Patient Simulator Using Wearable Robot: Representation of Invariant Sitting-down and Standing-up Motions of Patients with Knee-OA

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Abstract—It is ethically problematic to engage people with impaired motor functions in experiments, wherein they may face severe pain and unacceptable risk of injuries. We developed a wearable robot for healthy adults and its control algorithm to simulate the behaviors of patients with knee osteoarthritis. We focused on invariant motions, which are typically presented for avoiding knee pain. In order to simulate patient movements by a healthy person wearing a robot, we formulated a model motion that represented invariant patient motions. We then determined the output torque of the exoskeletal knee robot such that the wearer's motion followed the model motion. The effectiveness of the method was testified for standing-up and sitting-down motions, and some characteristic impaired motions such as body inclination to the healthy side and imbalanced right and left knee angles were manifested by the simulator.

I. INTRODUCTION

It is desired to establish a society in which people with impaired motor function can enjoy the same quality of life as enjoyed by healthy people. The reasonable and accurate assessment of the burdens incurred by impairment in patients' daily lives and the physical accessibility of public facilities contribute to the improvement of patients' quality of life. However, it is ethically unacceptable to engage impaired people in experiments to assess their burdens during certain activities especially when the experiment may involve severe pain or critical risks such as fall. To solve this problem, we proposed a novel method whereby a healthy person simulates the movements of an impaired person and participates in experiments.

Several research groups have attempted to simulate the impaired by the healthy. Pacala et al. reported a workshop for medical students to experience the inconvenience felt by elderly people needing caregiving [1]. They wore, for example, arm-hanging implements to imitate a symptomatic frozen shoulder. Wood held a similar workshop for participants to experience chronic diseases of the aged, using bandages and simulating the limited movements of knee joints caused by arthritis. She reported the educational effects of the study on the participants [2]. Positive effects of such experiences by wearing obstructive items have been also recognized by other researchers [3], [4]. Ullauri et al. employed a taping technique to imitate the limited ankle movements caused by the weakened muscles of the aged [5]. Aging simulators have been sold by some companies, with which the wearers are meant to experience motor, visual, and auditory deficiencies by

equipping their own bodies with weights, joint braces, goggles, etc. However, these methods only allow us to imitate passively obstructive factors. Recently, Huang et al. developed a patient simulator with actuated shoulder joints for training of patient transfer between a wheel chair and a bed [6]. However, they focused on the case of patient transfer and did not simulate a patient walking in the environment.

In contrast, Ishikawa et al. proposed a patient simulation method using exoskeletal robots for aiding in the training of the manual examination technique for physical therapy [7], [8]. Their robots were computer-controlled such that the wearer's joint resistances were similar to those of the impaired. In the study, Ishikawa et al. simulated several symptoms remaining a sufficient human likelihood unlike other robotic patient simulators. However, the wearers were supposed to relax and not move actively, as the purpose of the study was limited to simulating patients being passively examined or treated by clinicians.

The objective of the present study was the simulation of invariant and typical motions of people suffering from impaired knee motor function by using an exoskeletal knee robot. Such attempt is novel compared with the earlier patient simulator by passive obstructs [1]–[5]. We formulated model motions of knee osteoarthritis (OA) according to the characteristic abnormal movements of the impaired. We then determined the control rule for the wearable robot such that the wearer's motions resembled a model motion. We experimentally tested the developed system for sitting-down and standing-up motions. As a result, substantial similarities were observed between the model and simulated movements, indicating the effectiveness of our framework.

II. KNEE OSTEOARTHRITIS

We simulate the motions of knee-OA patients because of the significant effects of this impairment on patients' daily lives and popularity. In one report, the number of knee-OA patients in Japan with noticeable symptoms was estimated to be approximately ten million [9].

Knee-OA develops with aging and is typically accompanied by damage to the meniscus, cartilages, and bone spurs [10]. Patients suffer from symptoms such as pain during motion, a limited motion range, inflammation, and joint deformation [10], [11]. Patients with moderate knee-OA experience slight pain that hardly obstructs motions such as stair walking.
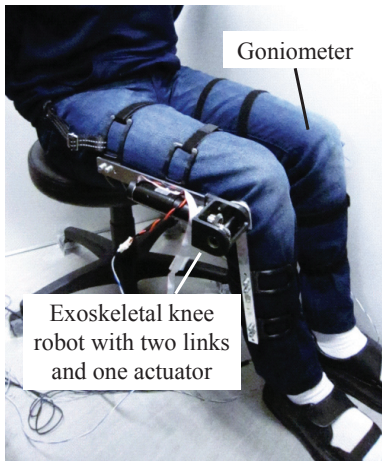
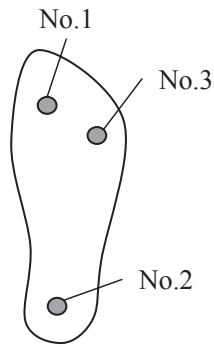Fig. 1: Exoskeletal knee robot worn by healthy human (right leg). Goniometer (left leg, hidden).

Fig. 2: Locations of force sensors (impaired side)

TABLE I: Characteristic abnormal sitting-down and standing-up motions of knee-OA patients

Slow movements due to the fear of pain.
Body inclination to the healthy side to avoid overloading on the impaired side. Typically, 70 and 60% of the weight is loaded on the healthy side during sitting and standing motions, respectively.
Patients stand up from the chair with their hands on their femoral thighs.
Fall to the seat at the last phase of the sitting-down motion.
Range of the joint motion is limited. Full knee extension and deep flexion are avoided.

Intermediate patients suffer from pain in moving their knee joints, especially with a large load applied to the joints. These patients' motions differ from those of healthy persons, as the patients tend to avoid motions causing pain. Severe patients undergo difficulties in their daily activities because of frequent and significant agony. In this study, we addressed the imitations of intermediate knee-OA patients. In particular, the following conditions were considered. Knee OA is developed at their right sides. The locomotive motions of patients are apparently distinct from healthy ones. Finally, they do not rely on supportive instruments such as a cane or hand rail.

III. EXPERIMENTAL EQUIPMENT

As shown in Fig. 1, a wearable exoskeletal knee robot was worn by a healthy adult on the right leg using two plastic braces for the femoral and lower thigh, respectively. A DC motor (RE-35, Maxon Motor, Netherlands, 90 W, continuous maximal torque 97.2 mN·m) with a reduction gear (GP32HP, Maxon Motor, 1/86) and an encoder (MR Type L, Maxon Motor, 1024 ppr) were installed on the knee joint. The motor was driven by a current driver (4-Q-DC ADS50/5, Maxon Motor) through a computer control at 5 kHz. On the left knee, a potentiometer-based goniometer was attached using Velcro tapes.

We measured the interaction forces between the right foot and the ground. Three three-axial force sensors (USL08-H18-1kN-AP, Tec Gihan, Japan) were embedded into a shoe sole at the locations shown in Fig. 2. The sensors were placed on the thenar and antithenar eminences and the heel. The sensor heads were appropriately leveled, such that they covered the load paths between the human and the shoe sole.

IV. INVARIANT MOTIONS OF KNEE-OA PATIENTS

Our final goal is to extract invariant motions from multiple actual knee-OA patients and stimulate those motions using a healthy adult with a wearable robot. However, to a greater or lesser extent, experiments measuring patients' motions impose physical burdens on the patients. Therefore, as a preliminary

step before experimentation on actual patients, we constructed model motions based on typical impaired motions performed by a healthy person. Even such a problem setting allowed us to testify whether our method, which is described later, can simulate specific abnormal motions that are different from the motions of healthy persons.

A. List of invariant motions

In this research, we selected sitting-down and standing-up motions as the target motions to be simulated. These motions occur often in daily lives and cause pain to patients. The model motions were determined by the following two steps. First, we listed the typical motions of knee-OA patients according to the literature [12] and opinions of a physical therapist (N.Y., one of the authors of this article). Second, a healthy adult imitated the typical motions of the impaired, under the supervision of N.Y.

Table I lists the invariant abnormal motions of the impaired while sitting-down on and standing-up from a chair. For example, the patients typically incline their upper bodies to the healthy side to avoid overloading on the impaired side.

B. Invariant motion of simulating target

According to the list of invariant motions and supervision by N.Y., a healthy person imitated the typical sitting-down and standing-up motions of knee-OA patients. To compare the impaired motions with the healthy motions, we also measured the healthy person's normal sitting-down and standing-up motions.

First, we checked whether the apparent features typical to the patients were exhibited by the performer. Fig. 3 shows the apparent movements of the model motions. Compared with the symmetric right and left knee changes for the healthy motion, the model motions exhibited an asymmetric body inclination to the healthy side. Furthermore we observed slower movements and falls to the seat in the last phase of the sitting-down motion.

Figs. 4 and 5 show the typical time courses of the knee angles and foot loads for the healthy (normal) and imitated model motions, respectively. In the case of the normal motions, both knee angles matched. In contrast, in the case of the model motions, they did not: the knee of the impaired side tended to take more extending positions than that of the healthy side. From the figure, we observed characteristics such as slow movements and the avoidance of full knee extension, which are consistent with the motion list in Table I.
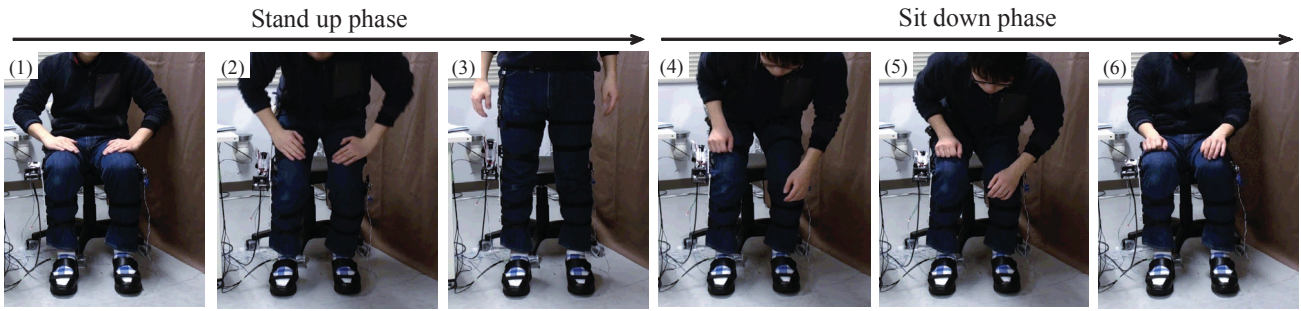
Fig. 3: Healthy person mimicked impaired standing-up and sitting-down motions. Subject's feet positions remained fixed. He imitated the abnormality in his right knee. Pictures were arranged chronologically from left to right. The robot was not actuated, but was used for data collection.
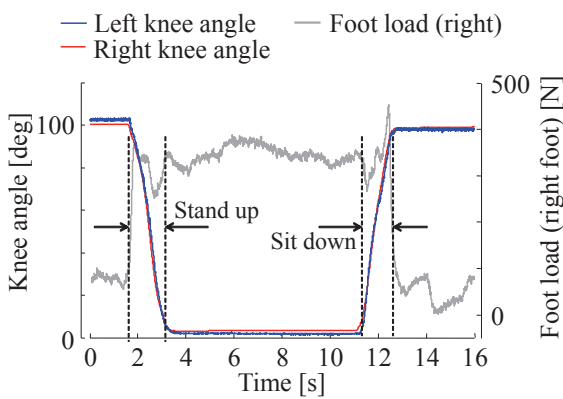

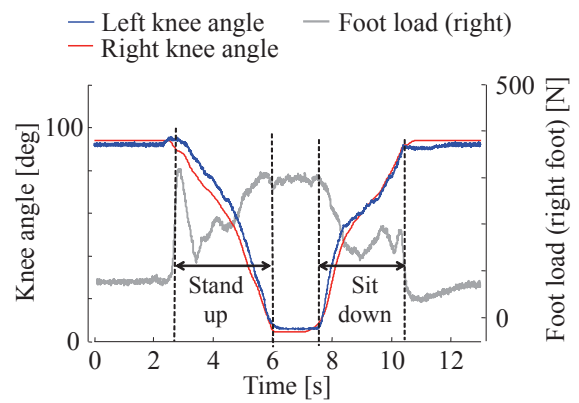Fig. 4: Knee angles and foot load of healthy motion.

Fig. 5: Knee angles and foot load of model motion of the impaired. Performed by a healthy adult without robot control.
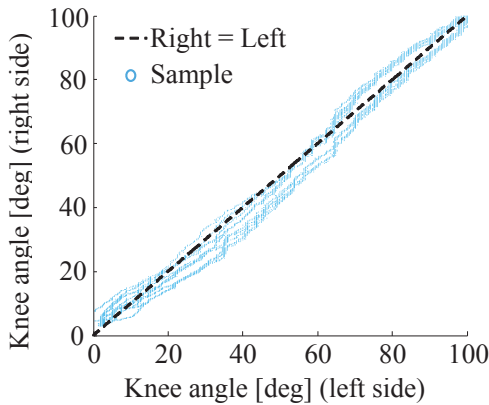

Fig. 6: Knee angles during normal sitting down on and standing up from chair. A dotted line is $\theta_{\mathrm{ref}} = \theta_{\mathrm{h}}$.
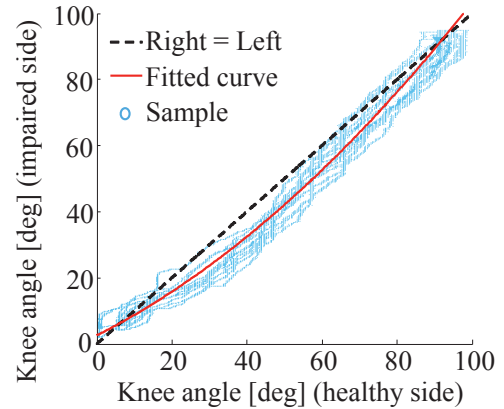

Fig. 7: Knee angles during model motion of sitting down on and standing up from chair with fixing feet positions. A red curve is a fitted quadratic function. A dotted line is $\theta_{\mathrm{ref}} = \theta_{\mathrm{h}}$.

We also observed some preferable characteristics regarding the foot loads. In the case of healthy motions, the loads were largely constant at 340 N during the standing-up (at 2.0–3.0 s in Fig. 4) and sitting-down (at 11.5–12.5 s) phases. The loads were almost equal for the right and left feet, provided that 340 N was approximately half of the subject's weight (70 kg). In contrast, in the case of the imitated motions, the foot load on the impaired side was obviously reduced (150–200 N) in the transition phases of the sitting and standing motions, as shown in Fig. 5 at the time intervals of 3.6–5.0 s and 8.0–10.5 s.

Stand up phase   Sit down phase
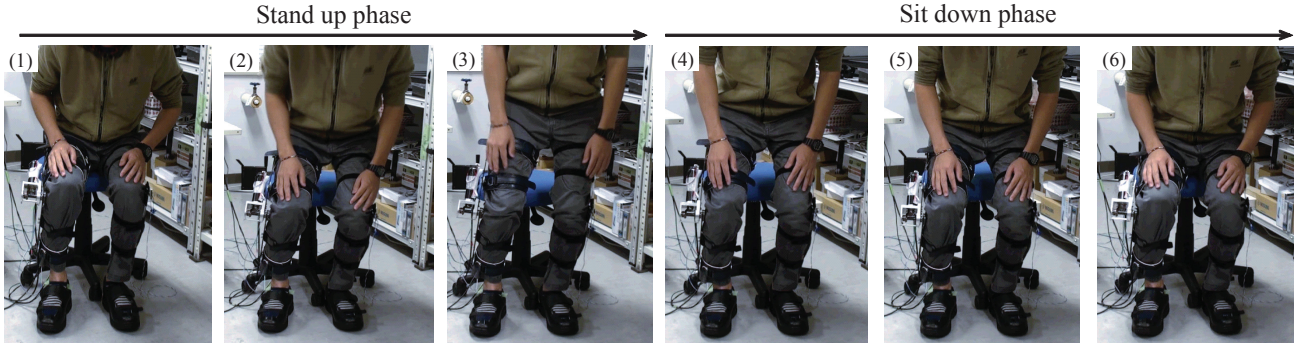
(1) (2) (3) (4) (5) (6)

Fig. 8: Simulated impaired standing-up and sitting-down motions with robot control. Subject's feet positions remained fixed. Pictures were arranged chronologically from left to right.

These loads were approximately 20–30% of the subject's body weight, and the results corresponded with the features in the motion list.

The aforementioned observations indicate that the model motions performed by the healthy person were consistent with those in the motion list in Table I. Therefore, the model motions can be references for the invariant motions of typical knee-OA patients.

V. TORQUE-CONTROL BASED ON THE FORMULATED TYPICAL MOTIONS OF KNEE-OA PATIENTS

A. Formulation of impaired knee angle

Because the wearer wore the robot on his right leg, only his right knee joint was directly influenced by the robot. To determine the set values of the right-knee angle, we formulated the model motions, which were defined in the previous section, as a function of the knee angle of the healthy (left) side. This enabled us to imitate the impaired motion using only a one-sided robot.

Fig. 6 shows the samples (green markers) of the right and left knee angles when a healthy adult repeated normal sitting-down and standing-up motions six times. For the healthy motions, $\theta_{\mathrm{ref}} = \theta_{\mathrm{h}}$ nearly held. In contrast, as shown in Fig. 5, the impaired knee angle tended to be smaller than that of the healthy side at moderate knee angles, and this trend did not rely on the type of motions, i.e., sitting-down and standing-up motions. Fig. 7 shows the samples when the subject imitated the abnormalities of the impaired. The two knee angles were effectively expressed by a quadratic function. Hence, we formulated the set angle of the impaired knee θ_{ref} as a quadratic function of the healthy knee angle θ_{h} as follows:

$$\theta_{\mathrm{ref}}(\theta_{\mathrm{h}}, t) = a_2 \theta_{\mathrm{h}}(t)^2 + a_1 \theta_{\mathrm{h}}(t) + a_0 \quad (1)$$

where a_2, a_1, a_0 were the coefficients specified by a least-squares method. The relationship between the two knee angles of the imitated motion (degree) was formulated as

$$\theta_{\mathrm{ref}}(\theta_{\mathrm{h}}, t) = 0.0043 \, \theta_{\mathrm{h}}(t)^2 + 0.57 \, \theta_{\mathrm{h}}(t) + 2.6. \quad (2)$$

The red curve in Fig. 7 is the function defined by (2). Using this function, we could specify the set knee angle of the impaired side from that of the healthy side.

B. Torque-control rules to realize model motions

We controlled the torque output of the robotic knee joint as follows:

$$\tau(t) = \tau_{\mathrm{a}}(t) + \tau_{\mathrm{b}}(t) \quad (3)$$

$$\tau_{\mathrm{a}}(t) = K_{\mathrm{p}} \frac{f(t)}{f_{\max}} (\theta_{\mathrm{ref}}(\theta_{\mathrm{h}}, t) - \theta_{\mathrm{i}}(t)) \quad (4)$$

$$\tau_{\mathrm{b}}(t) = -C_{\mathrm{p}} \frac{f(t)}{f_{\max}} \dot{\theta}_{\mathrm{i}}(t) \, g(\theta_{\mathrm{i}}) \quad (5)$$

$$g(\theta_{\mathrm{i}}) = \frac{1}{1 + \exp(-\mathrm{a}(\theta_{\mathrm{i}}(\mathrm{t}) - \mathrm{b}))} \quad (6)$$

where K_{p}, $f(t)$, and $f_{\max}$ were the proportional gain, total foot load of the impaired side, and weight of the wearer, respectively. $g(\theta_{\mathrm{i}})$ was a sigmoid function that smoothly switches from 0 to 1 around a given knee angle. a and b were set to 1.0 and 100, respectively such that $g(\theta_{\mathrm{i}})$ becomes 0 and 1 below and over: $\theta_{\mathrm{i}} = 100$ deg. K_{p} and C_{p} values were determined on a trial and error basis.

The output torque $\tau(t)$ was determined by two functions of $\tau_{\mathrm{a}}(t)$ and $\tau_{\mathrm{b}}(t)$. $\tau_{\mathrm{a}}(t)$ was in proportion to value of difference between measured knee angle of impaired side and reference one and made wearer's motions follow the model motions. $\tau_{\mathrm{b}}(t)$ prevented the wearer's motions from going out of the space of the model motions, concretely it prohibited the wearer from moving under deep flexion (higher than 100 deg, therefore $b = 100$ on (6)) of his impaired knee. Knee-OA patients avoid moving under such situation. Increase of foot load of the impaired side causes severe pain and it prevents knee-OA patients from moving. To represent this feature, both functions $\tau_{\mathrm{a}}(t)$ and $\tau_{\mathrm{b}}(t)$ were multiplied by the normalized foot load of the impaired side. For example, the robot exerts large torque when the wearer leans on the impaired side and no torque when the impaired foot does not contact the floor.

VI. EXPERIMENT

A. Tasks and participants

Our method to simulate the model motions was testified for standing-up and sitting-down motions. A healthy adult that was naive about our simulator wore the exoskeletal robot and repeated these motions for 20 times, of which 19 trials were
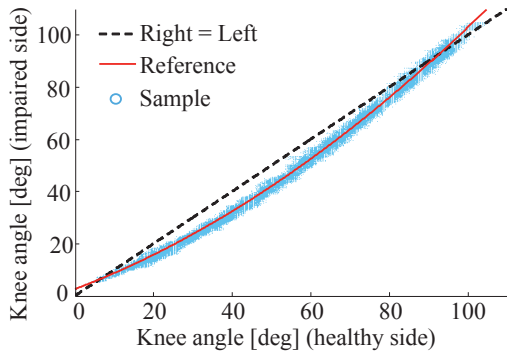
Fig. 9: Healthy and impaired knee angles of the simulated motions. The red curve is the model motion defined by (2).
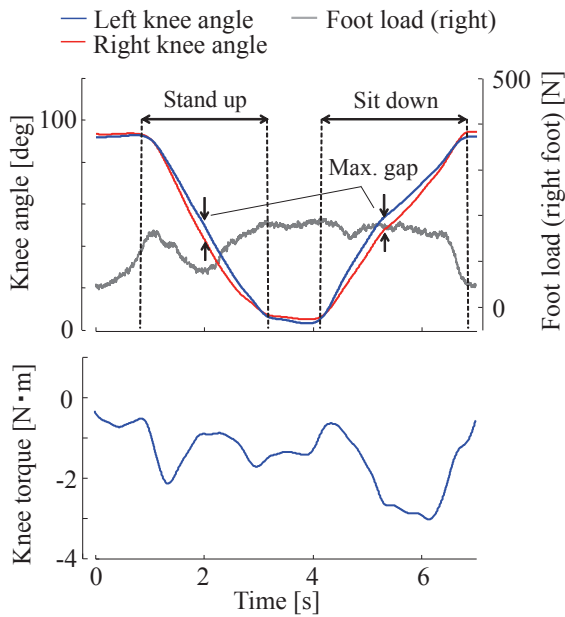

Fig. 10: Sample of simulated motion. Top: Knee angles and foot load of the impaired side. Bottom: Knee torque output by the wearable robot. Negative torque is to extend the wearer's knee.

valid and used for the later analysis. During the experiment, he was instructed not to attempt to resist the torque presented by the wearable robot, but to be guided by the robot. The robot does not exert torques large enough to forcefully move the wearer's knee, but it guides and impedes the wearer's motion.

B. Experimental results of movement simulation

Fig. 8 shows an example of the simulated motion acquired in the experiments. A body inclination to the healthy side, slower movements, and falls to the seat at the last phase of the sitting-down motions were observed in the simulated motions. These are consistent with the characteristic impaired motions listed in Table 1, suggesting that some features of the model motions were well represented.

Fig. 9 shows the samples of the right and left knee angles of the simulated motions for 19 trials, and the reference functions (red curve) defined by (2). They indicated that the knee angles of the simulated motions fairly matched the reference curve. A time course sample of the angles of the both knees shown in Fig. 10 also corroborate the similarity between the two types of motions, i.e. impaired model motion and simulated motion. Specifically, in the simulated motions, both knee angles did not exactly match, and the knee of the impaired side tended to be smaller (extension) than that of the healthy side.

Fig. 10 also shows the foot load measured at the impaired side. The foot load on the impaired side was smaller than 200 N for the large part of the motions whereas the weight of the wearer was 65 kg (637 N). In the case of the example shown in Fig. 10, they were 70–180 N (at 1.0–3.0 s in Fig. 10) and 50–180 N (at 8.0–10.0 s in Fig. 10) during the standing-up and sitting-down motions, respectively. This reduction of the foot load on the impaired side indicates the imbalanced body use of the wearer.

Fig. 10 (bottom) shows a sample of the torque output by the wearable robot. The negative torque is to extend the impaired knee. For this example, in the standing-up phase, the torque mostly acted to extend the impaired knee, which might have led to the body inclination to the healthy side. In the sitting-down phase, the torque was also generated in the direction of knee extension, which functioned as an impediment to the wearer's motion. As a result, the left leg preceded the right (supposedly impaired) leg.

One of the evident differences between the healthy and impaired motions lies in the imbalanced right and left knee angles. As previously described, for the impaired motion, right knee angle tends to be smaller (extension) than the left knee angle or the body inclines to the healthy side to avoid the loading on the impaired side. To test this angle differences between the right and left knees, we calculated the maximum difference during each of the simulated standing-up and sitting-down motions, as shown in Fig. 10, for all 19 trials. The means and standard deviations of the maximum knee angle differences were 7.2 ± 0.51 and 6.9 ± 0.64 deg for the standing-up and sitting-down motions, respectively, as shown in Fig. 11. These values are significantly higher than zero (standing up: $z = 61.8$, $p < 0.001$, sitting down: $z = 47.1$, $p < 0.001$), indicating that the right (impaired) knee angle was more extended than the left (healthy) knee angle.

The aforementioned results indicated that our torque-control rules physically guided a healthy wearer to the formulated model motions. In contrast, some apparent differences were observed between the model and simulated motions. The simulated motions were faster and the foot loads of the simulated motions did not completely match those of the model motions (comparison between Figs. 5 and 10). These differences were due to the limitation of the model function as determined by (1), which included only knee angles, but not angular velocities and force loads.

VII. CONCLUSION

To resolve the ethical barriers in experimental study involving motor-impaired patients, we developed a first prototype of a patient simulator using a wearable robot. We focused on
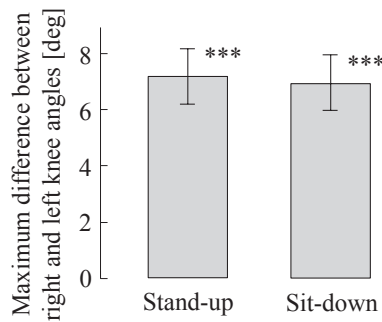
Fig. 11: Maximum differences between right and left knee angles of simulated motions. *** indicates statistically significant difference compared with zero at $p < 0.001$.

knee-OA patients because of their prevalence. Our strategy was first to formulate the typical motions of knee-OA patients. We then devised a control rule for the exoskeletal knee robot to guide the wearer's motions to match the designated patient's motions. In our experiment, sitting-down and standing-up motions were simulated by the wearer, and they were similar to those of typical knee-OA patients with regard to the temporal profiles of the knee angles, upper body inclination, and reduced foot load during the motions. The wearable patient simulator potentially allows us to conduct experiments to assess the accessibility of environments for people with motor impairment with few ethical challenges.

However, due to the primitiveness of the formulation of the model motions, the simulated motions did not dynamically match the model motions i.e. angular velocities of the knee and foot loads. Also, our method is not still applicable to other motions such as walking. Hence, some extension is fairly needed for further improvement.

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