Assistive Robot for Standing with Physical Activity Estimation based on Muscle Arrangements of Human Legs

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Keywords: Standing Assistance, Musculoskeletal Model, Joint Traction, Physical Activity Estimation.

Abstract: A physical activity estimation scheme is proposed for patients who use a robot for standing assistance. In general, conventional assistive robots do not require patients to use their own physical strength to stand, which leads to decreased strength of the elderly. Therefore, an assistive robot that maximally uses a patient's remaining physical strength is desired. The assistive robots can achieve this objective by estimating the physical activity of the patient when they stand. The activity estimation proposed here is primarily 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, which are normally stuck on patients. The proposed estimation is implemented with a prototype assistive robot that assists elderly patients to use their remaining physical strength based on the estimated results, thus testing the effectiveness of the proposed method.

1 INTRODUCTION

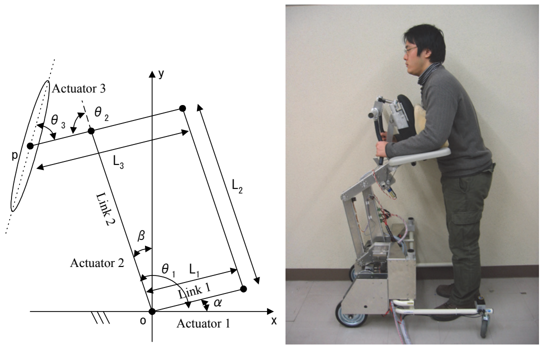
The act of standing may be the most serious and important activity in the daily life of an elderly person lacking physical strength (Alexander et al., 1999; Hughes et al., 1996). However, assisting elderly patients to stand is a heavy task for caregivers and this can be the primary source of the lumbago that many experience (Cabinet Office, Government of Japan, 2011). Therefore, creating a care service robot capable of assisting the elderly when they stand is important, and thus many such assistive devices have been developed and presented in previous works (Nagai et al., 2003; Funakubo et al., 2001).

In Japan, elderly people requiring assistance in daily life are classified into five different care levels (Cabinet Office, Government of Japan, 2011), where requiring care level 1 is a minor and requiring care level 5 is a serious condition. Generally, the elderly 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 use the patient's remaining physical strength. Thus, the patient's physical strength decreases (Hirvensalo et al., 2000). In fact, between 2002 and 2003, more than 10% of care level 1 patients were subsequently assigned to higher care levels in next year (Cabinet Office, Government of Japan, 2011). 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 (Chugo et al., 2012). The robot is based on a walker (a popular assistance device for aged people in normal daily life) and uses a support pad, which is 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 because the robot is required to coordinate its assistive force

Chugo D., Yamada T., Muramatsu S., Yokota S. and Hashimoto H..
Assistive Robot for Standing with Physical Activity Estimation based on Muscle Arrangements of Human Legs.
DOI: 10.5220/0005527400350043
In *Proceedings of the 12th International Conference on Informatics in Control, Automation and Robotics* (ICINCO-2015), pages 35-43
ISBN: 978-989-758-123-6



(a) Flame kinematic model. (b) Overview of our robot.

Figure 1: Our developed robot for standing assistance.

accordingly. However, generally, such estimations without biosensors, as electromyographs (EMG), are difficult; further, physical activity estimation with biosensors, which are required to be stuck on the patient, is impractical because assistance robots should be low cost and easy to use.

Previous works have proposed physical activity estimation using human models comprising linkages and joints without such biosensors (Nuzik et al., 1986; Hatsukari et al., 2009). These schemes evaluate the patient's physical activity using joint traction, which is calculated using the kinematical model as an index. However, many muscles generate human body movements, and traction, which muscles can generate maximally, changes according to the relative positions of bones and muscles. Therefore, maximum joint traction is not constant but it changes according to the patient's posture. During a standing motion, the 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 real-time physical activity estimation for patients using a standing assistance robot without additional biosensors. The paper is organized as follows: in Section 2, we propose an estimation scheme of a patient's activity according to their posture during the standing motion using a human musculoskeletal model of a lower limb, which expresses a biarticular muscle function; in Section 3, we demonstrate an assistance control scheme on our robot, which uses a patient's strength based on estimated results; in Section 4, we provide experimental results obtained using our prototype; and Section 5 concludes this paper.

2 PHYSICAL ACTIVITY ESTIMATION

2.1 Overview of Proposed Estimation Scheme

In the linkage model of a human (Nuzik et al., 1986; Hatsukari et al., 2009), a joint traction is used as an index of a patient's load. However, a joint traction does not consider the posture of the patient, and in some cases, this index and the experience of nursing specialists are different, especially when the patient is in a half-sitting posture. When the patient stands, the muscles shown in Fig. 2 generate the lifting motion (Nishida et al., 2011). Many muscles (shown in Table 1) are used to accomplish the standing movement, and the traction, which muscles can generate maximally, changes according to the relative position between frames and muscles.

Thus, we propose a novel physical activity estimation 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 the 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 $\tau_k^{\max}$ and waist joint $\tau_k^{\max}$) the muscles can generate for the posture at this time.
- Comparing the two derived tractions, we evaluate the physical activity of patient μ_i, which demonstrates how much the patient uses their own physical 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 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 (Nuzik et al., 1986). Using this model, we can derive the traction of each joint and estimate the patient's load.

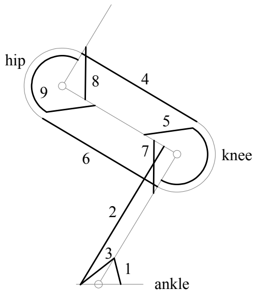


Figure 2: Muscle arrangements in the human leg.

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	Semimenbranosus (SM)	60.2
7	Biceps femoris and short head (BFSH)	8.7
8	Iliopsoas (IL)	23.0
9	Gluteus masimum (GMAX)	20.0

The assistance system is designed in such a way that patients lean on the pad and grasp the 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).

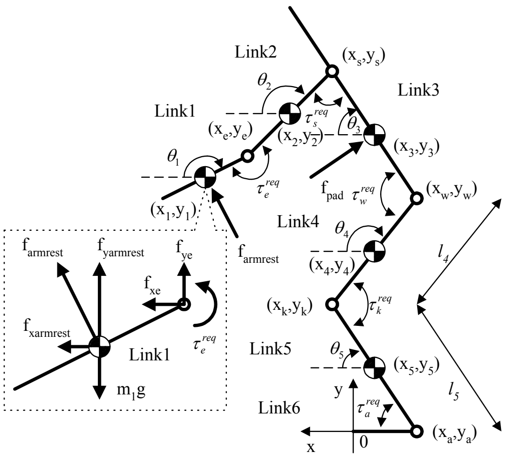


Figure 3: Linkage model of a human.

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 mass of the link ($i = 1, \cdots, 6$) 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.

Using the balance of applied force and its moment, we can derive the required traction of each joint as (7) and (8). Our robot measures the user's posture using the kinematical information provided by the assistance manipulator and a laser range finder, which is equipped as shown in Fig. 1(b), without additional biosensors that would have to be stuck onto the patient. Using these equations, we can calculate the required traction for each joint (τ_k^{req}, τ_k^{req}). For further information, please refer to our previous paper (Chugo et al., 2012).

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

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

Here, we use body parameters chosen from a standard body of data of Japanese adult males (Okada et al., 1996); 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.

2.3 Deriving the Maximum Traction

In the field of biomedicine, a human musculoskeletal model that considers the role of both an antagonistic muscle and a biarticular muscle has been proposed (see Fig. 4) (Oshima et al., 1999). This model shows that the antagonistic muscle and the biarticular muscle interact to generate human body movement.

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 mass ratio of the body segment to the mass 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.

We know from previous research (Oshima et al., 1999) that when the maximum force each muscle can realize at the ankle joint is F_{me1}, F_{me2}, F_{me3}, F_{mf1}, F_{mf2}, and F_{mf3}, 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 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 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 estimation scheme of 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 (2) and (3), respectively. From the kinematical relationship shown in Fig. 4, the force output vector (f_x, f_y) at the ankle joint is derived:

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

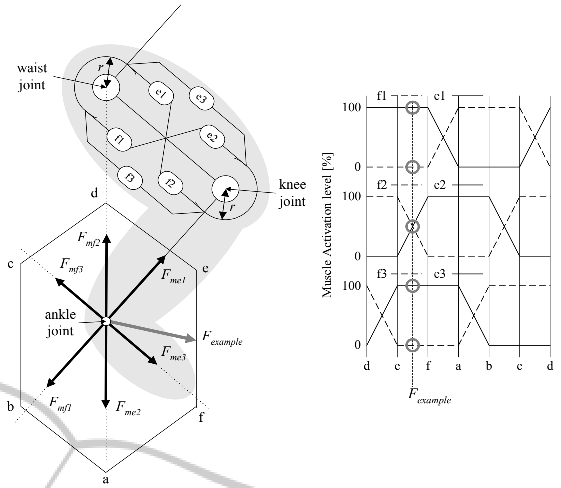


Figure 4: Musculoskeletal model considering the role of the antagonistic and biarticular muscles.

Second, our system derives the distribution of the output force at the ankle joint from the patient's posture. Then, our system 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 (Spector et al., 1980) that the maximum force $F_i^{\max}$ that the 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 a maximum force that the muscle per unit area can generate. In this study, we set $\sigma = 50[N/cm^2]$ (Oshima et al., 1999) and use the values shown in Table 1 for a cross-sectional area of each muscle (Okada et al., 1996). i is an identification number of the muscle.

When the muscle activation level is η_i, the maximum traction output of the waist joint $\tau_w^{\max}$ and the knee joint $\tau_k^{\max}$ that the muscle can generate with the posture at this time is derived as

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

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

where r is a moment arm of each joint (Hoy et al., 1990). $\tau_w^{\max}$ and $\tau_k^{\max}$ change according to the relative position between muscles and frames, which means that it reflects 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 we can evaluate the load is 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 nursing specialists (Oshima et al., 1999).

3 ASSISTANCE CONTROL

3.1 System Overview

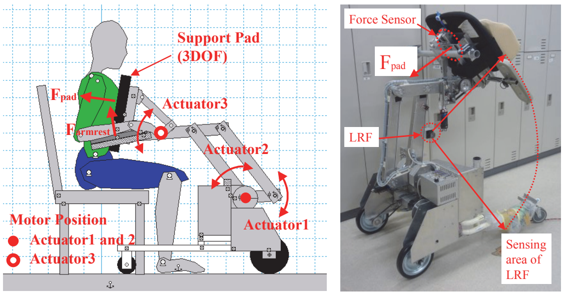
Fig. 5(a) shows our proposed assistance robot. The system consists of a support pad with three DOF and the walker. The support pad is activated by our new assistance manipulator, which has four parallel linkages (Chugo et al., 2012). The patient leans on the support pad and grasps the armrest while standing with assistance (see Fig. 1(b)). In general, fear of falling forward during the standing motion reduces elderly patients' standing ability (Maki et al., 1991). With the proposed scheme, patients can easily maintain their posture during the standing motion without the fear of falling forward.

Fig. 5(b) 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 walk. To measure a patient's posture, the prototype has a force sensor and a laser range finder in its body (see Fig. 5(b)).

Our physical activity estimation 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. 5(b)). To measure the patient's posture, we use a laser range finder; thus, calibration or special markers are not required to be stuck on the patient.

3.2 Standing Motion as Recommended by Nursing Specialists

Previous studies have proposed many types of assisted standing. Based on her experience as a nursing specialist, Kamiya proposed using the

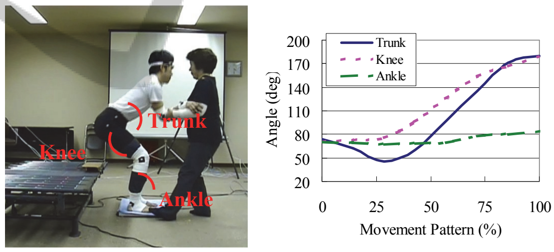


(a) The robot's actuators. (b) The robot's sensors.

Figure 5: Prototype of the assistive robot.

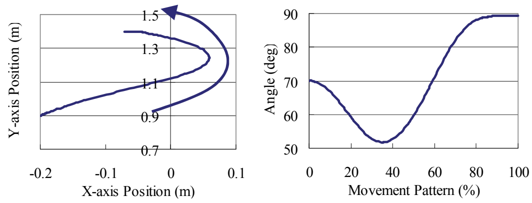
patient's maximum strength to stand, as shown in Fig. 6. For effective standing assistance, we use a control reference as shown in Fig. 7 (Kamiya, 2005). Fig. 7(a) shows the support pad's position tracks, and Fig. 7(b) shows its angle tracks. The movement pattern in Fig. 7(b) refers to a ratio of the standing motion as determined by (8). t_s is the time required to complete the standing operation, and t is the present time.

$$\hat{s} = \frac{t}{t_s} \tag{8}$$



(a) Standing motion. (b) Angular value of each joint.

Figure 6: Standing motion recommended by nursing specialists.



(a) Position of P (x_p, y_p) (b) Inclination θ_p

Figure 7: Derived control references. The coordination is defined in Fig. 1(a).

3.3 Assistance Control Scheme based on the Physical Activity

For using the remaining physical strength of a patient, our assistance system uses new control that combines damping control and position control (Chugo et al., 2007). Damping control is suitable for controlling objects with contact. From (2) and (3), the assistance force $F_y\left(= f_{yarmrest} + f_{ypad}\right)$ in the lifting direction will reduce the required traction of each joint (τ_w^{req} and τ_k^{req}) because coefficients of F_y , $-\left(x_s - x_w\right)$ and $\left(x_w - x_3\right)$ in (2) and $\left(x_w - x_k\right)$ in (3) will be negative in usual standing posture. Therefore, we can expect that the damping control which 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 heavy, our system uses the damping control for reducing the patient's load. On the other hand, if the activity of the patient is light, our system uses the position control, which does not assist the force, for using the remaining physical strength of the patient. In our previous works, our system uses a joint traction as an index of the patient's load for this algorithm (Chugo et al., 2012). In this paper, we extended our assistance algorithm using a proposed index of the patient's physical activity defined in (1).

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 of each actuator (No. 1, 2, and 3) (9) from the motion reference (shown in Fig. 6) using the following equation:

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

where v_i^{ref} is the velocity control reference $\left(i = 1,2,3\right)$, 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 (Chugo et al., 2012).

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 estimation, the system selects a suitable control scheme for damping and position controls. For this to happen, the output of each actuator is derived from

$$v_i = v_i^{ref} - B\left(F_y - F_{y0}\right) - K\left(x_i - x_i^{ref}\right) \qquad (10)$$

where $F_y\left(= f_{yarmrest} + f_{ypad}\right)$ is the applied force 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 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 that coordinate the ratio between the damping and position controls. Our system applies damping control when the maximum estimated load of each joint μ_i , which is defined in (1), exceeds the threshold $\mu^{\max} = 40[\%]$. *i* is the identification character. (For example, for the knee joint, *i* is *k*.) For applying damping control, the coefficient *B*, which validates damping control, is derived from

$$\begin{cases} B = b\left(\mu_i - \mu^{\max}\right) & if\ \left(\mu_i \ge \mu^{\max}\right) \\ B = 0 & if\ \left(\mu_i < \mu^{\max}\right) \end{cases} \qquad (11)$$

Using this parameter coordination, our system assists the patient with increased force when the patient's load is heavy. On the other hand, position control is always useful because it helps the patient maintain stable posture during the standing motion. Thus, we set coefficient *K,* which validates position control, as constant. In this study, the values *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, which is implemented on the basis of the proposed estimation 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 (Takeda et al., 2001).

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 with the force control mode when the subject's physical activity exceeds their capability threshold. In this case, the robot uses our proposed load estimation scheme, and we set the threshold of the subject's capability as $\mu^{\max} = 40\%$ based on the opinion of nursing specialists (Oshima et al., 1999). 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 (Chugo et al., 2012). In this case, we set the threshold of the subject's capability as $\tau_{prev}^{\max} = 0.5[Nm/kg]$ based on previous research (Omori et al., 2001).

In all cases, we use the standing motion recommended by nursing specialists (Kamiya, 2005) as specified in Section 3.2.

4.2 Experimental Results

The subject stands up as shown in Fig. 8. Fig. 9 shows the required traction τ_i^{req}, the maximum traction $\tau_i^{\max}$ (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 $\tau_i^{\max}$ 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 (Nuzik et al., 1986).

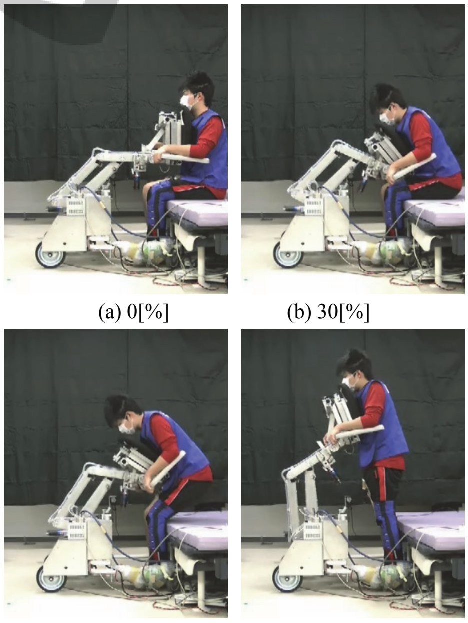
Furthermore, Fig. 10 shows the EMG data of a vastus lateralis (VAS) muscle that is normalized by maximum voluntary contraction. This data reflects the activity of the knee joint. The activity of the VAS muscle in Fig. 10(a) has the same tendency as our proposed load estimation index. In Fig. 10(b), the 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 estimation scheme is effective.

Fig. 11 shows the ratio ρ which shows the correct answer rate of the estimated physical activity from (12).

$$\rho = \frac{t_{match}}{t_s} \tag{12}$$

where t_s is the time required to complete the standing operation and t_{match} is the time when the estimated physical activity exceeds the threshold $\mu^{\max} = 40\%$ and the measured muscle activity exceeds 40%, too.

In Case 2 (Fig. 11(a)), our system uses the proposed activity ratio $\mu^{\max} = 40[\%]$ as an index of high physical activity; in Case 3 (Fig. 11(b)), our system uses joint traction $\tau_{prev}^{\max} = 0.5[Nm/kg]$ as the index. These results show that our proposed physical activity estimation scheme (Case 2) is more accurate than the previous index, which uses joint traction (Case 3). Two subjects (Subjects G and H) are hemiplegics and the estimation results for both cases are inaccurate because their standing motions were different from the motion recommended by nursing specialists (Kamiya, 2005); therefore, different muscles may be used when they stand up. Future work will discuss the muscle model for hemiplegics.

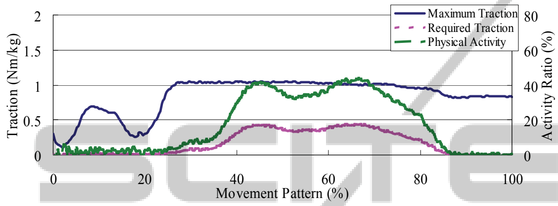


(a) 0[%] (b) 30[%]

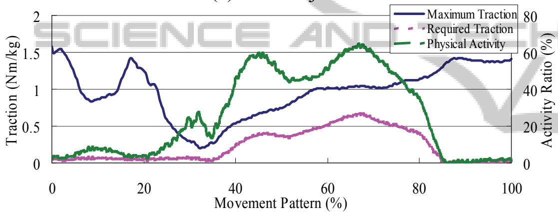
(c) 60[%] (d) 100[%]

Figure 8: Standing motion with our assistance robot (Case 1, Subject A).

Using the estimated physical activity of the subject, our robot assists with force control only when necessary. As a result, Fig. 12 shows the maximum traction output τ_{knee}^{req} (peak load) which the subject is required to output for standing completely and Fig. 13 shows the required output power for one standing motion of a knee joint. From Fig. 12(a) and (b), we see that the workload in Case 2 is larger than that in Case 3, which means that the subject uses more physical strength with our proposed load estimation (Case 2). On the other hand, from Fig. 13(a) and (b), 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.

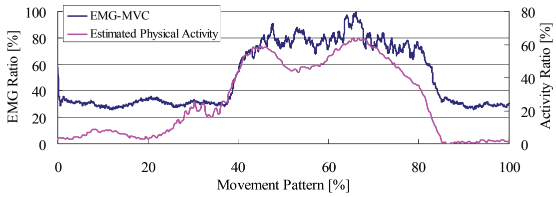


(a) A waist joint

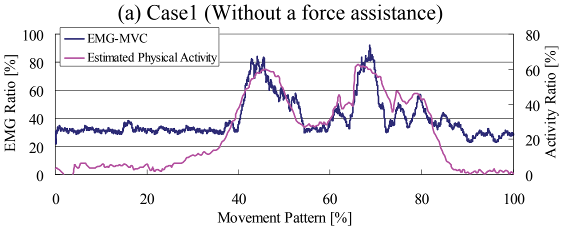


(b) A knee joint

Figure 9: A required traction, a maximum traction, and the estimated physical activity. (Case 1, Subject A).



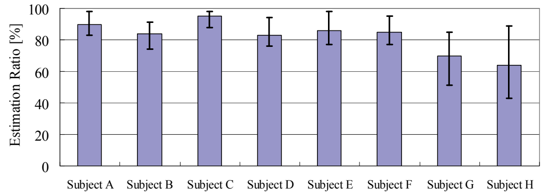
(a) Case1 (Without a force assistance)



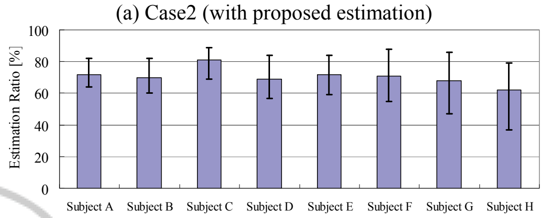
(b) Case2 (With a force assistance)

Figure 10: The estimated physical activity and the measured muscle activity during a standing motion. (Subject A).

These results show that our proposed load estimation allows the robot to assist with standing in such a way that the subject's remaining physical strength is used as much as possible.

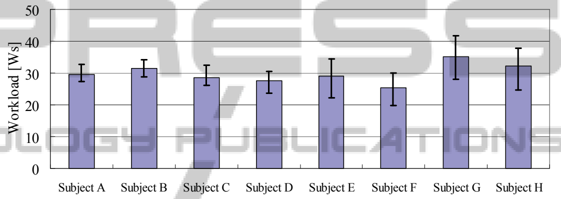


(a) Case2 (with proposed estimation)

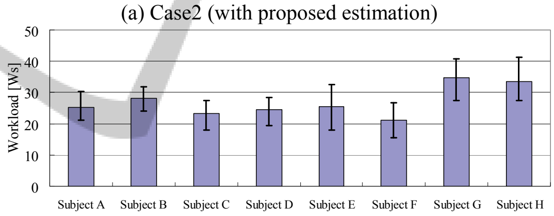


(b) Case 3 (with previous scheme)

Figure 11: Ratio of the estimated physical activity and the measured muscle activity.



(a) Case2 (with proposed estimation)



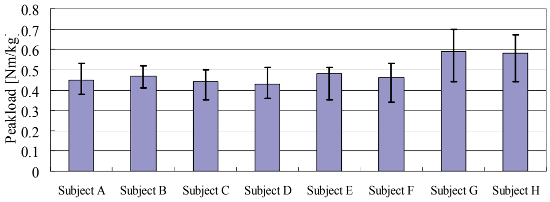
(b) Case 3 (with previous scheme)

Figure 12: Workload of the knee joint.

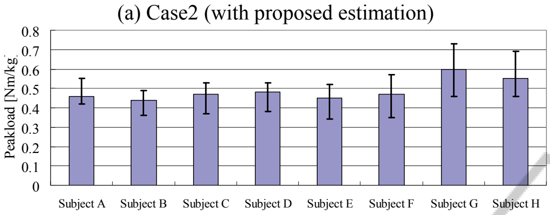
5 CONCLUSIONS

This paper proposes both a physical activity estimation scheme that considers muscle arrangements and a novel assistance system that uses those estimated results to take advantage of the patient's remaining physical strength in such a way that the patient's muscular strength will not decline 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 a cross-sectional area of each muscle. Previous researchers have proposed a derivation method of these values using easy gymnastics (Oshima et al., 1999). We plan to develop an automatic individual parameter derivation scheme in future work.



(a) Case2 (with proposed estimation)



(b) Case 3 (with previous scheme)

Figure 13: Peakload of the knee joint.

ACKNOWLEDGEMENTS

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

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