

A Standing Assistance Scheme Using a Patient's Physical Strength by a Load Estimation Considering the Muscle Arrangements of a Human Leg

Daisuke Chugo, Satoshi Muramatsu, Sho Yokota and Hiroshi Hashimoto

Abstract In this paper, we propose a standing assistance scheme that uses a patient's own physical strength, as evaluated by physical activity estimates. In general, conventional assistive robots do not require patients to use their own physical strength to stand, which leads to decreased strength in the elderly. Therefore, an assistive robot that allows a patient to maximally use their remaining physical strength is desired. Assistive robots can achieve this objective by estimating the physical activity of a patient when they stand. Therefore, the activity estimate proposed here is based on a human musculoskeletal model of a lower limb, which exhibits a biarticular muscle function. The patient generates a natural standing motion using the biarticular muscle function, and the proposed model enables the assistive robot to estimate the patient's physical activity without using biosensors such as electromyographs. Using the proposed estimated results, our prototype assistive robot can assist elderly patients to use their remaining physical strength maximally by selecting a suitable assistive control method.

D. Chugo (✉)

School of Science and Technology, Kwansai Gakuin University,
2-1 Gakuen, Sanda, Hyogo 6691337, Japan
e-mail: chugo@kwansai.ac.jp

S. Muramatsu

School of Information Science and Technology, Tokai University,
4-1-1, Kita-Kaname, Hiratsuka, Kanagawa 2591292, Japan
e-mail: muramatsu@tokai.ac.jp

S. Yokota

Faculty of Science and Engineering, Toyo University,
2100, Kujirai, Kawagoe, Saitama 3508585, Japan
e-mail: s-yokota@toyo.jp

H. Hashimoto

Master Program of Innovation for Design and Engineering,
Advanced Institute of Industrial Technology, 1-10-40, Higashi-Ooi,
Shinagawa, Tokyo 1400011, Japan
e-mail: hashimoto@aiit.ac.jp

© Springer International Publishing Switzerland 2016

J. Filipe et al. (eds.), *Informatics in Control, Automation and Robotics 12th International Conference, ICINCO 2015 Colmar, France, July 21-23, 2015 Revised Selected Papers*,
Lecture Notes in Electrical Engineering 383, DOI 10.1007/978-3-319-31898-1_8

Keywords Standing assistance · Musculoskeletal model · Joint traction · Physical activity estimate

1 Introduction

The act of standing may be the most serious and important activity in the daily life of an elderly person lacking physical strength [1, 2]. However, assisting elderly individuals in standing is a difficult task for caregivers and can be a primary source of the lumbago that many experience [3]. Therefore, creating a care service robot capable of assisting the elderly when they stand is important, and many such assistive devices have been developed and presented in previous works [4, 5].

In Japan, elderly people requiring assistance in daily life are classified into five different care levels [3], where care level 1 is minor and care level 5 represents a serious condition. Generally, the elderly people whose care level is 1 or 2 have difficulty in standing on their own but are able to perform normal daily life activities if standing assistance is provided. However, in many cases, standing assistance devices provide all the power necessary for the patient to stand and do not allow the patient to use their remaining physical strength. Thus, the patient's physical strength decreases [6]. In fact, between 2002 and 2003, more than 10% of care level 1 patients were subsequently assigned to higher care levels within the next year [3]. Thus, to improve the quality of life of elderly patients with low care levels, assistive robots should use the patient's remaining physical strength. However, no studies have been conducted toward this end.

Therefore, we have developed a novel assistive robot designed to aid patients in using their own physical strength to stand [7]. The robot is based on a walker (a popular assistance device for aged people in normal daily life) and uses a support pad actuated by manipulators with three degrees of freedom (Fig. 1) to assist patients in standing.

To maximally utilize the remaining physical strength of a patient while providing standing assistance, the robot is required to accurately estimate the physical activity of the patient so that it may coordinate its assistive force accordingly. However, generally, it is difficult to conduct such estimates without biosensors such as electromyographs (EMGs); furthermore, physical activity estimates with biosensors, which must be attached to the patient, is impractical because assistance robots should be low-cost and easy to use.

Previous works have proposed physical activity estimates using human models comprising linkages and joints without the use of biosensors [8]. These schemes evaluate the patient's physical activity using joint traction, which is calculated using the kinematic model as an index. However, many muscles generate human body movements, and the maximum amount of traction that muscles can generate changes according to the relative positions of the bones and muscles. Therefore, maximum joint traction is not constant, but changes according to the patient's posture. During

a standing motion, a patient’s posture changes considerably, which should be taken into consideration when evaluating a patient’s physical activity.

Therefore, in this paper, we propose a standing assistance scheme using a patient’s physical strength, which is evaluated by means of real-time physical activity estimates without additional biosensors. The paper is organized as follows: in Sect. 2, we propose the estimate scheme for a patient’s activity according to their posture during standing motion using a human musculoskeletal model of a lower limb, which expresses a biarticular muscle function; in Sect. 3, we propose a standing assistance control scheme on our robot, which uses a patient’s strength based on estimated results; in Sect. 4, we provide experimental results obtained using our prototype; and Sect. 5 concludes this paper.

2 Physical Activity Estimate

2.1 Overview of the Proposed Estimate Scheme

In the linkage model of a human body [9, 10], joint traction is used as an index of a patient’s load. However, this index does not consider the posture of the patient, and in some cases, this index diverges from the experience of nursing specialists, especially when the patient is in a half-sitting posture. When the patient stands, the muscles shown in Fig. 2 generate a lifting motion [11]. Many muscles (shown in Table 1) are used to accomplish the standing movement, and the traction, which muscles can

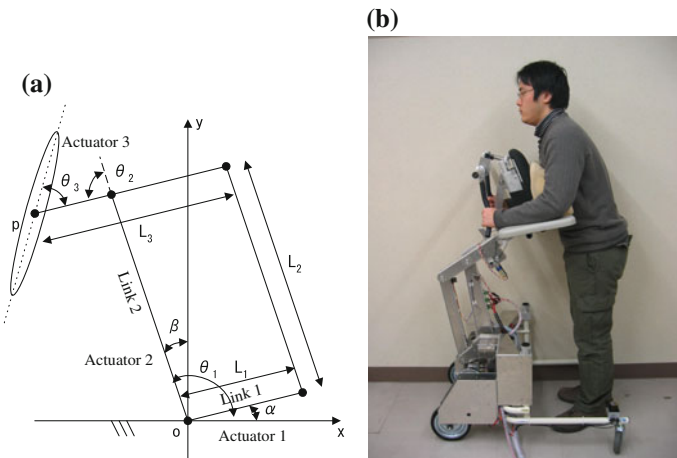


Fig. 1 Our developed robot for standing assistance. **a** Frame kinematic model. **b** Overview of our robot

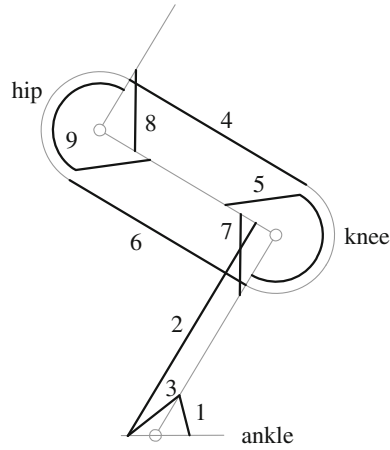


Fig. 2 Muscle arrangements in the human leg

Table 1 Human leg muscles

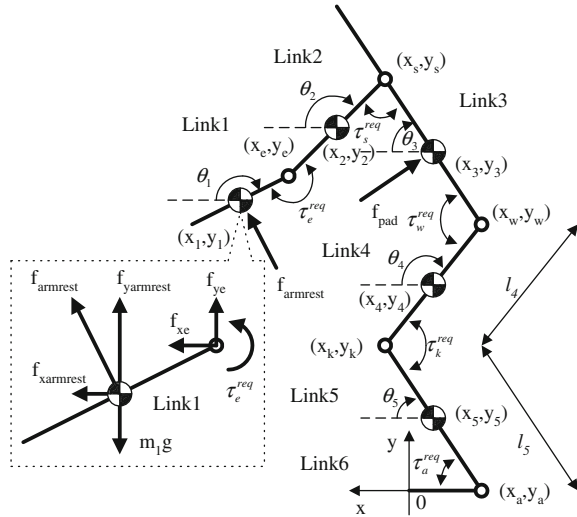
| No. | Muscle actuator | Physical areas (cm ²) |
|-----|--------------------------------------|-----------------------------------|
| 1 | Tibialis anterior (TA) | 19.7 |
| 2 | Gastrocnemius (GAS) | 99.1 |
| 3 | Soleus (SOL) | 247.6 |
| 4 | Rectus femoris (RF) | 43.5 |
| 5 | Vastus lateralis (VAS) | 248.1 |
| 6 | Semimembranosus (SM) | 60.2 |
| 7 | Biceps femoris and short head (BFSH) | 8.7 |
| 8 | Iliopsoas (IL) | 23.0 |
| 9 | Gluteus maximus (GMAX) | 20.0 |

maximally generate, changes according to the relative position between frames and muscles.

Thus, we propose a novel physical activity estimate scheme that takes all this into consideration. In this paper, we focus on the traction of the knee and waist joints, which are the main forces propelling patients to stand. Our proposed algorithm is as follows:

- First, we derive the required traction (knee joint τ_k^{req} and waist joint τ_k^{req}) to accomplish a standing motion with our assistive robot.
- Second, we derive the maximum traction (knee joint τ_k^{max} and waist joint τ_k^{max}) that the muscles can generate for the posture at the present time.
- Comparing the two derived tractions, we evaluate the physical activity of the patient, μ_i , which demonstrates how much the patient is required their own phys-

Fig. 3 Linkage model of a human body



ical strength as compared with their maximum power (1). i is the identification character (for example, in the case of the knee joint, i is k):

$$\mu_i = \frac{\tau_i^{req}}{\tau_i^{max}}. \tag{1}$$

2.2 Derivation of the Required Traction

To estimate the applied load to each joint, we approximate human motion based on the movement of the linkage model on a two-dimensional (2D) plane [9]. Using this model, we can derive the traction of each joint and estimate the patient's load.

The assistance system is designed in such a way that patients lean on a pad and grasp an armrest while standing with our assistance (we will explain our prototype more closely in the next section), which means that our system uses the pad to apply force to the patient's chest and the armrest to apply force to their forearm. These forces move vertically (at the pad) and horizontally (at the armrest). Considering these conditions, we propose a linkage model that approximates the human body with our assistance device (see Fig. 3).

This model consists of six linkages. The armrest applies the assistance force ($f_{armrest}$) to the center position of Link 1 and the support pad applies the force (f_{pad}) to the center position of Link 3. m_i is the mass of the link ($i = 1, \dots, 6$) and I_i is the moment of inertia. (x_i, y_i) is the position of the center of gravity on each link, and (x_i, y_i) ($i = a, k, w, s,$ and e) is the position of each joint. We assume that each linkage is in pillar form with its mass distributed uniformly:

Table 2 Human body parameters

| No. | Name | M (%) | C.G (%) | K (%) | Length (m) |
|-----|---------|-------|---------|-------|------------|
| 1 | Forearm | 3.2 | 41.5 | 27.9 | 0.35 |
| 2 | Humerus | 5.4 | 52.9 | 26.2 | 0.39 |
| 3 | Trunk | 57 | 49.3 | 34.6 | 0.48 |
| 4 | Femur | 22 | 47.5 | 27.8 | 0.61 |
| 5 | Leg | 10.2 | 40.6 | 27.4 | 0.56 |
| 6 | Foot | 2.2 | 59.5 | 20.4 | 0.26 |

M The ratio of the mass of the body segment to that of the body

C.G. The ratio of segmental length, which shows the location of the center of gravity on the longitudinal axis

K The ratio of the gyration radius of the body segment to the length of its segment

$$\begin{aligned}
\tau_w^{req} = & -(y_s - y_t) f_{xarmrest} - (x_s - x_t) f_{yarmrest} \\
& + (y_w - y_3) f_{xpad} + (x_w - x_3) f_{ypad} \\
& + m_3 \{(y_w - y_3) \ddot{x}_3 - (x_w - x_3) (\ddot{y}_3 - g)\} \\
& + m_2 \{(y_s - y_w) \ddot{x}_2 - (x_s - x_w) (\ddot{y}_2 - g)\} \\
& + m_1 \{(y_s - y_w) \ddot{x}_1 - (x_s - x_w) (\ddot{y}_1 - g)\} + I_3 \ddot{\theta}_3 - \tau_s^{req}
\end{aligned} \quad , \quad (2)$$

$$\begin{aligned}
\tau_k^{req} = & -(y_w - y_k) (f_{xarmrest} + f_{xpad}) \\
& + (x_w - x_k) (f_{yarmrest} + f_{ypad}) \\
& + m_4 \{(y_k - y_4) \ddot{x}_4 - (x_k - x_4) (\ddot{y}_4 - g)\} \\
& + m_3 \{(y_k - y_w) \ddot{x}_3 - (x_k - x_w) (\ddot{y}_3 - g)\} \\
& + m_2 \{(y_k - y_w) \ddot{x}_2 - (x_k - x_w) (\ddot{y}_2 - g)\} \\
& + m_1 \{(y_k - y_w) \ddot{x}_1 - (x_k - x_w) (\ddot{y}_1 - g)\} + I_4 \ddot{\theta}_4 - \tau_w^{req}
\end{aligned} \quad . \quad (3)$$

Here, we use body parameters chosen from a standard body of data from adult Japanese males [10]; see Table 2. To derive the required body parameters for calculating the moment force, we measure the length of each body segment and the mass of the entire body of each individual patient.

We know from previous research [12] that the maximum force that each muscle can realize at the ankle joint is F_{me1} , F_{me2} , F_{me3} , F_{mf1} , F_{mf2} , and F_{mf3} , and the output distribution of the force at the ankle joint is expressed kinematically as a hexagon (see Fig. 5).

The directions of F_{me1} and F_{mf1} are parallel to the leg, the directions of F_{me2} and F_{mf2} are parallel to the straight line that connects the waist and ankle joints, and the directions of F_{me3} and F_{mf3} are perpendicular to the leg. Furthermore, Oshima et al.'s previous research [12] demonstrates that there is a relationship between the force output vector and the activation level, η_i , of the muscle working in the force output direction. This relationship is shown in Fig. 4, and our system can estimate the

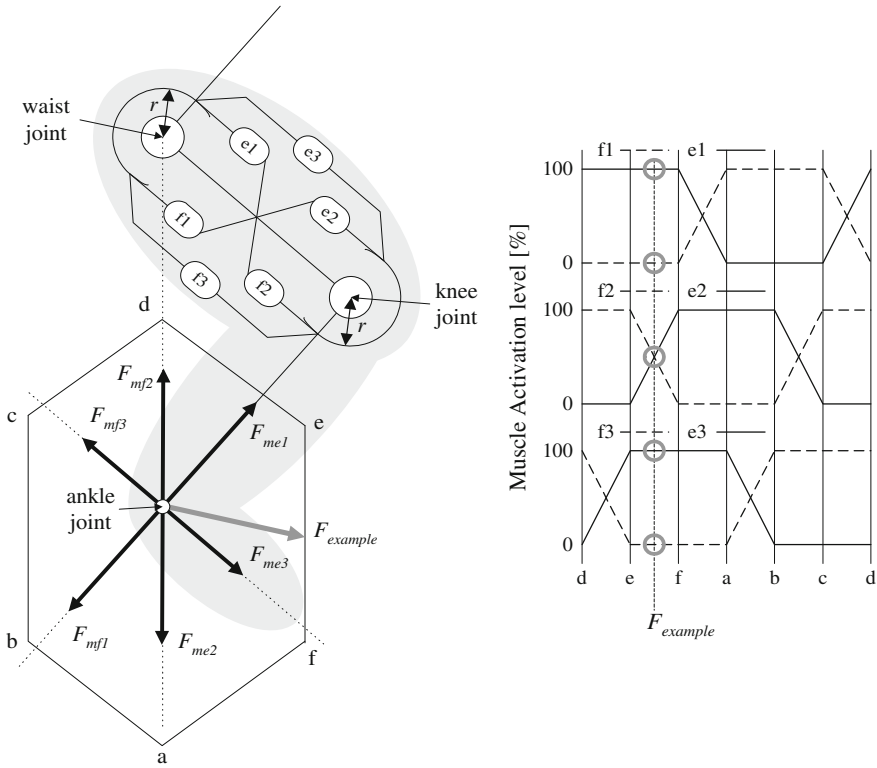


Fig. 4 Musculoskeletal model considering the role of the antagonistic and biarticular muscles

activation level of each muscle using the output force at the ankle joint. For example, when the output force is $F_{example}$, as in Fig. 4, the direction of the force vector is between e and f.

Therefore, the activation levels of each muscle are $\eta_{e1} = \eta_{e3} = 100(\%)$, $\eta_{f1} = \eta_{f3} = 0(\%)$, and $\eta_{e2} = \eta_{f2} = 50(\%)$, as shown in Fig. 4.

Using this model, we propose a physical activity estimate scheme for a patient according to their posture. First, our system calculates the required traction of the waist joint, τ_w^{req} , and of the knee joint, τ_k^{req} , using Eqs. (2) and (3), respectively. From the kinematic relationship shown in Fig. 4, the force output vector (f_x, f_y) at the ankle joint is derived as

$$\begin{aligned} & \begin{vmatrix} \tau_w^{req} \\ \tau_k^{req} \end{vmatrix} \\ &= \begin{vmatrix} l_5 \sin \theta_5 + l_4 \sin \theta_4 & -(l_5 \cos \theta_5 + l_4 \cos \theta_4) \\ l_5 \sin \theta_5 & -l_5 \cos \theta_5 \end{vmatrix} \cdot \begin{vmatrix} f_x \\ f_y \end{vmatrix}. \end{aligned} \tag{4}$$

Second, our system derives the distribution of the output force at the ankle joint from the patient's posture, and then adapts the force output vector (f_x, f_y) derived from (4) to the hexagon from Fig. 4, which expresses the distribution of the output force, and derives the muscle activation level, η_i , at this time.

We know from previous research [13] that the maximum force, F_i^{\max} , that a muscle can generate is

$$F_i^{\max} = A_i \sigma, \quad (5)$$

where A_i is the cross-sectional area of each muscle and σ is the maximum force that the muscle can generate per unit area. In this study, we set $\sigma = 50(\text{N}/\text{cm}^2)$ [12] and use the values shown in Table 1 for the cross-sectional area of each muscle [11]. i is the identification number of the muscle.

When the muscle activation level is η_i , the maximum traction outputs of the waist joint, τ_w^{\max} , and the knee joint, τ_k^{\max} , that the muscle can generate with the posture at a given time is derived as

$$\begin{aligned} \tau_w^{\max} = & (\eta_{e1} F_{e1}^{\max} - \eta_{f1} F_{f1}^{\max}) r \\ & + (\eta_{e3} F_{e3}^{\max} - \eta_{f3} F_{f3}^{\max}) r, \end{aligned} \quad (6)$$

$$\begin{aligned} \tau_k^{\max} = & (\eta_{e2} F_{e2}^{\max} - \eta_{f2} F_{f2}^{\max}) r \\ & + (\eta_{e3} F_{e3}^{\max} - \eta_{f3} F_{f3}^{\max}) r, \end{aligned} \quad (7)$$

where r is the moment arm of each joint [14]. τ_w^{\max} and τ_k^{\max} change according to the relative position between muscles and frames, which means that they reflect the posture of the patient.

Using (2), (3), (6), and (7), we can derive the physical activity of the patient, μ_i , as (1). If the physical activity (1) is a large value compared with the maximum activity that the muscles can generate, then the load is evaluated as being heavy. Usually, the patient does not use their maximum power, and in this study, we set the threshold showing the capability of the patient as $\mu^{\max} = 40(\%)$, which is based on the opinions of the nursing specialists [12].

3 Assistance Control

3.1 System Overview

Figure 5a shows our proposed assistance robot. The system consists of a support pad with three DOFs and a walker. The support pad is activated by our new assistance manipulator, which has four parallel linkages [7]. The patient leans on the support pad and grasps the armrest while standing with assistance (see Fig. 1b). In general,

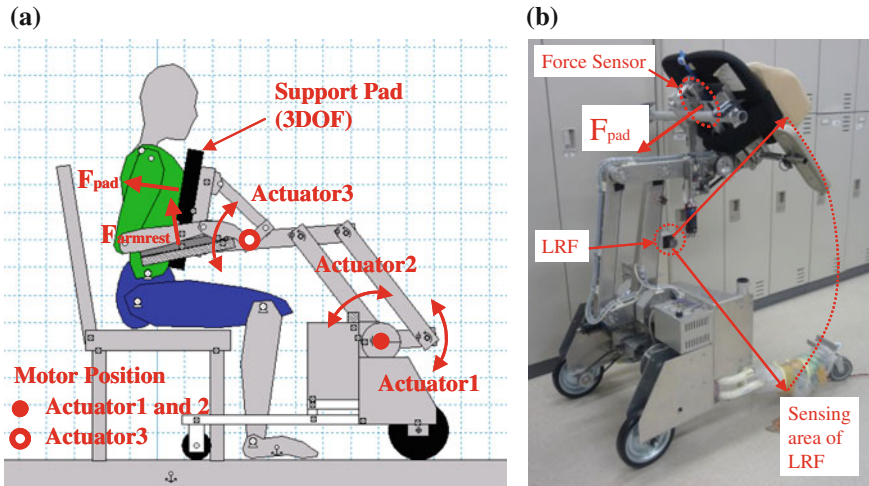


Fig. 5 Prototype of the assistive robot. a The robot’s actuators. b The robot’s sensors

fear of falling forward during the standing motion reduces elderly patients’ standing ability [15]. With the proposed scheme, patients can easily maintain their posture during a standing motion without the fear of falling forward.

Figure 5b shows the prototype of the proposed robot. The prototype is able to lift patients up to 180-cm tall and weighing up to 150 kg. Furthermore, because of its actuated wheels, the prototype can assist patients in walking. To measure a patient’s posture, the prototype has a force sensor and a laser range finder in its body (see Fig. 5b).

Our physical activity estimate scheme, which is proposed in the previous section, requires real-time data regarding its assistance force and the patient’s posture. To measure its assistance force, our support pad has two force sensors on its body that measure F_{pad} and $F_{armrest}$ (see Fig. 5b). To measure the patient’s posture, we use a laser range finder; thus, special markers do not need to be stuck onto the patient for a motion capture systems.

3.2 Standing Motion as Recommended by Nursing Specialists

Previous studies have proposed many types of assisted standing. Based on Kamiya’s experience as a nursing specialist, she proposed using the patient’s maximum strength to stand, as shown in Fig. 6. For effective standing assistance, we use a control reference, as shown in Fig. 7 [16]. Figure 7a shows the support pad’s position tracks, and Fig. 7b shows its angle tracks. The movement pattern in Fig. 7b refers to the ratio of the standing motion as determined by (8);

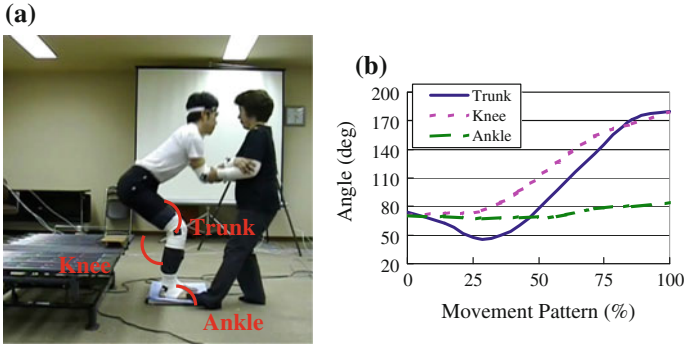


Fig. 6 Standing motion recommended by nursing specialists. **a** Standing motion. **b** Angular value of each joint

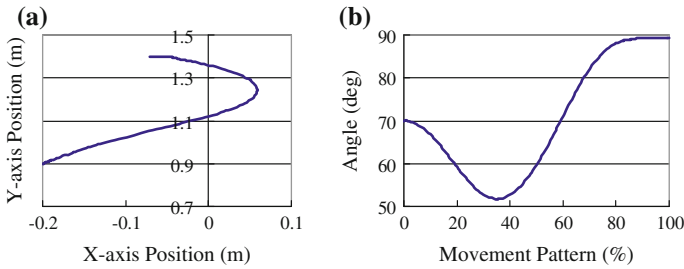


Fig. 7 Derived control references. The coordination of mechanisms is defined in Fig. 1a. **a** Position of $P(x_p, y_p)$. **b** Inclination θ_p

$$\hat{s} = \frac{t}{t_s}. \quad (8)$$

Here, t_s is the time required to complete the standing operation, and t is the present time.

3.3 Assistance Control Scheme Based on Physical Activity

In order to use the remaining physical strength of a patient, our assistance system uses a novel combination of damping and position control [17]. Damping control is suitable for controlling objects with contact. From (2) and (3), the assistance force, $F_y (= f_{yarmrest} + f_{ypad})$, in the lifting direction will reduce the required traction of each joint (τ_w^{req} and τ_k^{req}) because the coefficients of F_y , $-(x_s - x_w)$, and $(x_w - x_3)$ in (2) and $(x_w - x_k)$ in (3) will be negative in the usual standing posture. Therefore, we can expect that the damping control that increases F_y will reduce the required load of a patient during standing motion.

In our proposed control algorithm, if the physical activity of the patient is great, our system uses the damping control to reduce the patient's load. On the other hand, if the activity of the patient is small, our system uses the position control, which does not assist the force, but uses the remaining physical strength of the patient. In our previous works, our system used joint traction as an index of the patient's load for this algorithm [17]. In this paper, we extended our assistance algorithm using a proposed index of the patient's physical activity defined in (1).

Furthermore, for practical use, our system equips two assistance modes. One is a load-reducing mode. In this mode, our system assists the patient's body to reduce the fixed rate of the physical strength of a knee joint. The patient can set this fixed rate according to their bodily situation and in this paper, and we call this fixed rate the assistance force ratio. This mode is suitable for patients whose required care levels are serious. The other is a rehabilitation mode. In this mode, our system assists the patient's body only when the required physical strength of the patient exceeds that of which they are capable. The patient can set any value as the threshold and this mode is suitable for patients who have enough dexterity to stand up and require limited assistance.

3.3.1 Deriving the Reference

Before using the robot for assistance, we measure the height and mass of each patient individually. The length of each body segment is derived based on Table 2 and used by the reference generator as it derives the velocity control reference (9) of each actuator (Nos. 1, 2, and 3) from the motion reference (shown in Fig. 6) using the following equation:

$$\mathbf{v}_i^{ref} = \left[v_i^{ref}(0), \dots, v_i^{ref}(\hat{s}), \dots, v_i^{ref}(1) \right]^T. \quad (9)$$

Here, v_i^{ref} is the velocity control reference ($i = 1, 2, 3$), which is a function of the movement pattern \hat{s} defined in (8). For more details regarding the calculation process, please refer to our previous work [17].

3.3.2 Control Algorithm

Our system estimates the physical activity of the patient using the proposed scheme (1) while assisting patients as they stand. Based on this estimate, the system selects a suitable control scheme for damping and position controls. For this to happen, the output of each actuator is derived using

$$v_i = v_i^{ref} - B(F_y - F_{y0}) - K(x_i - x_i^{ref}), \quad (10)$$

where $F_y (= f_{yarmrest} + f_{ypad})$ is the force applied to the vertical direction on the support pad and armrest. x_i^{ref} is the angular position reference derived from (9), and x_i is the actual angular position. v_i is the updated reference value that our system inputs to the motor controller during the assisted standing motion. F_{y0} is the coefficient and force that the patient applies to the support pad while he or she stands. Using (10), our system can switch between the position control mode and the damping control mode.

3.3.3 Controller's Parameter Coordination

B and K are constants used to coordinate the ratio between the damping and position controls. Our system applies the damping control mode when the estimated physical activity of each joint μ_i , which is defined in (1), exceeds the threshold, μ^{max} . μ^{max} is derived in (11) in the case a in the load-reducing mode:

$$\mu^{max} = \max \{ \mu_k, \dots, \mu_j \}. \quad (11)$$

In the case of the rehabilitation mode, the patient may set the coefficient according to their own body situation:

$$\mu_i^{max} = (1 - r) \mu_i; \quad (12)$$

where r ($0 \leq r \leq 1$) is the assistance force ratio, which the patient can set.

To apply the damping control mode when the estimated physical activation, μ_i , exceeds μ^{max} , the coefficient B that validates the damping control mode is derived in (13):

$$\begin{cases} B = b (\mu_i - \mu^{max}) & \text{if } (\mu_i \geq \mu^{max}) \\ B = 0 & \text{if } (\mu_i < \mu^{max}) \end{cases} \quad (13)$$

On the other hand, the position control mode is always useful because it helps the patient to maintain stable posture during motion. Therefore, we set the coefficient, K , which validates the position control mode to a constant. Please note that the values of b and K are derived experimentally.

4 Experiments

4.1 Experimental Setup

To verify the effectiveness of our proposed scheme, eight subjects test the prototype robot based on the proposed estimate scheme. Two subjects (Subjects A and B) are young students and four subjects (Subjects C–F) are 54–72 years old with care levels of 1 or 2. Two subjects (Subjects G and H) are hemiplegics aged 32 and 64 years.

The young subjects (Subjects A and B) wear special clothing designed to limit their motion in order to simulate an elderly person’s limited mobility [18].

Unless otherwise noted, each subject tests the following three cases five times. In Case 1, the robot assists with the standing motion using only the position control mode. Only subjects A and B test this case because the robot does not assist with force and the subject is required to stand using only their own physical strength. In Case 2, the robot assists the subject using our proposed scheme. In this case, the robot uses the force control mode when the subject’s physical activity exceeds their capability threshold. In case 2A (rehabilitation mode), we set the assistance ratio to 30(%), and in case 2B (rehabilitation mode) we set the threshold of the subject’s capability to $\mu^{\max} = 40\%$ based on the opinion of nursing specialists [12]. In Case 3, the robot assists the subject with the force control mode as necessary, similar to Case 2. The difference between Cases 2 and 3 is that in Case 3, the robot estimates the physical activity of the subject using joint traction, as in our previous work [17]. In this case, we set the threshold of the subject’s capability as $\tau_{prev}^{\max} = 0.5(\text{Nm/kg})$ based on previous research [19].

In all cases, we use the standing motion recommended by nursing specialists [16], as specified in Sect. 3.2.

4.2 Experimental Results

The subject stands up as shown in Fig. 8. Figure 9 shows the required traction, τ_i^{req} , the maximum traction, τ_i^{\max} (defined in (2), (3), (6), and (7)), and the estimated physical activity of the subject, μ_i (defined in (1)) for each joint. As Fig. 8 shows, there are different tendencies between τ_i^{\max} and μ_i . The estimated load μ_i increases—especially at 40–75% movement in a knee joint, around which time the subject lifts their upper body and their load tends to be heavy. This result is similar to the experiences of nursing specialists [8].

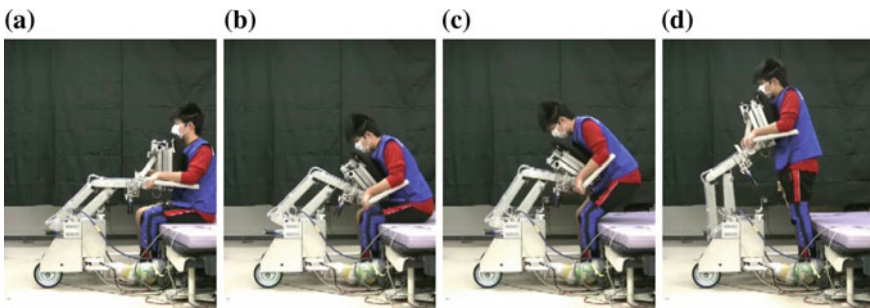


Fig. 8 Standing motion with our assistance robot (Case 1, Subject A). **a** 0(%). **b** 30(%). **c** 60(%). **d** 100(%)

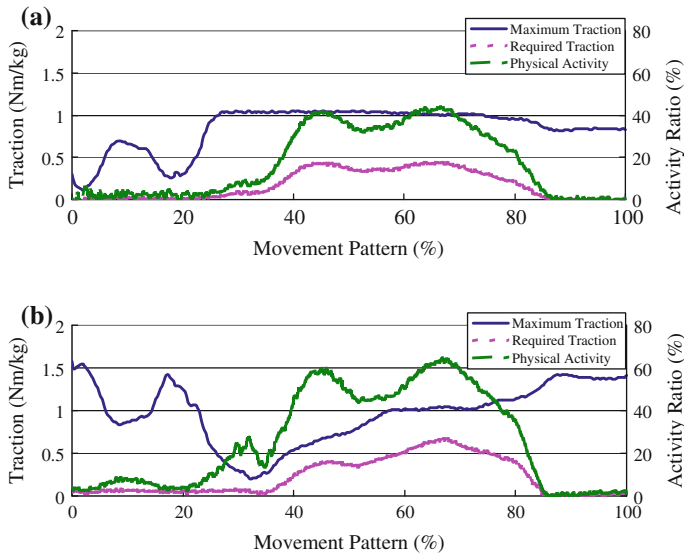


Fig. 9 A required traction, a maximum traction, and the estimated physical activity. (Case 1, Subject A). **a** A waist joint. **b** A knee joint

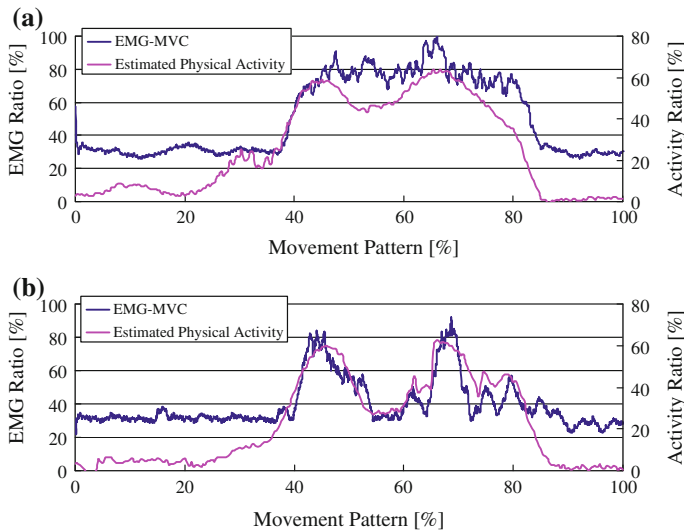


Fig. 10 The estimated physical activity and the measured muscle activity during a standing motion (Subject A). **a** Case 1 (without force assistance). **b** Case 2 (with force assistance)

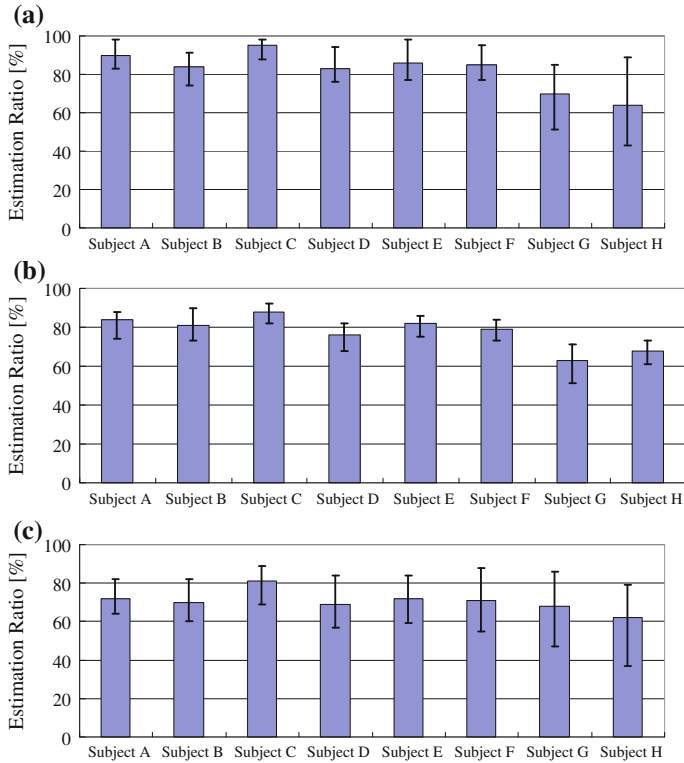


Fig. 11 Correct estimate ratio of the physical activity. **a** Case 2A (with proposed estimate, rehabilitation mode). **b** Case 2B (with proposed estimate, load-reducing mode). **c** Case 3 (with previous scheme)

Furthermore, Fig. 10 shows the EMG data of a vastus lateralis (VAS) muscle that is normalized by maximum voluntary contraction. This data reflects the activity of the knee joint. The activity of the VAS muscle in Fig. 10a has the same tendency as our proposed load-estimate index. In Fig. 10b, the estimated load exceeds the threshold ($\mu^{\max} = 40(\%)$), and our robot assists with force for the standing motion. Therefore, the load of the subject decreases during the knees' 40–75% movement. These results show that our proposed load estimate scheme is effective.

Figure 11 shows the ratio, ρ , which shows the correct answer rate of the estimated physical activity from (14):

$$\rho = \frac{t_{match}}{t_s}, \quad (14)$$

where t_s is the time required to complete the standing operation and t_{match} is the time at which the estimated physical activity exceeds the threshold μ^{\max} and the measured muscle activity exceeds this threshold as well.

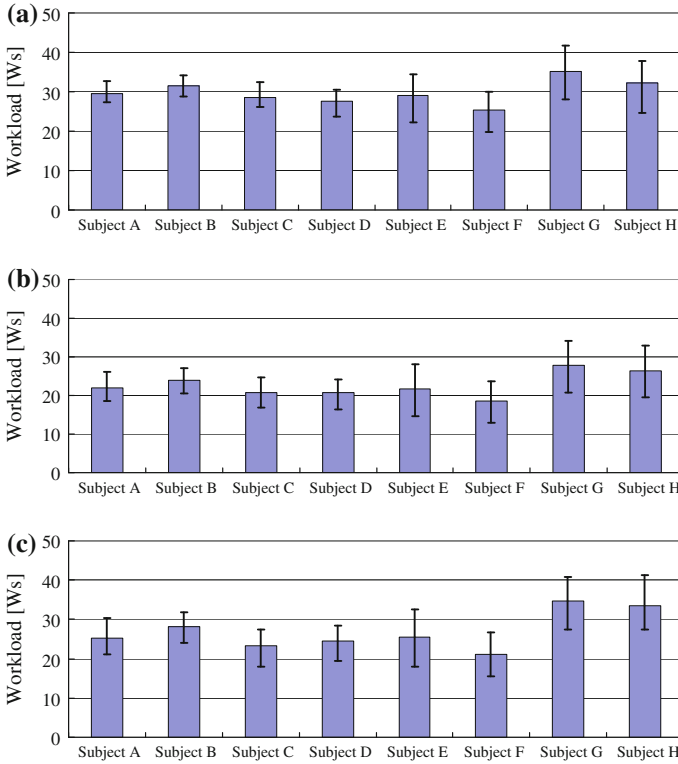


Fig. 12 Workload of the knee joint. **a** Case 2A (with proposed estimate, rehabilitation mode). **b** Case 2B (with proposed estimate, load-reducing mode). **c** Case 3 (with previous scheme)

In Case 2A (Fig. 11a), our system uses the rehabilitation mode and the proposed activity ratio $\mu^{\max} = 40(\%)$ as an index of high physical activity; in Case 2B (Fig. 11b), our system uses the load-reducing mode with assistance ratio $r = 0.3$ as the index; in Case 3 (Fig. 11c), our system uses joint traction $\tau_{prev}^{\max} = 0.5(\text{Nm/kg})$ as the index.

These results show that our proposed physical activity estimate scheme (Cases 2A and 2B) is more accurate than the previous index using joint traction (Case 3). Two subjects (Subjects G and H) are hemiplegics and the estimate results for both cases are inaccurate because their standing motions were different from the motion recommended by nursing specialists [16]; therefore, different muscles may be used when they stand up. Future work will discuss the muscle model for hemiplegics.

Using the estimated physical activity of the subject, our robot assists with force control only when necessary under the rehabilitation mode in Case 2A and the load-reducing mode in Case 2B. As a result, Fig. 12 shows the maximum traction output, τ_{knee}^{req} (peak load), which the subject is required to output to stand completely and Fig. 13 shows the required output power for one standing motion of a knee joint.

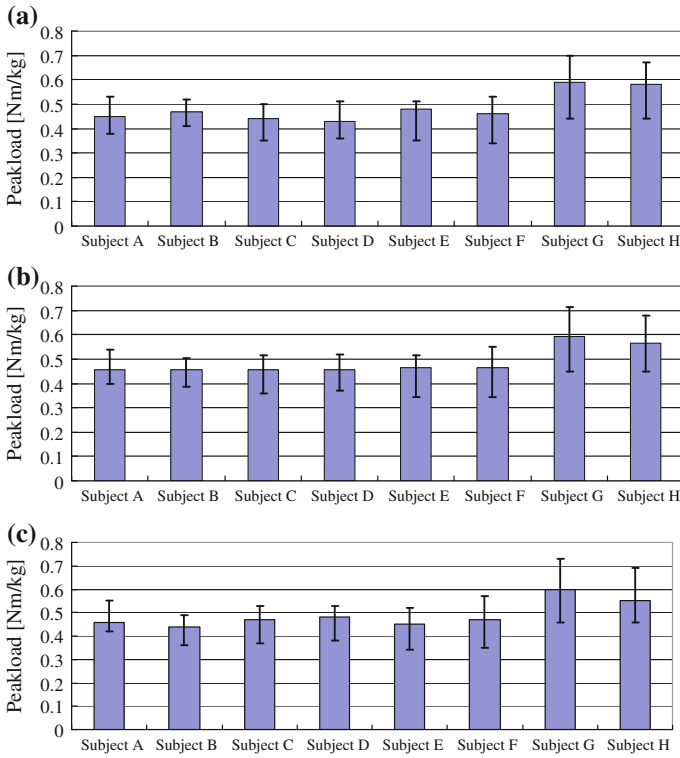


Fig. 13 Peak load of the knee joint. **a** Case 2A (with proposed estimate, rehabilitation mode). **b** Case 2B (with proposed estimate, load-reducing mode). **c** Case 3 (with previous scheme)

From Fig. 12a, c, we see that the workload in Case 2A is larger than that in Case 3, which means that the subject uses more physical strength in the rehabilitation mode with our proposed load-estimate (Case 2A). Furthermore, from Fig. 12b, c, we see that the workload in Case 2B is smaller than that in Case 3, which means that our system assists more efficiently in the load-reducing mode under our proposed load-estimate scheme (Case 2B).

On the other hand, from Fig. 13a–c, we see that the peak load is almost the same and does not exceed the capability of the subject, $\tau_{prev}^{max} = 0.5(Nm/kg)$, which means that our robot assists with enough force when necessary. These results show that our proposed load-estimate method allows the robot to assist with standing in such a way that the subject’s remaining physical strength is used as much as possible.

Finally, to investigate whether subjects find our assistance scheme oppressive, we administer a questionnaire survey to all subjects, as shown in Fig. 14. This Japanese questionnaire form is proposed by [20]. Figure 15 shows the questionnaire results. From Fig. 15, we find that, using our proposed idea, the feeling of oppression arising from a standing assistance system seems to be reduced. This means that our assistance

| | | | | | | | | |
|---|----------|-----|-----|-----|-----|-----|---|---------|
| ① | 不安な | ← 1 | — 2 | — 3 | — 4 | — 5 | → | 安心な |
| ② | 不快な | ← 1 | — 2 | — 3 | — 4 | — 5 | → | 快適な |
| ③ | 親しみを感じない | ← 1 | — 2 | — 3 | — 4 | — 5 | → | 親しみを感じる |
| ④ | 頼もしくない | ← 1 | — 2 | — 3 | — 4 | — 5 | → | 頼もしい |
| ⑤ | 恐怖を感じる | ← 1 | — 2 | — 3 | — 4 | — 5 | → | 恐怖を感じない |
| ⑥ | 違和感を感じる | ← 1 | — 2 | — 3 | — 4 | — 5 | → | 違和感がない |

Fig. 14 Questionnaire form. [不安な] means uneasy and [安心な] means safe. [不快な] means uncomfortable and [快適な] means comfortable. [親しみを感じない] means the subject does not feel familiarity and [親しみを感じる] means he feels it. [頼もしくない] means the subject feels the robot is not reliable and [頼もしい] means he feels it is reliable. [恐怖を感じる] means the subject feels fear and [恐怖を感じない] means he does not feel it. [違和感を感じる] means the subject feels incongruity and [違和感を感じない] means he does not feel it

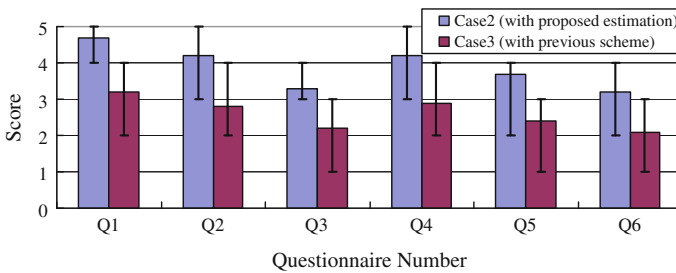


Fig. 15 Questionnaire results of the 8 subjects (A–H)

fits the condition of using a subject’s physical strength, and that subject does not feel fear of falling or other feelings of oppression. Thus, the proposed system succeeds in leading the standing motion by the subject.

5 Conclusions

This paper proposes both a physical activity estimate scheme that considers muscle arrangements and a novel assistance system that uses results of such estimates to take advantage of a patient’s remaining physical strength in such a way as to prevent their muscular strength from declining over time. By using our proposed scheme, our system can reduce a patient’s load when the patient’s posture is such that it is difficult to use any of the patient’s own physical strength.

In our system, the subject is required to set parameters, such as the cross-sectional area of each muscle. Previous researchers have proposed a derivation method of these values using easy gymnastics [11]. We plan to develop an automatic individual parameter derivation scheme in future work.

Acknowledgments This research is supported in part by Grant-in-Aid for Scientific Research C (25350693) from the Japan Society for the Promotion of Science (JSPS).

References

1. Alexander, N.B., Schultz, A.B., Warwick, D.N.: Rising from a chair: effects of age and functional ability on performance biomechanics. *J. Geom.: Med. Sci.* **46**(3), M91–98 (1991)
2. Hughes, M.A., Schenkman, M.L.: Chair rise strategy in the functionally impaired elderly. *J. Rehabil. Res. Dev.* **33**(4), 409–412 (1996)
3. Cabinet Office, Government of Japan, KOUREISHA HAKUSHO (The whitepaper on the aged society), p. 25 (2011). ISBN: 4904681010 (in Japanese)
4. Nagai, K., Nakanishi, I., Hanabusa, H.: Assistance of self-transfer of patients using a power-assisting device. In: *Proceedings of the IEEE International Conference on Robotics and Automation*, pp. 4008–4015 (2003)
5. Funakubo, A., Tanishiro, H., Fukui, Y.: Power assist system for transfer aid. *J. Soc. Instrum. Control Eng.* **40**(5), 391–395 (2001)
6. Hirvensalo, M., Rantanen, T., Heikkinen, E.: Mobility difficulties and physical activity as predictors of mortality and loss of independence in the community-living older population. *J. Am. Geriatr. Soc.* **48**, 493–498 (2000)
7. Chugo, D., Morita, Y., Sakaida, Y., Yokota, S., Kobayashi, H., Hashimoto, H., Takase, K.: Standing assistance control using a physical strength of a patient with load estimation. In: *Proceedings of 21st IEEE International Symposium on Robot and Human Interactive Communication*, pp. 234–239 (2012)
8. Nuzik, S., Lamb, R., Vansant, A., Hirt, S.: Sit-to-stand movement pattern. A kinematic study. *Phys. Ther.* **66**(11), 1708–1713 (1986)
9. Hatsukari, T., Kuroko, S., Miyake, N., Kawazoe, R., Higuchi, J., Hirata, Y., Kosuge, K.: Self-help standing-up method based on quasi-static motion. In: *Proceedings of the IEEE International Conference on Robotics and Biomimetics*, pp. 342–347 (2009)
10. Nishida, I., Maeda, M., Kawano, T., Shirase, K.: Estimation method of muscle forces of lower limb considering the role of antagonistic muscles and biarticular muscles-estimation of muscle forces of lower limb during vertical jumping. *J. Jpn. Ergonomics Soc.* **47**(6), 244–251 (2011)
11. Okada, H., Ae, M., Fujii, N., Morioka, Y.: Body segment inertia properties of Japanese elderly. *Biomechanisms* **13**, 125–139 (1996)
12. Oshima, T., Fujikawa, T., Kumamoto, M.: Functional evaluation of effective muscle strength based on a muscle coordinate system consisted of bi-articular and mono-articular muscles-contraction forces and output forces of human limbs. *J. Precis. Eng.* **65**(12), 1772–1777 (1999)
13. Spector, S.A., Gardiner, P.F., Zernicke, R.F., Roy, R.R., Edgerton, V.R.: Muscle architecture and force-velocity characteristics of cat soleus and medial gastrocnemius: implications for neural control. *J. Neuro-physiol.* **44**, 951–960 (1980)
14. Hoy, M.G., Zajac, F.E., Gordon, M.E.: A musculoskeletal model of the human lower extremity: The effect of muscle, tendon, and moment arm on the moment-angle relationship of musculotendon actuators at the hip, knee, and ankle. *J. Biomech.* **23**(2), 157–169 (1990)
15. Maki, E., Holliday, P.J., Topper, A.K.: Fear of falling and postural performance in the elderly. *J. Gerontol.* **46**(4), 123–131 (1991)
16. Kamiya, K.: Development and evaluation of life support technology in nursing. In: *Proceedings of 7th RACE Symposium Research into Intelligent Artifacts for the Generalization of Engineering*, pp. 116–121 (2005)
17. Chugo, D., Matsuoka, W., Songmin, J., Takase, K.: Rehabilitation walker with standing-assistance device. *J. Robot. Mechatronics* **19**(6), 604–611 (2007)
18. Takeda, K., Kanemitsu, Y., Futouy, Y.: Understanding the problem of the elderly through a simulation experience-difference in the effect between before and after clinical practice. *Kawasaki Medical Welfare J.* **11**(1), 64–73 (2001)

19. Omori, K., Yamazaki, Y., Yokoyama, H., Aoki, U., Kasahara, M., Hiraki, K.: The relationship between strength in the lower extremity and the ability to stand up from a chair in elderly inpatients. *Sogo Rehabil.* **30**(2), 167–171 (2001)
20. Matsui, Y., Kanoh, M., Kato, S., Itoh, H.: Generating interactive facial expressions of kansei robots using simple recurrent network. *J. Robot. Soc. Jpn.* **28**(3), 360–368 (2010) (in Japanese)