2014 IEEE International Conference on Robotics & Automation (ICRA)
Hong Kong Convention and Exhibition Center
May 31 - June 7, 2014. Hong Kong, China

Generation of Human Standing-up Motion with Muscle Synergies Using Forward Dynamic Simulation

Qi An[1], Yuki Ishikawa[1], Tetsuro Funato[2], Shinya Aoi[3], Hiroyuki Oka[4], Hiroshi Yamakawa[1], Atsushi Yamashita[1], and Hajime Asama[1]

Abstract— **The standing-up motion is one of the most important activities of daily livings. In order to understand the strategy to achieve the standing-up motion, muscle synergy analysis is applied to the measured data during human standing-up motion. In addition, musculoskeletal model which consists of three body segments and nine muscles in lower limb is developed to ensure that the standing-up motion can be generated by muscle synergies. As a result, three muscle synergies have been extracted from the human standing-up motion, and each synergy strongly corresponded to characteristic kinematic events: momentum flexion, momentum transfer, and posture stabilization. Results of forward dynamic simulation show that the standing-up motion can be achieved by controlling time-varying weighting coefficient of three muscle synergies instead of controlling individual nine muscles.**

I. Introduction

The number of the elderly people has increasing rapidly, and it has brought many serious issues to our society, such as decreased physical ability or increased social security cost. It is important to improve functional mobility of the elderly in order to avoid being bedridden and enhance their quality of life. Among the daily activities, especially the functional ability to perform the standing-up motion is an important criteria of activities of daily living [1].

We have developed the assistive device to support joint torque of the elderly people [2]. However, adding deficient joint force cannot fully improve the functional ability. In areas of medical or physical therapy, many training methodologies have been used to strengthen muscles, but it has been reported that increase of muscle strength is only seen in the same posture that people perform the training [3]. The study implies that training strongly depends on the environment or its context when the training is performed. Moreover, the importance of training multiple joints movement or muscles is pointed out for improvement of body function [4]. Therefore, it is necessary to understand how the standing-up motion is performed by clarifying the condition to achieve the standing-up motion or understanding the strategy to perform the motion in order to enhance the ability of it.

[1]Qi An, Yuki Ishikawa, Hiroshi Yamakawa, Atsushi Yamashita, and Hajime Asama are with Department of Precision Engineering, Graduate School of Engineering, The University of Tokyo, Tokyo, 1138656, Japan anqi@robot.t.u-tokyo.ac.jp

[2]Tetsuro Funato is with the Department of Mechanical Engineering and Intelligent Systems, The University of Electro-Communications, Tokyo, Japan

[3]Shinya Aoi is with the Department of Aeronautics and Astronautics, Graduate School of Engineering, Kyoto University, Kyoto, Japan

[4]Hiroyuki Oka is with the 22nd Century Medical and Research Center, Graduate School of Medicine, The University of Tokyo, Tokyo, Japan

Regarding the standing-up motion, many previous studies have focused on kinematic characteristics. For example, Shenkman et al. have divided the standing-up motion into four phases based on kinematic events: forward movement of shoulder (momentum flexion), rising hip (momentum transfer), minimum ankle doriflexion (extension), and maximum shoulder height (posture stabilization) [5].

In order to elucidate the mechanism of the standing-up motion, we have focused on the idea of muscle synergies. The idea of muscle synergy was firstly suggested by Bernstein [6] to decompose the complex human movement into small sets of modules of synchronized muscle activation (muscle synergy). Some previous studies have employed the muscle synergy analysis to show that basic movement of the frogs can be explained with common muscle synergies [7]. Additionally it has been shown that human locomotion can be achieved by five muscle synergies [8].

If there are muscle synergies in human standing-up motion, it would be useful knowledge for a training methodology. Also, contribution of muscle synergies to body kinematics should be clarified to elucidate the strategy of human standing-up motion. Previously, we have analyzed the standing-up motion with muscle synergies and their contribution was simulated from musculoskeletal model which is expressed as the neural network model [9]. Although this model could express relationship between muscle activation, joint torques, and body kinematics, it did not consider the effect of anatomical characteristics or dynamics of body.

The objectives of this study are to develop a musculoskeletal model which involves human body dynamics and anatomical characteristics of muscle and to clarify that human standing-up motion is generated from a small number of muscle synergies using forward dynamic simulation.

II. Methods

A. Synergy Model

This study assumes that muscle activation of human movement is generated from muscle synergies and time-varying weighting coefficients (eq. (1)).

$$\mathbf{A} \cong \mathbf{W}\mathbf{C}. \tag{1}$$

In the equation, $\mathbf{A}$ indicates the matrix of discrete time-varying activation of n muscles ($1 \leq t \leq T_{\max}$) as shown in eq. (2). Muscle synergy matrix $\mathbf{W}$ consists of N muscle synergy vector $\mathbf{w}_{j(j=1,2,\cdots,N)}$ and its component w_{ij} indicates i-th muscle activation of j-th muscle synergy (eq. (3)). Weighting coefficient matrix $\mathbf{C}$ is composed of the vectors

$\mathbf{c}_j$, and their components $c_j(t)$ indicate weighting coefficient of j-th synergy at time t (eq. (4)).

$$\mathbf{A} = \begin{pmatrix} \mathbf{a}_1(t) \\ \mathbf{a}_2(t) \\ \vdots \\ \mathbf{a}_n(t) \end{pmatrix} = \begin{pmatrix} a_1(1) & \cdots & a_1(T_{\max}) \\ \vdots & \ddots & \vdots \\ a_n(1) & \cdots & a_n(T_{\max}) \end{pmatrix}, \tag{2}$$

$$\mathbf{W} = (\mathbf{w}_1 \cdots \mathbf{w}_N) = \begin{pmatrix} w_{11} & \cdots & w_{1N} \\ \vdots & \ddots & \vdots \\ w_{n1} & \cdots & w_{nN} \end{pmatrix}, \tag{3}$$

$$\mathbf{C} = \begin{pmatrix} \mathbf{c}_1(t) \\ \mathbf{c}_2(t) \\ \vdots \\ \mathbf{c}_N(t) \end{pmatrix} = \begin{pmatrix} c_1(1) & \cdots & c_1(T_{\max}) \\ \vdots & \ddots & \vdots \\ c_N(1) & \cdots & c_N(T_{\max}) \end{pmatrix}. \tag{4}$$

Figure 1 shows a schematic design of muscle synergy model. It depicts that n muscle activation (Fig. 1 (b): gray part) is generated from muscle synergies and time-varying weighting coefficients. The bars in Fig. 1 (a) show simultaneous muscle activation in muscle synergies ($\mathbf{w}_{1,2,3}$) and corresponded time-varying weighting coefficients ($\mathbf{c}_{1,2,3}$). In Fig. 1 (b), dashed red, blue, and green lines show muscle activation generated from individual synergies.

Non-negative matrix factorization (NNMF) is used to decide muscle synergy matrix $\mathbf{W}$ and time-varying weighting coefficients matrix $\mathbf{C}$ [10]. In this study, different numbers of synergies are tested to decide the optimal synergy number to express muscle activation during standing-up motion.

One-factor analysis of variance (ANOVA) is conducted to evaluate how extracted muscle synergies ($\mathbf{WC}$) can explain muscle activation ($\mathbf{A}$). Mean squared error is used for evaluation. Significance level p is set to 0.05, and post-hoc test (Tukey-Kramer test) is applied to assess the effect of increase in the number of synergies if there is a significant difference among the number of synergies.

B. Musculoskeletal Model

This study focuses on sagittal movement of human body, and human body is expressed as two dimensional model with three solid links (shown in Fig. 2 (a)). In the standing-up motion, human barely moves their feet, and therefore their feet are fixed in our model. Each link respectively indicates shank, thigh, and HAT (head, arm, and trunk), and three joint angles ($\theta_{k=1,2,3}$) indicate ankle, knee, and hip joint angles from the horizontal direction to each link.

Muscle model employs nine muscles in lower limb including mono- and bi-articular muscles as described in Fig. 2 (b). All nine muscles are chosen based on their function to either extend or flex ankle, knee, or hip joints as shown below.

1) Tibialis Anterior (TA): dorsiflexes ankle
2) Gastrocnemius (GAS): plantarflexes ankle and flexes knee
3) Soleus (SOL): plantarflexes ankle
4) Rectus Femoris (RF): extends knee and flexes hip
5) Vastus (VAS): extends knee
6) Biceps Femoris Long Head (BFL): flexes knee and extends hip
7) Biceps Femoris Short Head (BFS): flexes knee
8) Illopsoas (IL): flexes hip
9) Gluteus Maximus (GMAX): extends hip

Equation of motion for the link model is shown in eq. (5).

$$\mathbf{M}(\Theta)\ddot{\Theta} + \mathbf{h}(\Theta,\dot{\Theta}) + \mathbf{g}(\Theta) + \mathbf{D}(\Theta,\dot{\Theta}) = \mathbf{T}_{\mathrm{JNT}} + \Phi(\Theta), \tag{5}$$

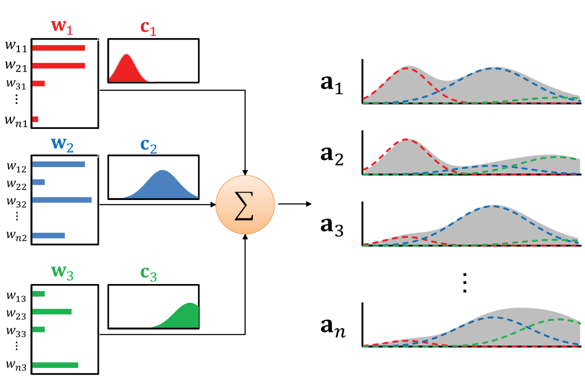
where $\mathbf{M}(\Theta) \in \mathbb{R}^{3\times3}$, $\mathbf{h}(\Theta,\dot{\Theta}) \in \mathbb{R}^{3\times1}$, and $\mathbf{g}(\Theta) \in \mathbb{R}^{3\times1}$ are obtained from Lagrange equation to indicate inertia term, non-linear term, and gravitational term. Vertical and horizontal reaction force ($\Phi(\Theta) \in \mathbb{R}^{3\times1}$) was generated at the hip from two elastic elements as in Fig. 2 (a). $\mathbf{D}(\Theta,\dot{\Theta}) \in \mathbb{R}^{3\times1}$ is the damping term of each joint, and its component $D_k(\Theta,\dot{\Theta})$ is the damping force generated in the joint k $(k = 1,2,3)$. It consists of the passive force (D_k^{RANGE}) and damping force (D_k^{DAMP}) (eq. 6). Passive force is generated at the limits of joint movement by passive joint structure and damping force is generated in proportion to joint angular velocity D_k^{RANGE} and D_k^{DAMP} are calculated from eqs. (7)–(8).

$$D_k(\theta_k,\dot{\theta}_k) = D_k^{\mathrm{RANGE}} + D_k^{\mathrm{DAMP}}, \tag{6}$$

$$D_k^{\mathrm{RANGE}} = x_{1k}\exp(-x_{2k}(\theta_k - x_{3k})) - x_{4k}\exp(-x_{5k}(x_{6k} - \theta_k)), \tag{7}$$

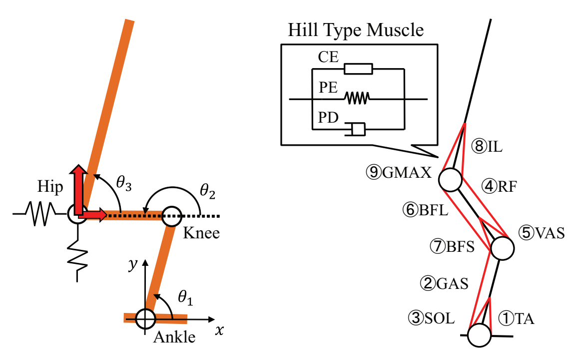
$$D_k^{\mathrm{DAMP}} = d_k\dot{\theta}_k. \tag{8}$$

$\mathbf{T}_{\mathrm{JNT}} \in \mathbb{R}^{3\times1}$ consists of joint torques which are generated from muscle torque ($\mathbf{RF}(L_i,\dot{L}_i,a)$) and posture stabilization



(a) Muscle Synergy (b) Muscle Activation

Fig. 1. Muscle Synergy Model



(a) Skeletal Model (b) Muscle Molde

Fig. 2. Musculoskeletal Model

torques ($\mathbf{T}_{\mathrm{FB}}$) (eq. (9)). In the equation, $\mathbf{R}$ indicates matrix of muscles moment arm expressed as in eq. (10). Components of the matrix $\mathbf{R}$ indicate the moment arm r_{ki} of muscle i ($i = \mathrm{TA, GAS, SOL, RF, VAS, BFL, BFS, IL, GMAX}$) to the joint k. Moment arm r_{ki} is zero if the muscle i is not attached to the joint k, otherwise moment arm r_{ki} takes either positive or negative values depending on how muscles contribute to each joint (extension or flexion). Moment arm length is considered to be constant regardless of the body posture. $\mathbf{F}$ consists of muscular tensions generated from each muscle ($[F_1, F_2, \cdots, F_n]^{\mathrm{T}}$).

$$\mathbf{T}_{\mathrm{JNT}} = \mathbf{RF}(L_i, \dot{L}_i, a_i) + \mathbf{T}_{\mathrm{FB}}, \tag{9}$$

$$\mathbf{R} = \begin{pmatrix} r_{11} & \cdots & r_{1n} \\ \vdots & \ddots & \vdots \\ r_{31} & \cdots & r_{3n} \end{pmatrix}, \tag{10}$$

$$r_{ki} = \begin{cases} 0 & \text{(no attachment)} \\ r_{ki} & \text{(extension)} \\ -r_{ki} & \text{(flexion)} \end{cases}. \tag{11}$$

To calculate muscle tension, Hill type muscle model is used [11]. It consists of three elements: contractile element (CE) and parallel elements (PE and PD). The muscular tension generated in CE, PE, and PD of muscle i (f_i^{CE}, f_i^{PD}, and f_i^{PE}) is calculated from eqs. (13–15).

$$F_i(L_i, \dot{L}_i, a_i) = f_i^{\mathrm{CE}} + f_i^{\mathrm{PD}} + f_i^{\mathrm{PE}}, \tag{12}$$

$$f_i^{\mathrm{CE}} = F_i^{CE} h(L_i) k(\dot{L}_i) a_i, \tag{13}$$

$$f_i^{\mathrm{PD}} = c_i^{\mathrm{PD}} \dot{L}_i, \tag{14}$$

$$f_i^{\mathrm{PE}} = k_i^{\mathrm{PE}} (\exp[15(L_i - \overline{L}_i)] - 1), \tag{15}$$

$$dL_i = r_{ki} d\theta_k. \tag{16}$$

In the Hill type muscle model, CE generates tension based on muscle excitation level (a_i), muscle length (L_i) and muscular velocity ($\dot{L}_i$). In the eq. (13), F_i^{CE} is the maximum isometric force, $\overline{L}_i$ is the rest muscle length, $h(L_i)$ indicates length-tension relationship and $k(\dot{L}_i)$ is velocity-tension relationship. However, muscle length of CE is fixed to the rest length in this paper.

On the other hand, muscle length of PE and PD varies according to the joint angles. PE is the elastic component which generates contraction tension when muscles are extended. PD is the damping component which generates tension in proportion to muscular velocity. Changes in muscular length (dL_i) are calculated from moment arm r_{ki} and change of joint angle θ_k from its neutral position as in eq. (16) [12].

The additional torque ($\mathbf{T}_{\mathrm{FB}}$) is generated on the joints in order for the model to be stabilized and follow human standing-up motion. $\mathbf{T}_{\mathrm{FB}}$ is determined by PD control.

Joint damping coefficients ($x_{1k \cdots 6k}$ and d_k), moment arm of the muscles (r_{ki}), kinetic and damping parameters of muscles (k_i^{PE} and c_i^{PD}), maximum isometric force (F_i^{CE}) and rest muscle length ($\overline{L}_i$) are decided from the previous study [13].

C. Generation of Motion

In the study, firstly joint torques are computed with the same link model explained in the previous section by inverse dynamics using measured body trajectories and reaction force. These joint torques are used to determine the amount of muscular tension which is necessary to achieve the motion. Joint torques are decomposed to each muscle activation, but muscle activation cannot be determined exclusively since the musculoskeletal system includes bi-articular muscles (GAS, RF, and BFL) and one of the muscles (IL) cannot be measured due to the inner muscle. Therefore, muscle activation ($\mathbf{a}'_i$) is calculated by optimization to minimize the following squared error (Z) from the measured muscle activation ($\mathbf{a}_i$).

$$Z = \frac{1}{2} ||\mathbf{a}'_i - \mathbf{a}_i||^2. \tag{17}$$

We conduct forward dynamic simulations to calculate body kinematics of human standing-up motion. When initial body posture is given, body kinematics is repeatedly calculated from the current body posture and muscular torques generated on the joint. As described in the previous section, muscular torque is mainly calculated from muscle activation which is generated from muscle synergies $\mathbf{w}_j$ and their time-varying weighting coefficient $\mathbf{c}_j$. In this paper, muscle synergies $\mathbf{w}_j$ are fixed and only time-varying weighting coefficients $\mathbf{c}_j$ are input to the musculoskeletal model to obtain body kinematics. For numerical simulation, fourth ordered Runge-Kutta method is used when dt is set to 0.001s.

III. Empirical Experiments with Humans

Optimal motion capture system (MAC3D) with eight cameras (HMK-80; Motion Analysis Corp.) was used to measure body trajectories in 200Hz. Measured body parts were decided based on Helen Hayse marker set, and joint angles were calculated from the software (SIMM; MuscuGraphics Corp.) The forceplate (Nitta Corp.) was used to measure reaction force from the hip in 64Hz. Using DL-3100 (S&ME Corp.), muscle activation was measured in 1,000Hz from eight muscles of the right leg (TA, SOL, GAS, RF, VAS, BFL, BFS, GMAX) since sagittal movement is focused.

Data 1.0s before and 2.25s after the time when the subject rises hip is used. All data is filtered with second order butter worth low-pass filter in 10, 25, and 25Hz respectively for body kinematics, reaction force, and muscle activation. In addition, muscle activation is centred, rectified, and normalized with minimum and maximum values during the experiment.

One participant (twenty-seven years old male, height: 1.77m, weight: 80kg) has participated in our experiment, and seventeen trials of standing-up motion were recorded. Chair height was set to the knee height and he was asked to have his arms crossed in front of his chest. Speed of the motion is not controlled clearly, but he was asked to perform the motion in comfortable speed. Before starting experiment, we have explained detail of the experiment, and consent was obtained. This study was conducted with approval by the Institute Review Board (IRB) of the University of Tokyo.

IV. Results

A. Extracted Muscle Synergies

In this study, $\mathbf{a}'_i$ is calculated from inverse dynamics and optimization from measured body trajectories, reaction force, and muscle activation of seventeen trials. Muscle synergies are extracted individually from seventeen trials of muscle activation $\mathbf{a}'_i$. Figure 3 shows mean squared error between observed muscle activation patterns and reconstructed activations from muscle synergy model. Error bars indicate standard deviation of squared error. As a results of ANOVA, there was a significant difference in squared error according to the number of synergies. Then, post-hoc analysis was applied to each neighboring number of synergies. It was obtained that there was a significant difference between one and two, and two and three. Therefore it is suggested that adding more muscle synergies would not improve the performance to represent muscle activation. In this study, the number of synergies was set to three.

Figure 4 indicates three extracted synergies from seventeen trials of the standing-up motion. Figures 4 (a, c, e) illustrate muscle activations included in each muscle synergy; separate bars show different muscle synergies of individual trials. Black solid bars show mean activation of each muscles.

On the other hand, time-varying weighting coefficients for each synergies are shown in Figs. 4 (b, d, f); thick sold lines show mean of time-varying weighting coefficients, and dashed lines indicate single trial of them. The vertical black lines indicates characteristic kinematic events during human standing-up motion reported in the previous study [5]: forward movement of shoulder (I: flexion momentum phase), rising hip (II: momentum transfer phase), minimum ankle dorsiflexion (III: extension), and maximum shoulder height (IV: posture stabilization phase).

B. Forward Dynamic Simulation

Link parameters for forward dynamic simulation are described in Table I: link length, mass, and position of center of mass, inertial moment. Link length is determined based on the measurement of a subject, and other parameters are decided based on the standard human body data [14]. Position of center mass indicates ratio of length from proximal end of the body. Proportional gain and differential gain for PD control to calculate $\mathbf{T}_{\mathrm{FB}}$ were 200 and 50 in this study. Target trajectories of PD control is set to the average measured body kinematics. Coefficients of two elastic elements in the floor model were set to 6000 and 7000N/m for horizontal and vertical directions in this study. Initial body posture for forward dynamic simulation is decided from beginning of average measured body kinematics.

During forward dynamic simulation, muscle synergy is fixed to the mean of synergies extracted from seventeen trials which is depicted as black bars in Fig. 4 (a, c, e).

Figures 5 (a)–(c) show results of generated three joint angles. The black solid lines show the joint angles generated from forward dynamic simulation whereas the red dashed lines show the mean of measured joint angles.

Figures 5 (d)–(f) show results of joint torques generated to each joint; the black solid lines indicate torques generated by muscles (**RF**) whereas the red dashed lines indicate torques to stabilize posture ($\mathbf{T}_{\mathrm{FB}}$). The ratio of joint torques to stabilize posture ($\mathbf{T}_{\mathrm{FB}}$) to torques actuated in joint angles ($\mathbf{T}_{\mathrm{JNT}}$) is 0.92, 0.49, and 7.7% for foot, knee and hip joints.

Figure 5 (g) shows comparison between simulated reaction force in the forward dynamic simulation and measured reaction forces in the horizontal and vertical directions. The red solid and dashed lines show forces in the horizontal direction. The black solid and dashed lines with circle markers illustrate simulated and measured reaction forces in the vertical direction.

Table II shows the coefficient of determination between simulated and measured joint angles and reaction forces. It is implied that the standing-up motion is successfully realized from forward dynamic simulation in terms of body kinematics and reaction force from the chair.

Figure 5 (h) shows stick picture of generated motion. Each stick picture shows body posture of every 0.25s.

V. Discussion

As a result, three muscle synergies are identified from human standing-up motion and correspond time-varying weighting coefficients are determined. Through the generation of the standing-up motion using forward dynamic simulation, it is confirmed that three muscle synergies are able to realize the standing-up motion rather than controlling individual muscles. Since posture stabilization torque ($\mathbf{T}_{\mathrm{FB}}$) is relatively small compared to the total joint torque (7.7% at most), the standing-up motion is mainly generated from coordination of each muscles.

Figures 4 (b, d, f) show that muscle synergies are activated in chronological order, and their time-varying weighting co-

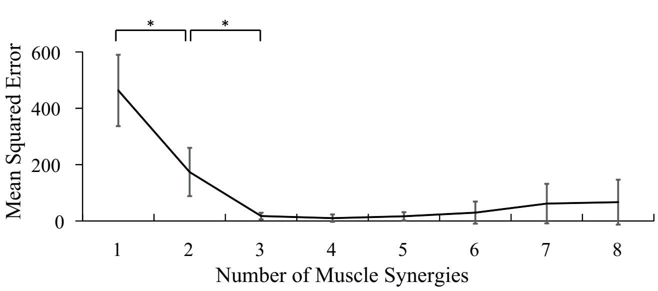


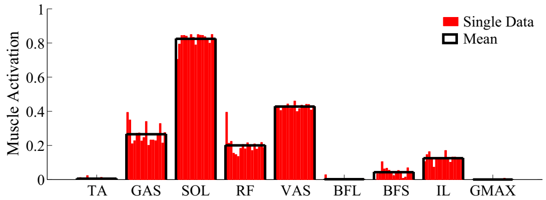
Fig. 3. Squares Error between Observed and Simulated Muscle Activation

TABLE I
LINK PARAMETERS

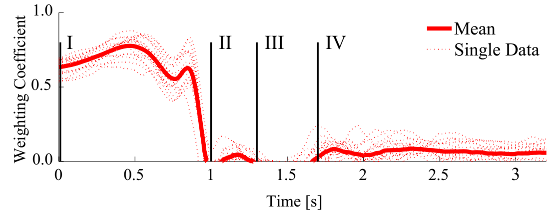
	Shank	Thigh	HAT
Length [m]	0.50	0.40	0.80
Mass [kg]	8.50	18.7	56.1
Center of Mass Position	0.41	0.42	0.20
Intertial Moment [kgm^2]	0.48	0.13	3.36

TABLE II
COEFFICIENT OF DETERMINATION

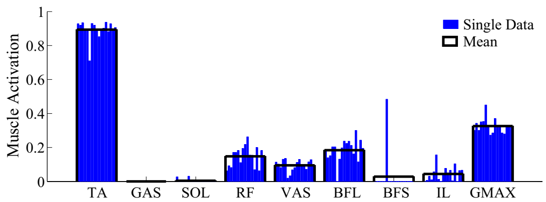
Vertical Reaction Force	Horizontal Reaction Force	Ankle Angle	Knee Angle	Hip Angle
0.997	0.994	0.999	0.999	0.972



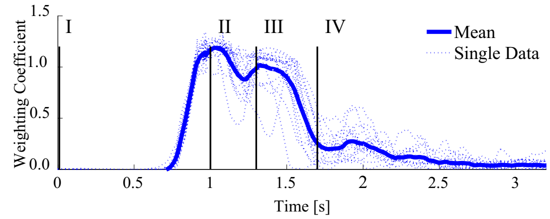
(a) Muscle Synergy 1



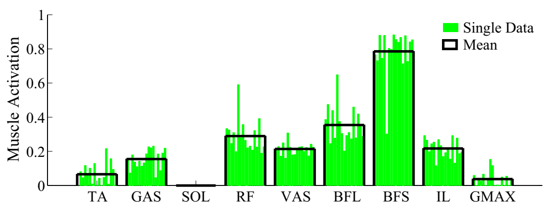
(b) Time-varying Weighting Coefficients of Synergy1



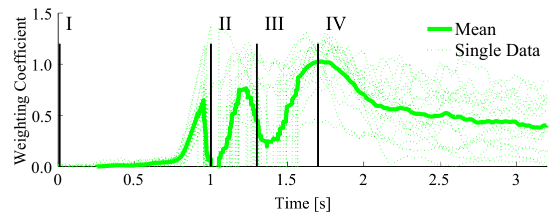
(c) Muscle Synergy 2



(d) Time-varying Weighting Coefficients of Synergy2

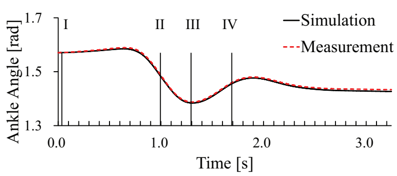


(e) Muscle Synergy 3

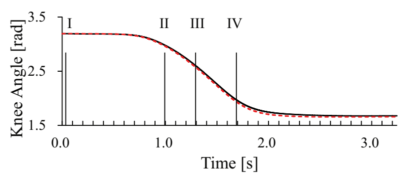


(f) Time-varying Weighting Coefficients of Synergy3

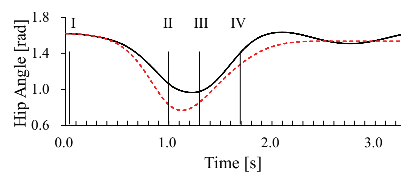
Fig. 4. Results of Extracted Synergies



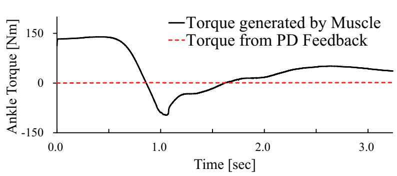
(a) Ankle Joint Angle



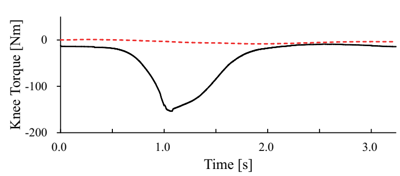
(b) Knee Joint Angle



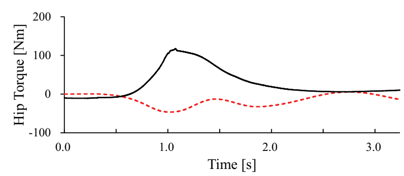
(c) Hip Joint Angle



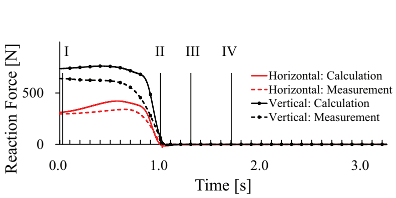
(d) Ankle Joint Torque



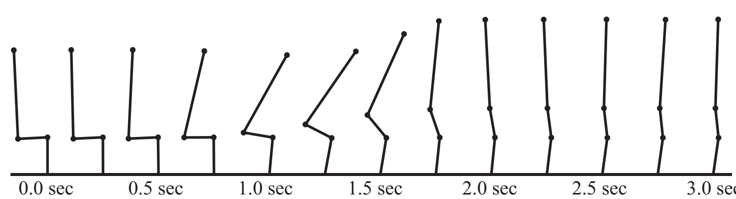
(e) Knee Joint Torque



(f) Hip Joint Torque



(g) Reaction Force



(h) Simulated Standing-up Motion

Fig. 5. Results of Simulated Standing-up Motion; I indicates the time when the subject starts moving his shoulder foward, II indicates rising hip, II indicates minimum ankle dorsiflexion, and III indicates the maximum shoulder height is achieved.

efficients clearly corresponded to the characteristic kinematic events defined in the previous study [5]. Figure 4 (b) shows the duration of muscle synergy 1 is between Phases I and II. Therefore it is implied that the first synergy works to flex their upper body to generate momentum. On the other hand, the maximum activation of the second synergy (Fig. 4 (d)) is found in the time when humans rise their hip (Phase II), and it continues during Phase II and III. It implies that the second synergy works as lifting up their hip and move trunk upward. Compared to other synergies, the third synergy (Fig. 4 (f)) has the maximum activation at the beginning of the Phase IV. It suggests that the last synergy has contribution toward posture stabilization. These results suggest that muscle synergies are related to the kinematic event during the standing-up motion, and synergies have own contribution toward body movement.

Currently, the function of posture stabilization is included in terms of joint torques $\mathbf{T}_{FB}$. However, humans usually balance their posture according to the sensory feedback from sensation of environmental changes. It will be necessary to implement the sensory feedback function to stabilize posture instead of joint torques $\mathbf{T}_{FB}$.

One of the reason for the hip joint to have the larger error than the other joints is because of illiopsoas (IL) which is flexor of the hip joint. Since IL is an inner muscle and cannot be measured in the experiment, muscle activation of IL is only estimated. Measuring additional muscles which contribute to hip flexor would improve the accuracy of our musculoskeletal model.

VI. Conclusion

In this study, a musculoskeletal model which considers human body dynamics and anatomical characteristics of muscles has been developed. Muscle activations of nine muscles in lower limb are decomposed into three muscle synergies. It has been shown that the standing-up motion could be generated by controlling time-varying weighting coefficient corresponded to extracted muscle synergies. Additionally it is implied that time-varying weighting coefficients are corresponded to characteristic kinematic events. The first synergy is activated when humans move shoulder forward (flexion momentum). The second synergy is strongly activated when people rises hip (momentum transfer). The third synergy has characteristic peak at the time when the highest shoulder position is achieved (posture stabilization). This result implies that human standing-up motion is composed of three muscle synergies.

Current study has employed the normalized situation to measure the standing-up motion, and the effect of link parameters is limited. Therefore structure of muscle synergies are thought to be similar among individuals, but different environment would affect the extracted synergies. For example, it is known that the standing-up motion is affected by the change of chair height or feet position. It will be interesting direction to investigate if muscle synergies are consistent in the environmental changes or humans need additional synergies to perform an adaptive motion.

Another important future direction is implementation of our findings into new training methods. If humans lose or forget the certain synergy due to injury or disease, they need to regain or relearn it. The previous study implies that if humans want to learn a new synergy, it would be the best to firstly try to use another existing synergy [15]. If there are similar synergies found in other movements, humans would be able to utilize these synergies by repeating the same movement. Otherwise, humans need to train and reconstruct it again. In terms of standing-up motion, it is implied that humans need to learn both specific muscle activation included in muscle synergies and how to switch muscle synergies based on kinematic phases. Therefore the new assistive method should focus on teaching how to manage synergies as well as adding deficient torques.

Acknowledgement

This work was in part supported by the MEXT KAKENHI, Gtant-in-Aid for Scientific Research (B) 24300198, JST RISTEX Service Science, Solutions and Foundation Integrated Research Program, and Grant-in-Aid for JSPS Fellows 24·8702.

References

[1] Alexander NB, Schultz AB, and Warwick DN, "Rising from a Chair: Effects of Age and Functional Ability on Performance Biomechanics", J. Gerontol., vol. 46, pp. 91-98, 1991.

[2] Chugo D, Kawabata K, Okamoto H, Kaetsu H, Asama H, Miyake N, and Kosuge K, "Force Assistance System for Standing-up Motion", Ind. Robot, vol. 34, pp. 128-134, 2007.

[3] Sale DG and MacDougall JD, "Specificity in Strength Training; a Review for the Coach and Athlete", Can. J. Appl. Sports Sci., vol. 6, pp. 87-92, 1981.

[4] Rutherford OM, "Muscular Coordination and Strength Training Implications for Injury Rehabilitation", Sports Medicine, vol. 4, pp. 196-202, 1988.

[5] Schenkman M, Berger RA, Patrick OR, Mann RW, and Hodge WA, "Whole-body Movements during Rising to Standing from Sitting", Phys. Ther., vol. 70, pp. 638-651, 1990.

[6] Bernstein N, "The Co-ordination and Regulation of Movement", Pergamon, Oxford, 1967.

[7] d'Avella A and Bizzi E, "Shared and Specific Muscle Synergies in Natural Motor Behaviors", PNAS, vol. 102, pp. 3076-3081, 2005.

[8] Ivanenko YP, Poppele RE, and Lacquaniti F, "Five Basic Muscle Activation Patterns Account for Muscle Activity during Human Locomotion", J. Physiol., vol. 556, pp. 267-282, 2004.

[9] An Q, Ikemoto Y, and Asama H, "Synergy Analysis of Sit-to-Stand in Young and Elderly People", J. of Robot. and Mech., vol. 25, pp. 1038-1049, 2013.

[10] Lee DD and Seun HS, "Learning the Parts of Objects by Non-Negative Matrix Factorization", Nature, vol. 401, pp. 788-791, 1999.

[11] Zajac FE, ' 'Muscle and Tendon: Properties, Models, Scaling, and Application to Biomechanics and Motor Control", Critical Reviews in Biomedical Engineering, vol. 17, pp. 359-411, 1989.

[12] Melissa GH, Felix EZ and Michael EG, "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., vol. 23, pp. 157-169, 1990.

[13] Davy DT and Audu, ML, "A Dynamic Optimization Technique for Predicting Muscle Forces in the Swing Phase of Gait", J. Biomech., vol. 20, pp. 187-201, 1987.

[14] Clauser CE, McConville JT, and Young JW, "Weight, Volume, and Center of Mass of Segments of Human Body", Wright-Patterson Air Force Basem Ohion, AMRL Technical Report, pp. 69-70, 1969.

[15] Richard GC, "Changes in Muscle Coordination with Training", J. of Appl. Phys., vol. 101, pp. 1506-1513, 2006.