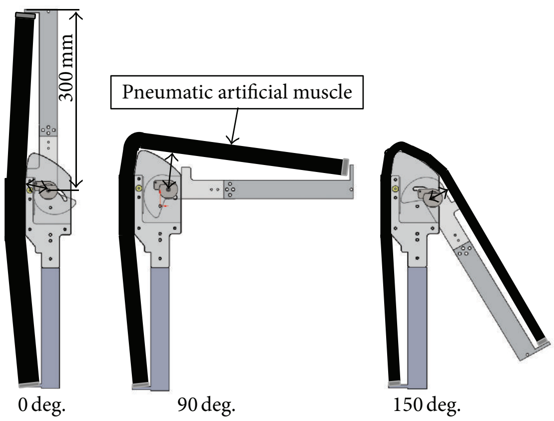


FIGURE 9: Structure and motion of the device.

compression is required to achieve 150 degrees of flexion. Thus, we selected the AIR-MUSCLE as the actuator.

To meet the goal of Figure 3 with the force characteristics of Figure 5, the adjustment of the lever arm is required. Hence, we designed the shape of the outer surface as the patella of the device (Figure 9). As shown in Figure 5, the actuator can generate a large force over a low range of the contraction rate. The force gradually decreases with the contraction and reaches zero at a specific contraction rate. However, the required torque remains constant within knee angles 80–150 degrees (Figure 3). To meet this requirement, the shape was designed to have a large lever arm at 90-degree flexion of the knee.

To simulate the torque generation with this structure, we assumed two friction areas, as shown in Figure 10. We attached fluororesin tapes onto the surface of the lower part, and then the friction coefficient between the upper and lower parts was defined as 0.3. The friction coefficient between the slit (aluminum) and rotational shaft (stainless steel) was defined as 0.4. The calculation result is shown in Figure 11. The designed curve almost achieves the goal.

*3.3. Hinge Mechanism for Flexibility.* To enhance the compatibility of the wearable device with the human body, the mechanism should have suitable flexibility. However, rigidity is required for the control of direction. To satisfy these conflicting requirements, we use a pair of hinge joints,

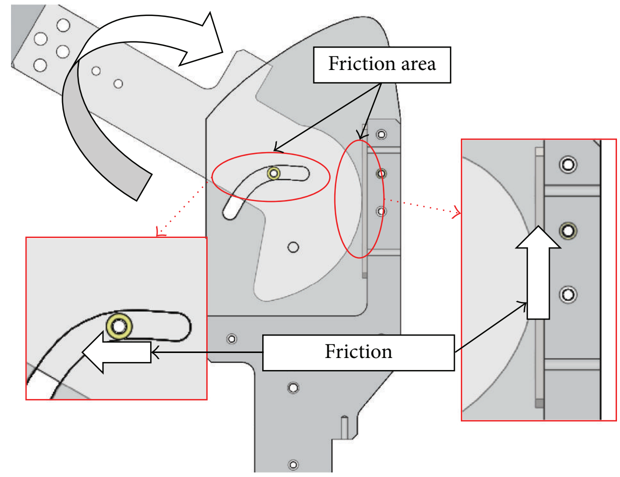


FIGURE 10: Motion of the torque adjustment mechanism.

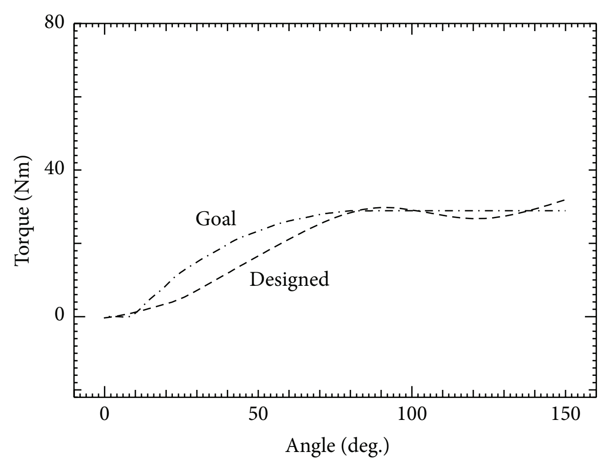


FIGURE 11: Designed torque.

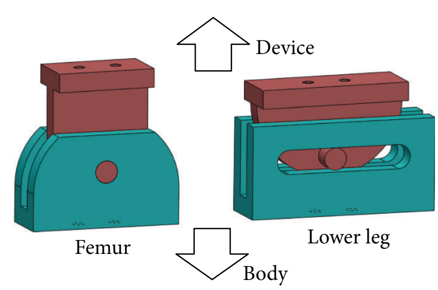


FIGURE 12: Hinge parts for flexibility.

as shown in Figure 12, between the human body and the machine. One of the joints has a simple rotational joint and is attached to the femur (upper) part of the device. The other has a pin and a slot to allow for rotation and sliding and is attached to the lower part.

The hinge parts were installed between the knee mechanism and the plastic cuffs. The plastic cuffs were tightened on the user's thigh and shin by using wide and stretchable supporters. Figure 13 shows the motion of the knee


FIGURE 13: Installation of hinge parts and their motion.

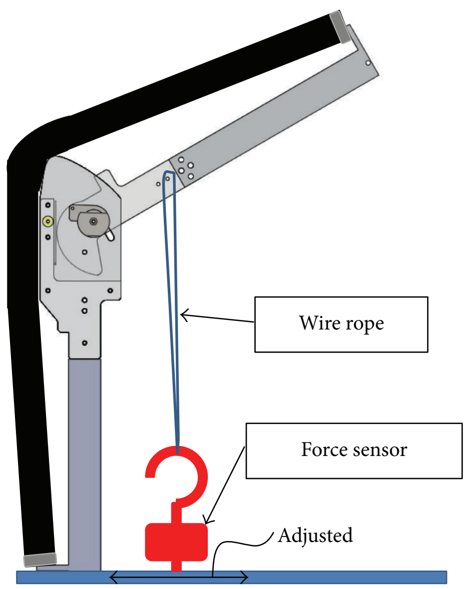


FIGURE 14: Setup for the torque measurement.

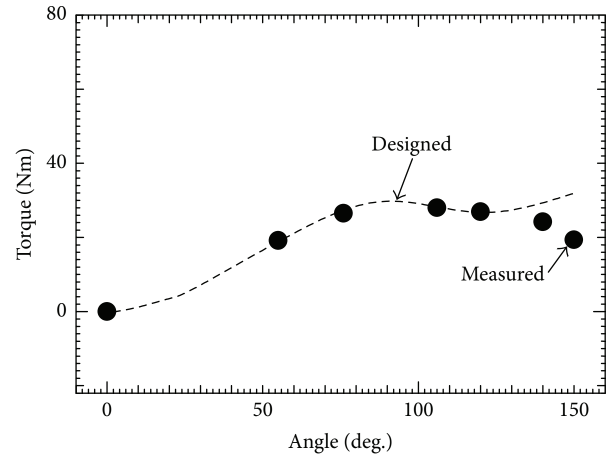


FIGURE 15: Comparison of the measured and design torque.

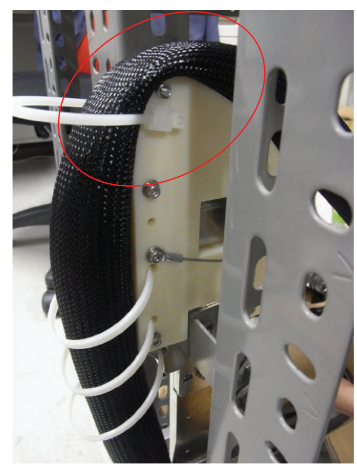

FIGURE 16: Flattened tube at deep flexion of the knee.

mechanism with the hinge parts. The upper and lower cuffs fit on the legs during abduction and adduction of the knee. The smooth flexion and extension of knee were achieved.

## 4. Evaluation of the Torque Output

*4.1. Method.* To evaluate the torque output, we prepared an experimental setup shown in Figure 14. The basic structure of the assist device was fixed on a rigid base. A wire rope was hooked at a specific point on the device with one end. The other end was hooked with a force sensor and the force sensor was manually fixed onto the base such that the wire rope is perpendicular to the base. The static torque measurement was conducted on each knee angle (7 points between 0 and 150 degrees). A constant air pressure of 0.3 MPa was applied to the actuator for each angle of knee (0–150 degrees). The torque around the axis of the knee joint was calculated with the torque and lever arm.

*4.2. Result.* Figure 15 shows the experimental results (plots) and the design torque curve. The measured torque for angles less than 120 degrees fit the designed curve. The friction model mentioned in the Section 3.2 adequately estimated the static assistive torque of the device for angles less than 120 degrees. However, an error exists between the measured and design torque in deep flexion for angles of over 140 degrees. At the maximum flexion, the measured torque was 33% less than the design torque.

*4.3. Discussion.* To explain the cause of the error in deep flexion, Figure 16 shows the condition of the pneumatic actuator at a knee angle of 150 degrees. As shown in the picture, the main tube of the pneumatic actuator is flattened in the deep flexion; this separates compressible volume in this state. This flattened state possibly corresponds to a 33% loss from the maximum output of the actuator. This problem can be solved by using a longer actuator; however, the required margin of air volume results in the delayed response at the initial moment of standing motion. We will attempt to solve this problem in the future.

## 5. Evaluation of Assistance

*5.1. Method.* Despite the shortness of output in the deep flexion condition, we conducted evaluation tests using a human

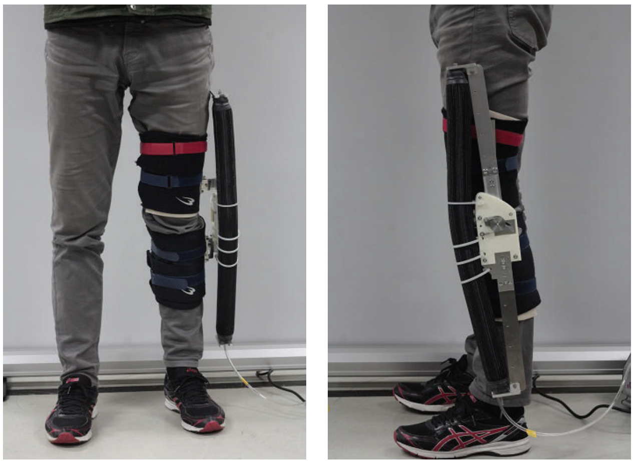

Figure 17: Device attached on the body.

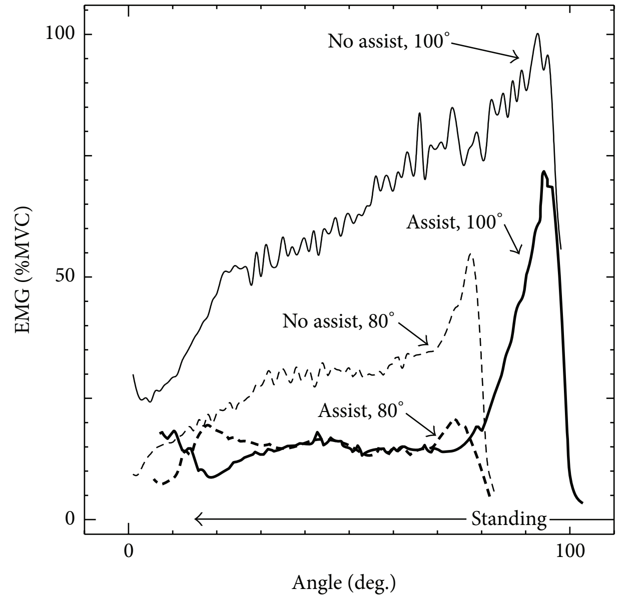


Figure 18: Muscle activation with and without assistance.

subject. A male subject (22 years old, weight: 65 kg, and height: 177 cm) without any physical or mental disabilities was recruited. He wore the assistive device on his left leg (Figure 17). The plastic cuffs were tightened on his thigh and shin by using wide and stretchable supporters. He sat on seats of two different heights (210 mm and 420 mm) at the initial state before standing. A wireless electromyogram (EMG) sensor (Logical Product, LP-WS1221) was attached on the musculus rectus femoris of his left thigh, and the EMG was measured during standing. A Flexible Goniometer System (Biometrics Ltd., K800 and SG110) was used to measure the knee angle. We just opened a valve for the air tube of the actuator at the instant of standing with assistance from the device. The air pressure was maintained at 0.3 MPa during assistance. The start switch was pushed by the subject. The experiment was repeated five times for each condition.

*5.2. Result.* The knee angles at the initial state were 100 degrees for the 210 mm chair and 80 degrees for the 420 mm chair. Figure 18 shows the experimental results for the four conditions described as follows:

(1) EMG without assistance from 100 degrees.
(2) EMG with assistance from 100 degrees.
(3) EMG without assistance from 80 degrees.
(4) EMG with assistance from 80 degrees.

The horizontal axis shows the knee angle; the angle starts from 100 degrees for conditions (1) and (2) and 80 degrees for conditions (3) and (4).

The root mean square (RMS) with 100 ms interval was calculated from the raw EMG signals. The results of repeating the test five times are averaged for the same angles. The maximum for condition (1) was defined as the maximum voluntary contraction (MVC).

*5.3. Discussion.* The maximum outputs appeared just after standing, followed by a gradual decrease. This trend is consistent with the characteristics of knee torque during standing (Figure 3). The peak EMGs with the assistance show 30% and 63% reduction from that without the assistance for 100 degrees' standing and 80 degrees' standing, respectively. After the peaks, the EMGs with the assistance were significantly smaller than those without the assistance. From these results, the developed device was found to successfully help standing only with the user's effort at the initial moment of standing. Although the designed assistive torque is just half of the reference torque (see Figures 3 and 15), the device significantly reduces the muscle activation without the initial period of standing. The reason for this result remains unclear; however, there is a possibility that the assist device also helps to reduce the muscle activation for maintaining posture. In this study, we conducted the experiments for a healthy subject. However, the final goal of this device is to assist the standing of disabled people. To achieve this goal, we must establish the biofeedback control using the EMG signals. One of the merits of this device is the simplicity of the control. A feedback control system was not used in the proposed device. We only input a constant pressure of air in the pneumatic actuator because the output characteristics are mechanically adjusted for the standing motion. This merit is helpful for developing a rehabilitative assistance suit for standing.

## 6. Evaluation of Stress

*6.1. Method.* Misalignments of assist suits generate unexpected stress on users' joints and skins. In order to reduce such stress, we proposed the bioinspired knee joint and flexible hinge parts. In this section, we measured the stress applied to the device in three conditions:

(1) A single-axis knee joint (its rotational center is fixed at the position of knee extension) without the flexible parts.

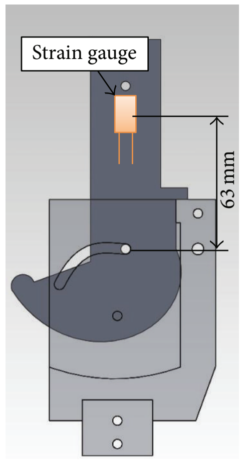


Figure 19: Position of strain gauge.

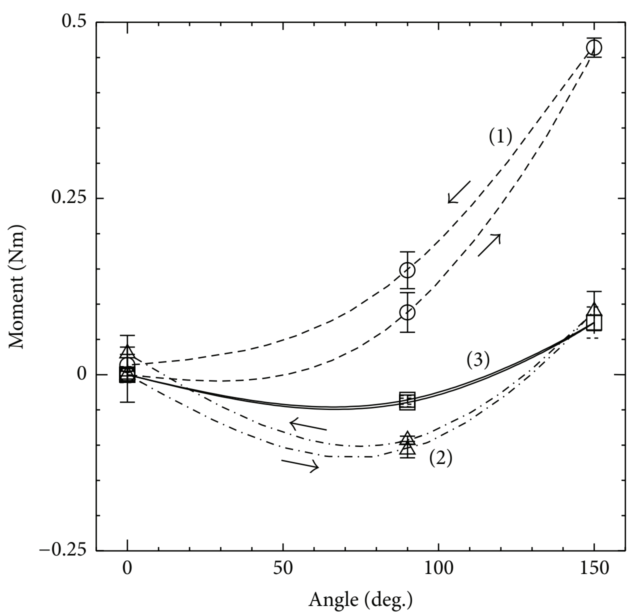


Figure 20: Bending moment on knee joint.

(2) The single-axis knee joint with the flexible parts.

(3) The bioinspired knee joint with the flexible parts.

A strain gauge was attached on the upper part of the knee joint (Figure 19). A healthy subject wore the knee mechanism and flexed (0–150 degrees) and extended (150–0 degrees) his knee joints 5 times without the assistance. The bending moment in the frontal plane applied to the frame of the device was calculated from the measured stress.

*6.2. Result.* The experimental results are shown in Figure 20. The vertical axis is the bending moment on the upper part, and the horizontal axis is the angle of knee joint. The negative direction of the bending moment represents the deformation away from the user. Each mark denotes the average value in each condition. The error bars denote the standard deviations.

*6.3. Discussion.* The bending moments at the maximum flexion are 0.46 Nm, 0.09 Nm, and 0.07 Nm for conditions (1), (2), and (3), respectively. But, for condition (2), the maximum absolute value reached 0.11 Nm in negative direction. During flexion of the knee joint, the femur normally has a 5–10 degrees' abduction against the tibia [11]. This motion generates undesired stress on both users and assist suits. The flexible parts reduced the stress 80% at the maximum flexion. In addition, the maximum variations are 0.20 Nm and 0.11 Nm, for conditions (2) and (3), respectively. The bioinspired knee joint reduced 45% of the variation of stress during standing.

## 7. Conclusions

In this study, we developed a new knee joint mechanism that includes a torque adjustment mechanism. We evaluated the motion, torque characteristics, and stress of the device. The combination of the sliding mechanism and wire-pulley mechanism fitted the subject's flexion-extension motions. The result of the EMG measurement during standing shows 30% and 63% reduction with the assistance of motion over 100-degree and 80-degree knee angles, respectively. In addition, the proposed device reduced up to 80% of stress in the frontal plane during standing.

## Competing Interests

The authors declare that they have no competing interests.

## Acknowledgments

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