



Original Article

## New estimation model of the initial lower limb angle to improve angle estimation during the extension phase of standing-up movement

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**Abstract.** [Purpose] An estimation model of the knee and ankle joint angles during the extension phase was proposed in the previous study. However, it had limited use because of the fixed initial lower limb angle before standing up. This study aimed to propose a new estimation model of the initial lower limb angle to improve the angle estimation during extension phase. [Subjects and Methods] Seven healthy male volunteers were enrolled. The new estimation model approximated the initial lower limb angle using a force sensor plate that measured the plantar pressure of the subjects. The estimated angle and force were compared to those obtained by a motion capture system and force plate. [Results] The new estimation model of initial lower limb angle showed no significant difference compared with the true values obtained by motion capture, except for the subject who had a greater foot-pressure measurement error compared with the force plate measurement, with maximum errors of 5.98° and 6.31°, respectively. [Conclusion] The proposed model in this study can estimate the initial lower limb angle before standing and can be applied to the angle estimation model during the extension phase of the standing-up movement.

**Key words:** Standing-up motion, Initial lower limb angle, Plantar pressure

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### INTRODUCTION

Standing-up from a chair is directly related to walking and is frequently performed every day. This is difficult for some elderly individuals because of weakened muscles or motor abilities. Training elderly individuals to attain a standing position and assisting them with the standing-up motion involved in rising from a chair are important to their Quality of Life (QOL), which has become an important issue in modern-day Japanese society<sup>1)</sup>.

Analysis of the posture parameters during the standing-up motion is useful for the physical therapists and caregivers for rehabilitation training or assisting in movement. Posture parameters could give feedback of movement during rehabilitation training of standing-up motion, and make body fit better with assistance devices to avoid pain or discomfort feel during standing-up motion assistance.

Motion capture system can precisely measure changes in body posture in any direction<sup>2)</sup>. However, its daily use is difficult because of high cost and specific space requirements. It also causes discomfort to the users because of the many reflective markers attached on whole parts of body. Incidentally, we think body posture would be expressed by combining joint angles (hip, knee ankle joint angles). There were some studies which focused on the phase division of the standing-up motion according to changes in three joint angles<sup>3, 4)</sup>. Moreover, joint angle can be used as a visualized parameter which is easier to analyze during rehabilitation train of standing-up motion, and also easier to coordinate with support device during assistance.

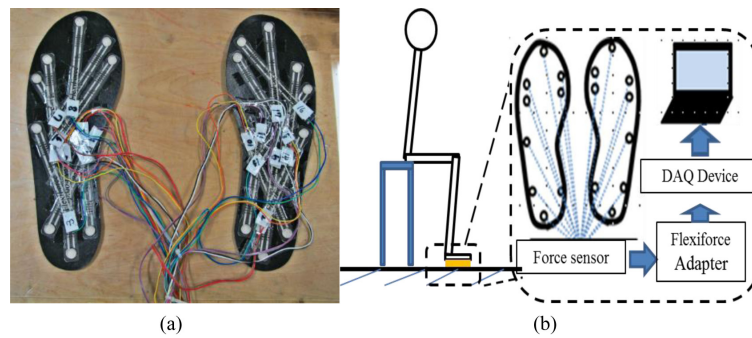
Therefore, development of an estimation system with lower cost and easier to use so that provide kinematical analysis of

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**Fig. 1.** (a) Force sensor plate and arrangement of sensors (b) Schematic diagram of plantar force measurement system.

Arrangement of sensors: the first toe, the third toe, the fifth toe, the first metatarsal bone, the fifth metatarsal bone, cubic bone, calcaneus bone, and the calcaneus were supposed to support three points.

standing-up motion focusing on angle estimation is necessary in daily life.

As we know, standing-up movement can be divided into flexion phase and extension phase, and it is difficult for elderly and disabled people to combine two phases, so each phase is performed separately<sup>5</sup>. Estimation of extension phase, which starts from hip and knee extension to the upright position, is important for analysis of standing-up movement. A three-link model and related equations for daily use were designed for angle estimation during the extension phase of standing-up movement in a previous study<sup>6</sup> to provide posture information on the extension phase, instead of using a motion capture system. In the estimate model of a previous research, however, the initial knee and ankle joint angles before standing up were fixed at  $80^\circ$  and  $70^\circ$ , respectively. The previous estimate model could not be used when the initial knee and ankle joint angles were unknown. Therefore, the use of the previous estimate model was restricted because of different initial knee and ankle joint angles. Furthermore, this fixed angle led to estimation errors because the initial angle was erroneous. Thus, it is necessary to determine the initial knee and ankle joint angles in a stable state before commencement of the standing-up motion. Previous research showed that when rising from a chair, 18% of the body weight was concentrated on the foot before forward tilting of the trunk<sup>7</sup>, and foot placement manipulations were used to modify weight-bearing distribution<sup>8</sup>. From these results, the change in the ground reaction force applied to the feet is found to be related to knee and ankle joint angles in the stable state before commencement of the standing-up motion. A force plate can measure the ground reaction forces generated by a body standing on or moving across it, but it cannot be used daily because of its high cost and weight. It is possible to measure the vertical ground reaction force applied on the foot using a force sensor and eliminate the sense of restraint experienced by the user<sup>9</sup>. A new estimation model for initial angles using a force sensor plate that can measure foot pressure was proposed in this study.

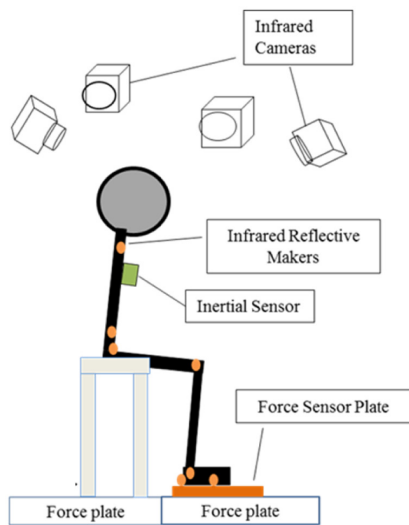
## SUBJECTS AND METHODS

Seven healthy male volunteers were enrolled in this study (mean age  $25 \pm 3$  years; mean height  $168.1 \pm 3.7$  cm; and mean weight  $64.7 \pm 9.4$  kg). This research was approved by the Ethics Committee for Human Research of Graduate School of Life Science and Systems Engineering, Kyushu Institute of Technology. All subjects received a description of this study, and signed a written, informed consent form before participating in this study.

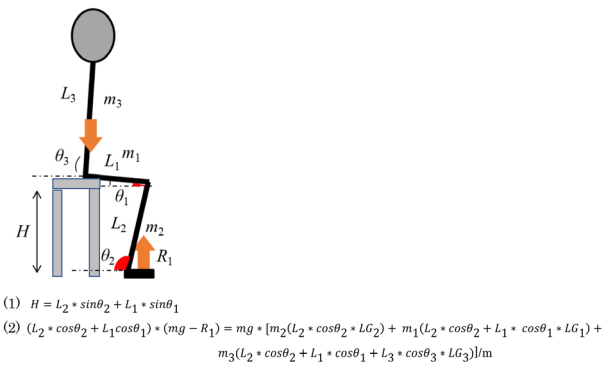
Experiment design for evaluation consisted of estimation of initial angle and estimation of body movement during extension phase based on the estimated initial angle results.

Figure 1 shows the force sensor plate designed for the measurement of the ground reaction force applied to the foot and a schematic diagram of the measurement device proposed by this study. The force sensor plate was designed as a removable plate with a pair of force sensor insoles. In the insole, nine FlexiForce button sensors (Tekscan Product Corporation) were arranged for one foot, with 18 button sensors for both feet. Arrangement of the sensors was based on the previous study<sup>10</sup>, as shown in Fig. 1. In the proposed measurement device (1) the pressure value was converted to resistance by a force sensor in the insole, located on the removable plate; (2) the Flexiforce Adapter (Phidgets Inc.) was used to load the interface of Flexiforce sensor onto the computer using the cable with an adapter to connect it to the force sensor; (3) to capture the signals from the FlexiForce button sensor and send the signal to the PC, NI cDAQ 9172 and 9205 (National Instrument Inc.), which are capable of simultaneous sampling from 18 channels, were used. The measurement of signal value from the pressure sensor can be recorded using C# program.

Figure 2 shows the experimental set-up. Subjects were asked to sit on a 40-cm high chair. The chair and feet of the subject were set on two pieces of the force plates (AMTI JAPAN). One force sensor plate, designed as described above, was set



**Fig. 2.** Experimental set-up.



**Fig. 3.** Initial lower limb angles estimate model and equations.  $L_3, L_2, L_1$ : length of trunk, length of lower leg, length of thigh.  $\theta_2, \theta_1$  were defined as initial lower limb angles in stationary state before standing-up motion.  $\theta_1$  is the angle between the horizontal line and the thigh,  $\theta_2$  is the angle between the horizontal line and the crus,  $\theta_3$  is the angle between horizontal line and body trunk.  $R_1$  is the vertical ground reaction force applied on the foot during stationary state before commencement of standing-up motion.  $H$  is the chair height.

under feet of the subject. The initial knee and ankle joint angles were set to around  $80^\circ$  and  $70^\circ$ , respectively. Each angle was decided by the subject. A 9-axis wireless motion sensor (Logical Product Corporation) was attached with an elastic strap to the chest (front of body of sternum, T5–T6) of each subject, same as that in the previous study<sup>6</sup>. The body movement was simultaneously measured using a motion capture system which provide true value in experiment. Fourteen infrared reflection markers were placed on the main joints of the body, such as the shoulders, iliac crests, greater trochanters, knees, and lateral malleolus. Six infrared cameras (OptiTrack, Natural Point Inc.) followed the markers.

The subjects sat on a chair and maintained a stationary state for three seconds before standing up. This study was designed for elderly individuals; therefore, the subjects were asked to rise from the chair at a slightly slower speed than they would do usually in their daily life. The trial started with the signal for measurement until the assumption of an upright posture. Each subject was asked to perform 10 trials. The signals from the pressure sensor and inertial sensor were recorded simultaneously using C# program. The sampling rate for all measuring devices was set at 100 Hz. The estimation model of the initial lower limb angle and related equations are shown in Fig. 3. The total force applied to the ground is balanced by the amount of force applied to the human body in a stationary state before standing up. The initial lower limb angles were calculated using the MATLAB software based on estimate equations.

Descriptive statistical analysis was used to obtain the means and standard deviations of the results from the proposed system and the motion capture system for analysis of the root-mean-square error (RMSE) of lower limb angles, as well as joint angles during extension phase. Paired t-tests on SPSS (IBM SPSS Statistics ver.21.0) showed a significant difference between the estimated result and true value obtained by the motion capture system in the test for initial lower limb angles. Wilcoxon's rank-sum test was used for the two subjects because of non-normal distribution

## RESULTS

The data obtained one second after the experiment started were used for estimation of the initial lower limb angles in this study. The average RMSE values of the initial estimated lower limb angles ( $\theta_1$  and  $\theta_2$ ) compared with true values obtained by the motion capture system are summarized in Table 1, as are the means and standard deviations (SDs). The maximum RMSE value was  $6.31^\circ$ , which performed by subject 6 for the estimation of  $\theta_2$ . The minimum RMSE value was  $2.78^\circ$ , which performed by subject 1 for the estimation of  $\theta_1$ . The estimated error ranges of  $\theta_1$  and  $\theta_2$  were  $2.78$ – $5.98^\circ$  and  $4.87$ – $6.31^\circ$ , respectively. The paired t-test showed that the estimated results for initial angle among the six subjects have no significant differences in the true value. The estimated result for  $\theta_2$  of subject 6 showed a significant difference compared with the value from the motion capture system ( $p < 0.05$ ).

Table 2 shows the average RMSE value of initial lower limb angle compared with true value before and after estimation. As seen in Table 2, the RMSE value of angle  $\theta_1$  was  $4.23$ – $7.54^\circ$  before the initial angle estimation, whereas for angle  $\theta_2$ , it was  $8.47$ – $10.96^\circ$ . After initial angle estimation, the RMSE value of angle  $\theta_1$  was  $2.78$ – $5.98^\circ$ , whereas angle  $\theta_2$  was  $4.92$ – $6.31^\circ$ . The RMSE values of the initial lower limb angle decreased after estimation.

**Table 1.** Average RMSE values of estimate angle ( $\theta_1$  and  $\theta_2$ ) compared with true value from motion capture system (seven subjects)

Subject	RMSE (degrees)	RMSE (degrees)
	$\theta_1$	$\theta_2$
1	2.78 ± 0.75	5.83 ± 1.47
2	3.05 ± 1.38	5.01 ± 1.71
3	4.01 ± 0.55	4.92 ± 1.15
4	4.52 ± 1.06	5.47 ± 0.70
5	3.16 ± 1.17	4.87 ± 1.31
6	5.98 ± 1.56	6.31 ± 1.08*
7	4.03 ± 1.12	5.63 ± 1.03

RMSE is root-mean-square error.  
 RMSE data are means ± SD.  
 Paired t-test, \*p<0.05.

**Table 2.** Average RMSE value of initial lower limb angles ( $\theta_1$  and  $\theta_2$ ) compared with true value before and after initial angle estimation (seven subjects)

Subject	RMSE (degrees)		RMSE (degrees)	
	$\theta_1$		$\theta_2$	
	Before estimation	After estimation	Before estimation	After estimation
1	6.32 ± 1.02	2.78 ± 0.75	9.57 ± 2.03	5.83 ± 1.47
2	7.54 ± 0.87	3.05 ± 1.38	9.96 ± 1.67	5.01 ± 1.71
3	4.23 ± 0.89	4.01 ± 0.55	10.36 ± 1.03	4.92 ± 1.15
4	7.19 ± 1.07	4.52 ± 1.06	8.47 ± 0.7	5.47 ± 0.70
5	6.96 ± 1.56	3.16 ± 1.17	10.96 ± 1.58	4.87 ± 1.31
6	7.26 ± 2.14	5.98 ± 1.56	8.97 ± 0.98	6.31 ± 1.08
7	7.25 ± 1.85	4.03 ± 1.12	9.36 ± 1.47	5.63 ± 1.03

RMSE is root-mean-square error.  
 RMSE data are means ± SD.

## DISCUSSION

The new estimation model for the initial lower limb angle showed the following results: The initial lower limb angle estimated using a foot sensor plate in the stationary state before standing up showed no significant difference compared with the true value among the six subjects; Among the six subjects, the maximum and minimum estimated errors were 5.83 degrees and 2.78 degrees, respectively. These angle estimated errors were decreased compared to estimated error by using predetermined lower limb angle in the previous research<sup>6)</sup>, and estimation error range of initial lower limb angles were decreased from 4.23–7.54 degrees to 2.78–4.52 degrees and, from 8.47–10.96 degrees to 4.92–5.83 degrees, respectively (as shown in Table 2). Based on estimated initial lower limb angle, the estimation of joint angles during extension phase was performed. Table 3 shows the average RMSE values of the knee and ankle joint angles during the extension phase. The maximum error and minimum error for the knee joint angle estimation were 4.49° and 2.61°, respectively, whereas the maximum error and minimum error for ankle joint estimation were 3.79° and 2.38°, respectively. In the previous study, the maximum error and minimum error for knee joint angle estimation were 4.58° and 3.68°, respectively, whereas the maximum error and minimum error for ankle joint estimation were 4.50° and 3.10°, respectively, with fixed initial knee and ankle joint angles<sup>6)</sup>. Therefore, angle estimation error during the extension phase was decreased with estimation of the initial lower limb angle. Kiyama et al. reported that the average error range of position sense between s between imitation angle and setting angle in the knee joint angle was -4.1 to 4.5°<sup>11)</sup>, and it is normal for the knee imitation angle to cause errors above 10 degrees.

Some errors in initial lower limb angle estimation still occurred and were thought to be related to the accuracy of the estimation model and force sensor. In the proposed three-link model shown in Fig. 3, body segment parameters based on previous research were set to obtain the mass, length, and center of gravity of each body part. To some extent, error caused in this research was related to individual differences of parameter. Furthermore, Segment parameters from previous research<sup>12)</sup> were used in this study and were obtained by conducting measurements on Americans, who have higher mean weight than Japanese.

**Table 3.** Average RMSE value of knee and ankle joint angles compared with true value during extension phase (six subjects)

Subject	RMSE (degrees)	
	Knee joint angle	Ankle joint angle
1	2.99 ± 0.56	3.03 ± 0.66
2	2.61 ± 0.53	3.79 ± 1.36
3	4.49 ± 0.64	2.38 ± 0.21
4	3.26 ± 0.32	2.86 ± 1.05
5	4.06 ± 0.32	3.65 ± 1.02
7	3.43 ± 0.47	3.02 ± 0.68

RMSE is root-mean-square error.  
RMSE data are means ± SD.

**Table 4.** Average errors between force sensor and force plate (seven subjects)

Subject	Average error (N)	Error percentage (%)
1	18.12 ± 3.40	18.82
2	24.71 ± 4.92	19.10
3	27.68 ± 4.32	20.67
4	27.09 ± 5.52	19.81
5	28.01 ± 4.93	24.03
6	37.98 ± 4.85	49.1
7	29.71 ± 2.33	23.00

Error percentage was obtained by comparing average error with the force applied on feet before commencement of standing-up motion.

It was observed that there was difficulty in obtaining an accurate measure of the force using the force sensor because of accuracy<sup>10</sup>. The force sensor measurement errors were related to factors such as the calibration method and state of the force sensor<sup>13</sup>. Previous research has reported that the first subject had a more accurate force sensor measurement than did the later subjects because of the trend of decreasing accuracy of force sensor measurement over time<sup>10</sup>. Table 4 shows the average errors between force sensor and force plate in this study. As shown in Table 4, the force sensor error of subject 6 was higher than for the other subjects, thus resulting to a larger error in the initial angle estimation compared to that for other subjects, with a significant difference compared with the true value. In subject 6, the force applied on the feet during the stationary state was very small because of lower weight. Moreover, the foot size of subject 6 was smaller than the insole in which the force sensors were placed. In this study, the insole with force sensors is 27.5 cm in length. Among the seven subjects, three subjects have a foot length of 27 cm; two, 26.5 cm; one, 26 cm; and subject 6, 25 cm. For subject 6, the arrangement of sensors did not correspond with points under feet, compared to that in other subjects, resulting in a larger error in pressure measurement.

In summary, based on the three points discussed above, the estimation model of the initial lower limb angle can be used for initial angle estimation under different foot placement situations as an improved model for angle estimation in the extension phase. There is little research focused on angle estimation for standing-up motion. Moreover, this study is advanced compared with the previous study and is novel because it estimated the initial lower limb angle using plantar pressure.

Future studies can focus on the following points: 1) improving the accuracy of estimation of the initial lower limb angle by decreasing the measurement error of the force sensor; 2) revising the estimated initial lower limb angle according to a statistic analysis of error caused by individual difference of parameters. Putting measured force plate data and the recorded segments data (trunk, thigh and lower leg) from motion capture into the model algorithm presented in Fig. 3, so that estimate the initial lower limb angle which eliminate the error caused by individual difference of parameters. 3) increasing the number of subjects, and the addition of some individuals with disabilities, such as Parkinson's disease.

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### Conflict of interest

None.

## REFERENCES

- 1) Eto H, Nagamoto H: Development of an elbow-supporting device to assist standing-up motion. *Robomech J*, 2015, 2: 10. [CrossRef]
- 2) Park SW, Park HS, Kim JH, et al.: 3D displacement measurement model for health monitoring of structures using a motion capture system. *Measurement*, 2015, 59: 352–362. [CrossRef]
- 3) Schenkman M, Berger RA, Riley PO, et al.: Whole-body movements during rising to standing from sitting. *Phys Ther*, 1990, 70: 638–648, discussion 648–651. [Medline] [CrossRef]
- 4) Kotake T, Dohi N, Kajiura T, et al.: An analysis of sit-to-stand movements. *Arch Phys Med Rehabil*, 1993, 74: 1095–1099. [Medline] [CrossRef]
- 5) Wada C, Tang YT, Arima T: Development of a standing-up motion guidance system using an inertial sensor. *Advanced Intelligent System.*, 2014: 179–187.
- 6) Jin F, Nagasaki T, Wada C: An estimation of knee and ankle joint angles during extension phase of standing up motion performed using an inertial sensor. *J Phys Ther Sci*, 2017, 29: 1171–1175. [Medline] [CrossRef]
- 7) Nozawa M, Mekuro, Yamamoto Y, et al.: A study on the relation between trunk forward and foot weight load in rising movement. *Jpn J Allied Health Rehabil*,

2002, 1: 19–22 (In Japanese).

- 8) Lecours J, Nadeau S, Gravel D, et al.: Interactions between foot placement, trunk frontal position, weight-bearing and knee moment asymmetry at seat-off during rising from a chair in healthy controls and persons with hemiparesis. *J Rehabil Med*, 2008, 40: 200–207. [[Medline](#)] [[CrossRef](#)]
- 9) Razak AH, Zayegh A, Begg RK, et al.: Foot plantar pressure measurement system: a review. *Sensors (Basel)*, 2012, 12: 9884–9912. [[Medline](#)] [[CrossRef](#)]
- 10) Wada C, Tokunaga M: Estimating standing-up motion by using foot pressure to assist standing up for the elderly. *SICE Annual Conference*, 2012: 1956–1958. 2012,08,23, Akita, Japan.
- 11) Kiyama T, Kawamura M, Ida K, et al.: The position sense in the normal knee of the young. *Jpn Phys Ther Assoc*, 1992, 19: 597–599 (In Japanese).
- 12) David AW: *Biomechanics and motor control of human movement* 4th edition. Translated by Akinori N and Shinsuke Y. 2009, pp 82–89, John Wiley & Sons, Inc. (In Japanese).
- 13) Fujii Y, Maru K: Self-correction method for dynamic measurement error of force sensors. *Exp Tech*, 2011, 35: 15–20. [[CrossRef](#)]