2016 IEEE International Conference on Systems, Man, and Cybernetics • SMC 2016 | October 9-12, 2016 • Budapest, Hungary

Patient Simulator Using Wearable Robot to Estimate the Burden of Knee-Osteoarthritis Patients during Sitting-down and Standing-up Motions

Ryu Kubo
Nagoya University
Nagoya, Japan

Ayaka Hirukawa
Nagoya University
Nagoya, Japan

Shogo Okamoto
Nagoya University
Nagoya, Japan

Naomi Yamada
Nagoya University
Nagoya, Japan

Yasuhiro Akiyama
Nagoya University
Nagoya, Japan

Yoji Yamada
Nagoya University
Nagoya, Japan

***Abstract*—The estimation of the physical burdens from which people with motor impairment suffer helps us establish welfare techniques comprising personal care equipment and assessment of critical risks, such as fall risks. However, the involvement of actual patients in the evaluation and development of this equipment is costly and involves the exposure of patients to long and exhausting experiments. To solve this problem, we developed a robot wearable by a healthy person and the associated control algorithm to simulate typical motions of patients with knee osteoarthritis, which is a common symptom for the elderly and causes pain during movement. To estimate the physical burdens inflicted by knee malfunctions, we computed the knee flexion and extension moment of the simulated patient during the standing-up and sitting-down motions. The moments, estimated under certain conditions, are qualitatively consistent with those considered clinically, which corroborates the validity of our patient simulation techniques.**

I. Introduction

Recently, the development of personal care equipment and promotion of impediment removal at public facilities have become essential. For the fulfillment of the functional requirements of these objectives, it is necessary to estimate the burdens of people with motion impairments. However, the participation of impaired people in experiments that estimate their burdens or risks, during which they may suffer various degrees of danger or pain, is ethically unacceptable. Although these experiments are important to ensure the safety of patients, the subjects are likely to face significant dangers, especially in experiments involving hazardous situations that may cause a fall. Therefore, experiments in hazardous and painful situations can not be conducted by an impaired person.

This paper reports on the development of a patient simulator that provides a framework to simulate the motions of an impaired person using a healthy person [1], to engage simulated patients in various experiments, and to estimate the burdens or risks of the impaired while avoiding ethical problems and risks. Experiments conducted by a healthy person instead of a real patient can lower the subject's risks to an acceptable level in various cases. Typically, risk assessment and evaluation experiments in the preliminary step of the development of personal care equipment are conducted by healthy subjects. However, these tests provide only limited data because of the large differences between the motor functions of healthy and impaired people. The introduction of our simulator to these experiments removes or alleviates these limitations and contributes to the development of personal care equipment.

Thus far, some research groups have reported that it is useful for healthy people to experience impaired motor functions. Wood and Verkey et al. [2], [3] reported on a workshop in which medical students experienced the motor functional difficulties of aging people by using several orthoses simulating their movements. Ullauri et al. [4] attempted to simulate the elderly's gait by restraining the muscle activity at the lower thighs using a taping technique. Many researchers conducted studies related to experiences of impaired motor functions that have contributed to the improvement of patients' quality of life [5]–[8]. Several companies (for example Koken Co., Ltd, Japan, and Sanwa Manufacturing Co., Ltd, Japan) sell aging simulators with springs and weights; however, these products are intended for enable healthy people to experience the inconvenience that patients feel and not to estimate the burdens that patients may face.

Some researchers have developed robots to simulate the motions of people with motion functional disabilities. Huang et al. [9] developed a robotic patient with actuators in both of shoulder joints for the training of medical students in the transfer of patients between a wheelchair and a bed. Ishilawa et al. [10], [11] constructed a robot that was wearable on a knee joint for the training of physical therapists in care and examination procedures. However, these studies did not consider situations in which patients actively move; for example, walking, standing up from or sitting down on a chair, and climbing stairs.

Several researchers have investigated the motions of the impaired and reported physical quantities related with the physical burdens incurred by the patients Mak et al. [12] compared the standing-up and sitting-down motions of healthy subjects and of Parkinson's disease patients and suggested that the patients' slow movements were caused by the reduction of the hip flexion and extension moment. Astephen et al. [13] analyzed the gait motions of patients with knee osteoarthritis (knee-OA) of different severities and proposed a relation between the severity of the symptom and joint moments during a gait. Anan et al. [14] reported the inefficiency of knee-OA patients' motions in terms of physical energy. These studies examined cases of real patients but not simulations of the impaired.

The purpose of the present study is the estimation of the physical burdens that impaired people experience in their daily lives by using our simulated patient that has been reported in the previous article [1]. We verified the proposed method by experiments involving standing up from and sitting down on a

chair. The standing-up and sitting-down motions are common motions in daily life and are easily influenced by a motor impairment.

In our experiment, a healthy person wearing an exoskeletal knee robot performed sitting-down and standing-up motions under different conditions, such as with or without a simulated impairment, different chair heights, and the use of a supportive hand guide. Subsequently, we compared the physical burdens evaluated during the simulated motions for all different conditions and examined their consistency with those considered in clinical settings to validate our physical burden estimation approach. This simulator can be applied to the evaluation of personal care equipment in its early development phase and to assess potential risks caused by the malfunction and ill design of this equipment.

II. Knee osteoarthritis

We selected knee-OA as our target disease because of its frequent occurrence and severe effects, which obstruct the patient's motions in daily life. According to a report of the Ministry of Health, Labor, and Welfare [15], the number of knee-OA patients in Japan, including those with underlying symptoms, was approximately 30 million.

Knee-OA develops with age and is generally accompanied by damage to the bone spurs, cartilages, and meniscus [16]. Patients suffer from symptoms such as a limited motion range, pain during motion, joint deformation, and inflammation [16], [17]. Patients with moderate knee-OA experience a little pain that scarcely obstructs motions like stair climbing. Intermediate-severity patients suffer from pain when moving their knee joints, especially when large loads are applied to the joints. The motions of these patients differ from those of healthy persons, as the patients tend to avoid painful movements. Severe-knee-OA patients experience difficulties in their daily activities because of significant and frequent pain. In this research, we focused on intermediate-knee-OA patients. In particular, the following conditions were considered: OA develops in the right knee. The apparent motions of patients are distinct from those of healthy people. Finally, the patients do not need supportive instruments, such as hand rail or a cane.

III. Experimental equipment

As shown in Fig. 1, a healthy adult wore a wearable exoskeletal knee robot on the right leg by two plastic braces for the lower and femoral thigh. We installed 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) in the knee joint. This motor was driven by a current driver (4-Q-DC ADS50/5, Maxon Motor) through a computer control unit at 5 kHz. A three-axial accelerometer (KXM1050, Kionics, America) was attached on the middle point of the lower link of the robot. A potentiometer-based goniometer was installed on the left knee using Velcro tapes.

The interaction forces between the ground and the right foot were measured by a shoe in which three three-axial force sensors (USL08-H18-1kN-AP, Tec Gihan, Japan) were mounted. The sensors were placed on the heel and the thenar and antithenar eminences. These sensors were appropriately leveled, such that they covered the load paths between the human and the shoe sole.

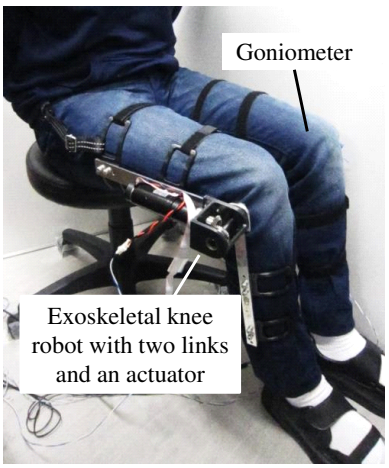

Fig. 1: Exoskeletal knee robot worn by healthy subject (right leg) and goniometer (left leg, hidden). [1]

IV. Method of patient simulation by a healthy person

We adopted the method developed in our previous study [1] to simulate typical motions of knee-OA patients. In this method, an exoskeletal knee robot leads the wearer through motions modeled after typical impaired motions.

First, we determined the representative movements of knee-OA patients using the literature [18] and observations of a physical therapist (N.Y., one of the authors of this article). Secondly, a healthy adult performed these representative movements as a model motion under the supervision of N.Y. Since actual knee-OA patients generally develop other diseases, our approach aided us to focus on representative behaviours caused by knee-OA. While performing this motion, he remained his feet on the ground. Fig. 2 shows both knee angles and the foot load of the impaired side of the model motion. The impaired knee shows a greater tendency to extend than the healthy knee and the foot load of the impaired side decreases during the standing-up and sitting-down motions in the model motion. These typically impaired motions develop for avoiding the possible pains on the impaired knee.

Fig. 3 shows samples (gray markers) of both knee angles during the standing-up and sitting-down model motions performed by the healthy subject, which were repeated five times for each. We formulated the desired angle of the impaired knee, θ_{ref}, as a quadratic function of the knee angle of the healthy side, θ_{h}, as follows:

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

where a_2, a_1, a_0 are coefficients calculated by the least-squares method. From this approximate equation, we can determine the reference angle of the impaired knee that corresponds to the measured angle of the healthy knee at a certain moment.

We controlled the torque output of the exoskeletal robot according to this equation:

$$\tau(t) = K_{\mathrm{p}}\,\frac{f(t)}{f_{\max}}(\theta_{\mathrm{ref}}(\theta_{\mathrm{h}}, t) - \theta_{\mathrm{i}}(t)) \tag{2}$$

where $f(t)$, $f_{\max}$, and K_{p} are the total foot load of the

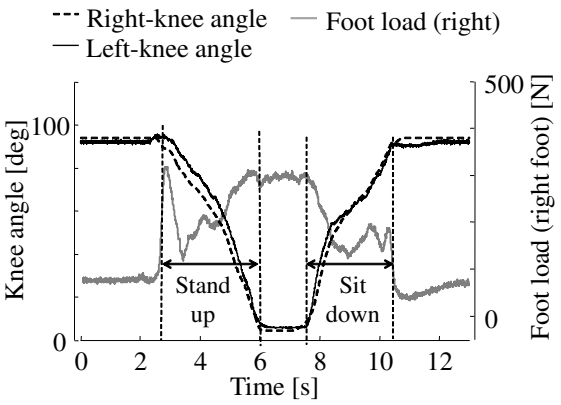

Fig. 2: Knee angles and foot load of model impaired motion performed by a healthy adult without robot control.

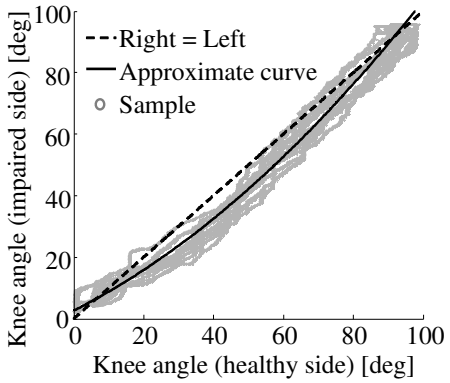

Fig. 3: Knee angles during the model motions of sitting down on and standing up from a chair with fixed feet positions. A black curve is a quadratic function used to approximate the samples. The dotted line corresponds to $\theta_{\mathrm{ref}} = \theta_{\mathrm{h}}$.

impaired side, the weight of the wearer, and the proportional gain, respectively. The output torque $\tau(t)$ is proportional to the difference between the measured impaired-knee angle $\theta_i(t)$ and the reference one and guided the wearer's motions to the model motions.

We have verified this simulation method in a previous study [1]. One healthy adult who had no background information on our apparatus repeated the standing-up and sitting-down motions under the guidance of the robot. We observed several similarities between the motions of the subject and those of the model; for example, an inclination of the upper body, load reduction on the impaired knee, and a significant extension of the impaired knee compared to the healthy knee. These results indicate that our method allows us to simulate the apparent motions of the patients by a healthy person using the wearable robot.

V. Estimation of the Patient's Burden Based on the Knee Extension and Flexion Moments

Our simulator aims to estimate the patient's physical burdens through experiments conducted by a simulated patient.

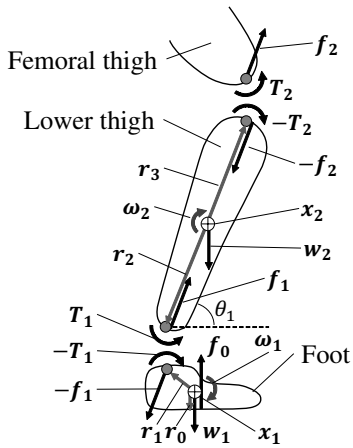

Fig. 4: Multibody diagram of a leg in the sagittal plane.

Earlier studies indicated that the adduction moment of the impaired knee is correlated with subjective pain during motion [17]. Some researchers have analyzed the efficiency of the sit-to-stand motion based on the moment of the knee joints [12], [19]. Hence, it is reasonable to consider that the flexion and extension moments of the knee, which are kinetic strains, are related with subjective burdens experienced by knee-OA patients.

First, we describe the method used to calculate the moments. Fig. 4 shows a multibody diagram of a lower limb on a sagittal plane. We formulate the motion equation of the foot as follows:

$$\boldsymbol{f}_0 - \boldsymbol{f}_1 + \boldsymbol{w}_1 = m_1 \ddot{\boldsymbol{x}}_1 \tag{3}$$

$$\boldsymbol{f}_0 \times \boldsymbol{r}_0 + \boldsymbol{f}_1 \times \boldsymbol{r}_1 - \boldsymbol{T}_1 = I_1 \dot{\boldsymbol{\omega}}_1 \tag{4}$$

and that of the lower thigh as follows:

$$\boldsymbol{f}_1 - \boldsymbol{f}_2 + \boldsymbol{w}_2 = m_2 \ddot{\boldsymbol{x}}_2 \tag{5}$$

$$\boldsymbol{f}_1 \times \boldsymbol{r}_2 + \boldsymbol{f}_2 \times \boldsymbol{r}_3 + \boldsymbol{T}_1 - \boldsymbol{T}_2 = I_2 \dot{\boldsymbol{\omega}}_2 \tag{6}$$

where $\boldsymbol{r_0}$, $\boldsymbol{r_1}$, $\boldsymbol{r_2}$, $\boldsymbol{r_3}$ are the position vector, $\boldsymbol{T_1}$, $\boldsymbol{T_2}$ are the moment at each joint, $\boldsymbol{f_0}$ is the floor reaction force, and, $\boldsymbol{f_1}$, $\boldsymbol{f_2}$ are the joint force. $\boldsymbol{w_1}$, $\boldsymbol{w_2}$ are the gravity force, $\boldsymbol{\omega_1}$, $\boldsymbol{\omega_2}$ are the angular velocity of the center of gravity, $\boldsymbol{x_1}$, $\boldsymbol{x_2}$ are the position vector of the center of gravity, I_1, I_1 are the moment of inertia, and m_1, m_2 are the mass of each body link. We assume $\ddot{\boldsymbol{x}}_1 \simeq 0$ and $\dot{\boldsymbol{\omega}}_1 \simeq 0$ because the subject's feet did not leave the ground and hardly moved from their initial positions during the standing-up and sitting-down motions. We also assume $\ddot{\boldsymbol{x}}_2 \simeq 0$ because the acceleration of the lower limb was significantly smaller than the other physical quantities. Finally, using equations (3)–(6) we formulate the knee flexion and extension moments during the motions as follows:

$$\begin{aligned}\boldsymbol{T}_2 =& \boldsymbol{f}_0 \times \boldsymbol{r}_0 + (\boldsymbol{f}_0 + \boldsymbol{w}_1) \times \boldsymbol{r}_1 + (\boldsymbol{f}_0 + \boldsymbol{w}_1) \times \boldsymbol{r}_2 \\ &+ (\boldsymbol{f}_0 + \boldsymbol{w}_1 + \boldsymbol{w}_2) \times \boldsymbol{r}_3 - I_2 \dot{\boldsymbol{\omega}}_2.\end{aligned} \tag{7}$$

VI. Experiment

A. Objectives

We conducted experiments to verify our method for estimating the burdens experienced by patients. Generally, knee-OA patients try to limit burdens on their impaired knees

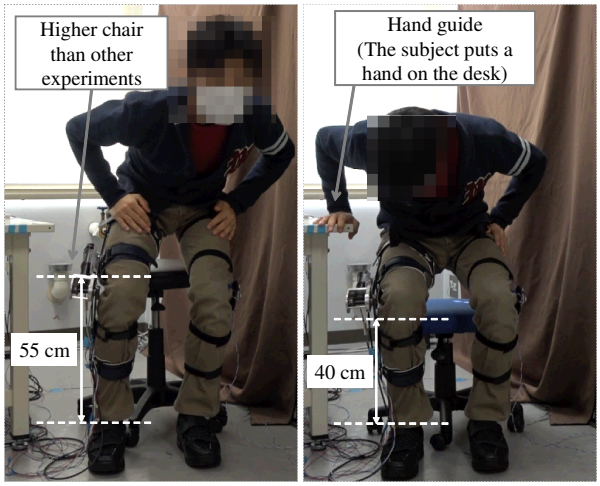

Fig. 5: Experimental scenes. Left: the case of a high chair (experiment 3). Right: the case of a hand guide (experiment 4).

because overloading the impaired knee causes pain. Additionally, physical therapists instruct patients on movements that reduce loads on the impaired knee. We conducted standing-up and sitting-down experiments under certain conditions, namely, control motion by a healthy person, simulated impaired motion, and simulated impaired motion following the instructions of a physical therapist. We then computed and analyzed the flexion and extension moments of the knee for each condition. We expected that the moments would be large in order of the control motion, the simulated motion, and the simulated motions following clinical instructions. If the burdens estimated under these conditions qualitatively match those observed clinically, then it corroborates the validity of our patient simulation technique.

B. Participants

The participant was one healthy man who had no background information on our apparatus and experimental objectives.

C. Tasks

We conducted experiments of standing-up and sitting-down motions under the following conditions:

Experiment 1: Case of a healthy person
Experiment 2: Case of a simulated patient
Experiment 3: Case of a simulated patient with a high chair
Experiment 4: Case of a simulated patient with a hand guide.

The height of the high chair was 55 cm and that of the normal chair was 40 cm. Fig. 5 shows the experimental environments of experiments 3 and 4 according to clinical settings.

D. Analysis

We analyzed the maximum and mean values of the flexion and extension moments during the examined motions using a t-test. We adjusted the significance level based on the Bonferroni correction for comparing of experiments 2-3 and 2-4.

E. Results

The number of valid trials in experiments 1, 2, 3, and 4 were 10, 8, 8, and 8, respectively.

1) Apparent and representative features of motions under each condition: Figs. 6–9 show the knee angles and the flexion and extension moments during the examined motions in a representative trial. In experiments 2, 3, and 4, we observed slower movements and a body inclination to the healthy side (not shown in the figures). Experiment 1 showed a full knee extension in the standing position that was not observed in the other experiments. These results are consistent with the characteristics of knee-OA patients.

In experiment 1, the moments exhibited large values, which were approximately 20 N·m at the times when the participant's hip separated from (time t_{A}) and contacted to (time t_{B}) the seat. In experiment 2, the moments showed peaks at times t_{A} and t_{B}, but the measured values were smaller than 10 N·m and lower than those of experiment 1. In experiment 3, the moments did not showed a peak at times t_{A} and t_{B}, and their values were lower than 10 N·m throughout the motions. In experiment 4, we observed small peaks of the moments at times t_{A} and t_{B} and their values were lower than 10 N·m, similarly to experiment 3. The simulated patient exhibited lower moments than the healthy person, and the two cases (experiments 3 and 4) in which they followed clinical instructions showed lower values than the other cases. These results reflect the characteristics of knee-OA patients and the clinical instructions provided to reduce the burden on the impaired side.

2) Statistical analysis: Figs. 10 and 11 show the maximum and mean values of the flexion and extension moments of the impaired (right) knee in each condition. We compared these values between experiments 1-2, 2-3, and 2-4 using t-tests with the Bonferroni correction in order to test the differences of the physical burdens due to the disease, the height of the chair, or the hand guide.

2-1) Validity and effect of simulated impairment: We compared experiments 1 and 2, which differed in terms of the presence of the disease, to test the effect of our simulated impairment. In experiment 1, both the maximum and mean values of the moments of the right (impaired) knee were significantly larger during the standing-up and sitting-down motions than those in experiment 2 ($p < 0.001$). These results indicated that the burden on the impaired knee was smaller for the patient than for the healthy person. In general, knee-OA patients lean their loads on the healthy side and reduce the burden applied to the impaired side. The experimental results were in good agreement with this general trend of knee-OA patients and confirmed that our simulated patient developed one of the typical characteristics of actual knee-OA patients.

2-2) Effects of chair-height and a hand guide on burdens: By employing our simulated patient, we confirmed that the knee flexion and extension moments decrease when following clinical instructions for knee-OA patients.

First, we compared experiments 2 and 3, in which the chairs of 40 and 55 cm height were used, respectively. The maximum moment of the impaired knee was significantly larger in experiment 2 than in experiment 3 during the sitting-down motion ($p < 0.001$) and the standing-up motion ($p < 0.1$). Similarly, the mean moment value in experiment 2 was considerably larger than that observed in

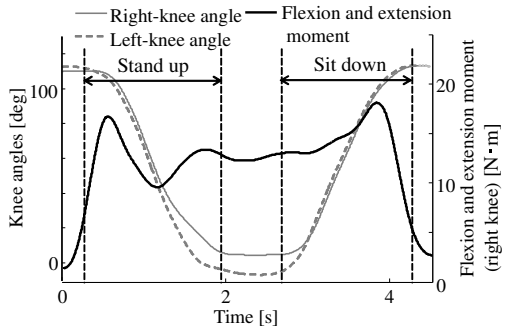

Fig. 6: Experiment 1. Knee angles and flexion and extension moment of the right knee in the case of a healthy subject. The chair height was 40 cm and no hand guide was used. A greater the moment is observed at the beginning of the standing-up motion and the end of the sitting-down motion.

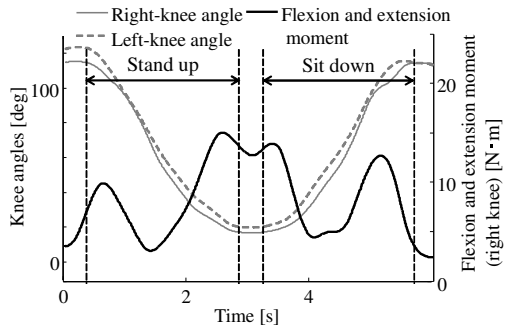

Fig. 7: Experiment 2. Knee angles and flexion and extension moment of the impaired knee. A healthy adult simulated knee-OA patients with robot control. The chair height was 40 cm and no hand guide was used.

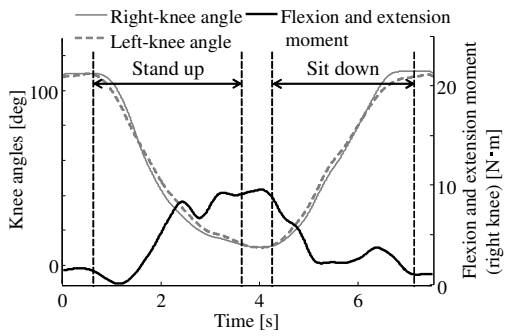

Fig. 8: Experiment 3. Knee angles and flexion and extension moment of the impaired knee. A healthy adult simulated knee-OA patients with robot control, using a high chair (height = 55 cm), and no hand guide.

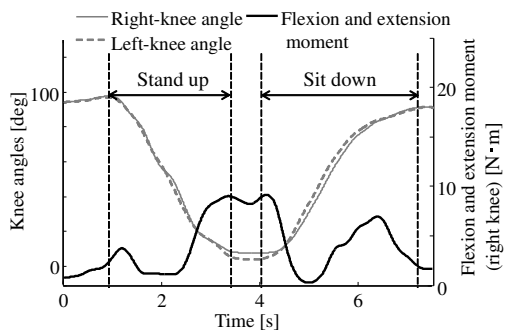

Fig. 9: Experiment 4. Knee angles and flexion and extension moment of the impaired knee. A healthy adult simulated knee-OA patients with robot control and a hand guide. The chair height was 40 cm.

experiment 3 during the standing-up and the sitting-down motions ($p < 0.001$). Using a high chair following clinical instructions enables patients to move their center of gravity easily at the times t_{A} and time t_{B}. Furthermore, it reduces the burden on the impaired knee. We confirmed the same tendency for our simulated patient, whose knee moments were smaller for the higher chair than for the shorter one.

Secondly, we compared experiments 2 and 4, which differed in terms of the use of a hand guide. Both the maximum and the mean values of the flexion and extension moments of the impaired knee were significantly larger in experiment 2 than in experiment 4 during the standing-up (maximum moment: $p < 0.001$, mean moment: $p < 0.01$) and sitting-down motions (maximum moment: $p < 0.01$, mean moment: $p < 0.01$).

According to the clinical instructions for knee-OA patients, the use of a hand guide reduces the burden on the impaired knee by distributing the body weight to three points. In the experiment involving our patient simulation, the moment along the impaired knee decreased because of the use of the hand guide, which is in agreement with the clinical cases of actual patients.

The general effects of the clinical instructions on the burdens experienced by actual patients are consistent with those on the knee moments of the simulated patient. As previously stated, the physical burdens on the impaired knees of knee-OA patients that follow clinical advice or instructions from therapists are smaller than those of patients who do not. Furthermore, because patients avoid overloading their impaired knees, the physical burdens on these knees are smaller than those of healthy people. These trends were represented using our patient simulation, in which the moment around the knee was used as an index of physical burden. These results corroborate the validity of the patient simulation, although further validation from different viewpoints remains to be studied.

VII. Conclusion

To solve economic, safety, and ethical problems related to experiments involving motor-impaired patients, we have developed a patient simulator with which a healthy person mimics the impaired motion with the guidance of an exoskeletal robot. Such techniques will help us evaluate the effectiveness of personal care equipment by allowing us to estimate the physical burdens that motor-impaired users may face.

In this study, we demonstrated a simulation of the sitting-down and standing-up motions of knee-OA patients. Using

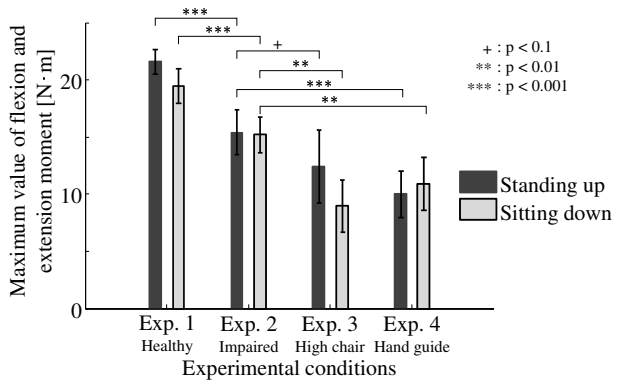

Fig. 10: Maximum values of the flexion and extension moments of the knee during each of the standing-up and sitting-down motions. Comparison between experiments 1-2, 2-3, and 2-4 using t-tests with the Bonferroni p-value correction. The error bars are the standard deviation calculated for all trials.

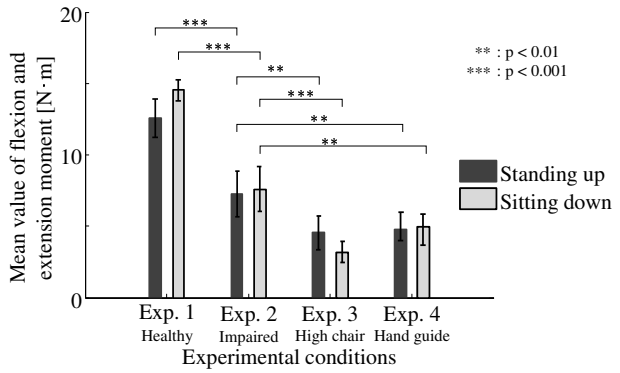

Fig. 11: Mean values of the flexion and extension moments of the knee during each of the standing-up and sitting-down motions. Comparison between experiments 1-2, 2-3, and 2-4 using t-tests with the Bonferroni p-value correction. The error bars are the standard deviation calculated for all trials.

the flexion and extension moment around the impaired knee, we evaluated the degree of physical burden during these motions. We compared the knee moments under four different conditions: control (healthy), simulated impairment, simulated impairment with a high chair, and simulated impairment with a hand guide. We then found that the order of the maximum and average knee moments measured in our experiments was in good agreement with those observed in clinical settings, which in part corroborates the validity of our patient simulation method. Whereas, further detailed validations are necessary. The developed patient simulator may enable us to estimate realistically the physical burdens of patients with motor impairment by employing healthy subjects.

REFERENCES

[1] R. Kubo, S. Okamoto, S. Nezaki, N. Yamada, Y. Akiyama, and Y. Yamada, "Patient simulator using wearable robot: Representation of invariant sitting-down and standing-up motions of patients with knee-OA," *IEEE Industrial Electronics Conference*, pp. 4808–4813, 2015.

[2] M. D. Wood, "Experiential learning for undergraduates: A simulation about functional change and aging," *Gerontology & Geriatrics Education*, vol. 23, no. 2, pp. 37–48, 2003.

[3] P. Varkey, D. S. Chutka, and T. G. Lesnick, "The aging game: Improving medical students' attitudes toward caring for the elderly," *Journal of the American Medical Directors Association*, vol. 7, no. 4, pp. 224–229, 2006.

[4] J. Beltran Ullauri, Y. Akiyama, N. Yamada, S. Okamoto, and Y. Yamada, "Muscle activity restriction taping technique simulates the reduction in foot-ground clearance in the elderly," *IEEE International Conference on Rehabilitation Robotics*, pp. 559–564, 2015.

[5] J. T. Pacala, C. Boult, and K. Hepburn, "Ten years' experience conducting the aging game workshop: Was it worth it?" *Journal of the American Geriatrics Society*, vol. 54, no. 1, pp. 144–149, 2006.

[6] S. B. Robinson and R. B. Rosher, "Effect of the half-full aging simulation experience on medical students' attitudes," *Gerontology & Geriatrics Education*, vol. 10, no. 3, pp. 3–12, 2001.

[7] B. W. Henry, C. Dourglass, and I. M. Kostiwa, "Effects of participation in aging game simulation activity on the attitudes of allied health students toward older adults," *Internet Journal of Allied Health Sciences and Practice*, vol. 5, no. 4, 2007.

[8] M. A. O'Brien, C. B. Fausset, E. L. Mann, and C. N. Harrington, "Using impairment simulation tools to demonstrate age-related challenges in everyday tasks and promote universal design," *Proceedings of the Human Factors and Ergonomics Society Annual Meeting*, vol. 58, no. 1, pp. 2402–2405, 2014.

[9] Z. Huang, T. Katayama, M. Kanai-Pak, J. Maeda, Y. Kitajima, M. Nakamura, K. Aida, N. Kuwahara, T. Ogata, and J. Ota, "Design and evaluation of robot patient for nursing skill training in patient transfer," *Advanced Robotics*, DOI:10.1080/01691864.2015.1052012, 2015.

[10] S. Ishikawa, S. Okamoto, Y. Akiyama, K. Isogai, and Y. Yamada, "Simulated crepitus and its reality-based specification using wearable patient dummy," *Advanced Robotics*, vol. 29, no. 11, pp. 699–706, 2014.

[11] S. Ishikawa, S. Okamoto, K. Isogai, Y. Akiyama, N. Yanagihara, and Y. Yamada, "Assessment of robotic patient simulators for training in manual physical therapy examination techniques," *Plos One*, vol. 10, no. 4, p. e0126392, 2015.

[12] M. K. Mak, O. Levin, J. Mizrahi, and C. W. Hui-Chan, "Joint torques during sit-to-stand in healthy subjects and people with Parkinson's disease," *Clinical Biomechanics*, vol. 18, no. 3, pp. 197–206, 2003.

[13] J. L. Astephen, K. J. Deluzio, G. E. Caldwell, and M. J. Dunbar, "Biomechanical changes at the hip, knee, and ankle joints during gait are associated with knee osteoarthritis severity," *Journal of Orthopaedic Research*, vol. 26, no. 3, pp. 332–341, 2008.

[14] M. Anan, K. Shinkoda, K. Suzuki, M. Yagi, T. Ibara, and N. Kito, "Do patients with knee osteoarthritis perform sit-to-stand motion efficiently?" *Gait & Posture*, vol. 41, no. 2, pp. 488–492, 2015.

[15] Ministry of Health, Labour, and Welfare, "The provision for motor system disease to advance preventive care," http://www.mhlw.go.jp/shingi/2008/07/dl/s0701-5a.pdf, 2015.2.26.

[16] R. D. Altman, "The Merck manual for health care professionals - Osteoarthritis (OA)," http://www.merckmanuals.com/home/bone_joint_and_muscle_disorders/joint_disorders/osteoarthritis_oa.html, 2015.2.26.

[17] N. Kito, K. Shinkoda, T. Yamasaki, N. Kanemura, M. Anan, N. Okanishi, J. Ozawa, and H. Moriyama, "Contribution of knee adduction moment impulse to pain and disability in Japanese women with medial knee osteoarthritis," *Clinical Biomechanics*, vol. 25, no. 9, pp. 914–919, 2010.

[18] M. Anan, K. Tokuda, N. Kito, and K. Shinkoda, "Kinematic analysis of sit-to-stand motion in knee osteoarthritis," *Physiotherapy Science*, vol. 25, no. 5, pp. 755–760, 2010.

[19] Y. C. Pai and M. W. Rogers, "Speed variation and resultant joint torques during sit-to-stand," *Archives of Physical Medicine and Rehabilitation*, vol. 72, no. 11, pp. 881–885, 1991.