An estimation of knee and ankle joint angles during extension phase of standing up motion performed using an inertial sensor

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Abstract. [Purpose] Motion capture system is difficult to use in daily life. The aim of this study was to propose an estimation model for knee and ankle joint angle measurements and locate body center of gravity (COG) of the extension phase during standing-up motion. [Subjects and Methods] Seven healthy male volunteers were enrolled. An estimation model was proposed for the knee and ankle joint angle measurements by combining the angle and acceleration of the trunk, based on readings from the inertial sensor attachment on the subject’s chest, during the extension phase. Joint angles and COG position were compared to those obtained by a motion capture system. [Results] The joint angles and COG position demonstrated high correlation coefficients which represent strong correlation between the proposed model and the motion capture system. The proposed model could estimate the joint angle during extension phase, with a maximum error of 4.58 degrees, as well as COG position in the horizontal and vertical directions with maximum errors of 4.48 cm and 3.19 cm, respectively. [Conclusion] The proposed system could be used instead of motion capture system to estimate knee and ankle joint angles; however, the estimation of the COG position was insufficient because of lacked accuracy.

Key words: Standing up motion, Extension phase during standing up, Inertial sensor

INTRODUCTION

Standing up motion is a complex activity, the final goal of which is to move the body center of mass from a fully stable position, the sitting position (3-point base), to an upright position, the standing position (2-point base). This transition is different because of the different strategies of standing up motion employed by various individuals. In order to overcome the inability to perform the standing up motion, there has been much research focused on analysis and strategy modification of standing-up motion by evaluating the parameters of this motion. One study found that by changing the strategy of standing up to modify the trajectory of center of mass during standing up motion has biomechanical impacts on body. The kinematics of the standing-up motion have been investigated through the time histories of joint angles in previous research. Some other sit-to-stand studies have compiled kinematic data, such as COG of body, by using 3D motion capture systems and kinetic data from force plate in order to analyze and clarify the standing up motion.

With regard to the importance of standing up motion in daily-life activity, and due to the complexity of this movement, analysis of it is useful for therapists and doctors. In some hospitals, the medical staff tries to train the elderly to stand up from a chair without any support, according to the analytical results of the motion capture and force plate system. However this system is difficult for the elderly to incorporate and use in their daily lives.

Standing up motion can be divided into two phases: the phase in which COG is moved forward (flexion phase), and the phase
in which COG is move upward (extension phase). Healthy people can perform standing up motion successfully by combing the two phases, but some elderly people cannot stand up successfully because they cannot move their COG forward during the flexion phase. In our laboratory, we conducted some research of the flexion phase using a guidance system which mainly included an inertial sensor. It can help the elderly perform the flexion phase by informing them of the most suitable time to leave the chair, as well as provide an estimation and analysis of the flexion phase6).

The aim of this study was to estimate knee and ankle joint angles in order to provide parameters for extension phase analysis by using the inertial sensor which we used in the guidance system in previous research.

COG is a geometric property of any object. It is the average location of the weight of a body. We can completely describe the motion of a body through space in terms of the translation of the COG of the body from one place to another. In order to describe the motion of the extension phase, the position of the COG was calculated based on body angle data.

To evaluate the accuracy of this system, the angle data obtained using proposed system were compared with those measured by a motion capture system, as well as the transformation of the COG position.

SUBJECTS AND METHODS

Seven healthy male volunteers were enrolled (mean age 24 ± 0 years; mean height 171.6 ± 3.5 cm; and mean weight 64.0 ± 9.9 kg). This study was conducted in accordance with the ethical principles of the Declaration of Helsinki. All subjects received a description of this study and provided their written, informed consent before participating in this study.

Figure 1 shows the experimental set-up. A 9-axis wireless motion sensor (Logical Product Corporation) was attached to the chest (front of body of sternum, T5–T6) of each subject with an elastic strap and was measured with a sampling rate of 100 Hz. The inertial sensor has three-axis gyro sensor, three-axis acceleration sensor and three-axis geomagnetic sensor, memory size is 32MB, 1 to 1,000 Hz of sampling frequency, number of quantization is 12 bit, and it can connect the computer by USB for wireless. Subjects were asked to sit on a chair with a height of 40 cm, because the chair height of 40 cm is usually used in daily life, such as in family or company, and the chair and feet of subject were respectively set on two sheets of force plate (AMTI JAPAN). Body movement while standing up was simultaneously measured using a motion capture system. 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, and one marker was placed on the inertial sensor to measure the position transformation of sensor and the length between sensor and greater trochanters. Fifteen infrared cameras (OptiTrack, Natural Point Inc.) followed the markers. The force plate was measured at a sampling rate of 100 Hz. The time when the buttock left the chair was determined to be the time when pressure on the chair was zero. For synchronization, the two systems were cable-connected.

The three link rigid body model was used in this study and shown in Fig. 2. The body was simplified to three parts: trunk, thigh and lower leg. And weight of the upper limb which consists of upper arm, forearm and hand belong to trunk due to its high flexibility. In Fig. 2, the body markers refer to previous research7. The weight and length of each part of body could be obtained if we know the height and weight of the subject.

The trunk angle data could be measured by inertial sensor, and the data processing algorithm was created using Visual C#. The knee joint angle (θ′) and ankle joint angle (θ″) during extension phase were determined using a new three-link angle model and equations which are shown in Fig. 3. The new model can estimate lower limb angles by combining angle and acceleration of trunk, which came from the inertial sensor during the extension phase. The body COG position in extension phase could be calculated by the data of body angle, weight and length of each part.

In order to investigate whether this proposed system could estimate the body extension phase at different standing up motion speeds, subjects were asked to stand up following three specific patterns, and each pattern was performed ten times by each subject. Pattern 1: Standing up at faster speed than normal; Pattern 2: Standing up at normal speed; Pattern 3: Standing up at slower speed than normal speed.

A descriptive statistical analysis was carried out based on the means and standard deviations for the analysis of angle and COG position root-mean-square error (RMSE) between our system and the motion capture system. Pearson correlation coefficient was also designed for results of angle and COG position (IBM SPSS Statistics ver. 21.0).

RESULTS

The average value of root mean squared error (RMSE) and Pearson correlation coefficient of angles (knee and ankle joint angles) between our system and the motion capture system for all subjects are illustrated in Table 1. Mean values, standard
deviation (SD), and Pearson correlation coefficient (PCC) were indicated. The maximum RMSE value was knee joint angle estimation (4.58 degrees) at slow speed. The minimum RMSE value was ankle joint angle estimation (2.32 degrees) at normal speed. Estimation error range of knee and ankle joint angles in the slow speed pattern were 3.68–4.58 degrees and 3.10–4.52 degrees, while they were 2.40–3.89 degrees and 2.35–3.30 degrees in the fast speed pattern. All angle correlation coefficient values between our system and the motion capture system were above 0.795, which represents characteristic value for the strong correlation, and the correlation was significant at the 0.01 level.

Average RMSE results and Pearson correlation coefficient of COG position in horizontal and vertical directions between our system and the motion capture system for all subjects are shown in Table 2. COG positions, which were calculated from angle data, have some errors. Mean RMSE values of horizontal and vertical directions at fast speed were 4.02 cm and 2.86 cm, while they were 3.47 cm and 2.98 cm at normal speed, and 4.48 cm and 3.19 cm at slow speed. The estimation error of COG position in the horizontal direction was higher than in the vertical direction for all patterns.

**DISCUSSION**

The estimated angle result of this study demonstrated a high Pearson correlation coefficient, which represent strong correlation between our system and motion capture, although our system had some estimation errors, but the error was smaller than that of previous research, and the experiments of this study caused no fail trails. In previous research, position of COG and positioning discrepancy in the body joints were used to estimate body joint angles during the extension phase of standing up motion, but it was easy to cause fail trails and there were bigger estimation errors. Our system was able to estimate knee and ankle joint angles during the extension phase with maximum error of 4.58 degrees and minimum error of 2.32 degrees. Double integration was used to calculate the joint angles, so we determined that part of the error was caused by integration of acceleration data. For body joint, there is a range of normal values for joint position sense. Previous research proposed by Kiyama et al. focused on the position sense of the normal knee of the young person, and reported that the average error range of the knee joint angle was –4.1 to 4.5 degrees between imitation angle and setting angle (it is normal for the knee imitation angle to cause errors above 10 degrees). So, we hold that our system is accurate enough for knee and ankle angle estimation during the extension phase of standing up motion.

The estimation error range of knee and ankle joint angles at slow speed was greater than normal and fast speed. We posit that the reason why slow speed demonstrated a larger estimation error is related to the integral acceleration data at slow
3.5 ± 0.9 cm in the horizontal direction, which was bigger than some others with an error of 3.2 ± 0.8 cm in the vertical direction and research. Previous research has estimated center of mass COG position with a maximum error of 4.48 cm in the COP in medio-lateral is 2.01 cm, and the maximum amplitude in the antero-posterior directions is 4.32 cm.11) Experiment, the average displacement vector in the vertical placement vector of COG during standing up motion. In our system and motion capture, our system could horizontal and vertical directions showed strong correlation between our system and motion capture.

Body sway at quiet standing posture by using a force plate, and the results showed that the maximum amplitude described by the COP in medio-lateral is 2.01 cm, and the maximum amplitude in the antero-posterior directions is 4.32 cm.11) Fang Wang et al. estimated body sway in terms of centroid trajectory by using an inexpensive webcam system. The results demonstrated that the average sway amplitude for anterior-posterior sway and lateral sway, respectively, were 14.04 cm and 13.8 cm (peak-peak)13). In this study, body sway exists in order to maintain a balanced position during standing up motion, and it can be captured by a motion capture system accurately, though it was difficult to measure using our system. Therefore, we hold that a part of the difference in COG position between our system and the motion capture system may be caused by body sway.

According to this study, we can provide the parameters of joint angles for extension phase analysis by using the proposed speed, which produces a larger integral error.

The Pearson correlation coefficient of COG position in horizontal and vertical directions showed strong correlation between our system and motion capture. Our system could estimate COG position with a maximum error of 4.48 cm in the horizontal direction, which was bigger than some other research. Previous research has estimated center of mass with an error of 3.2 ± 0.8 cm in the vertical direction and 3.5 ± 0.9 cm in the horizontal direction10), so accuracy improvement of our system for COG estimation in the vertical direction is needed.

COG estimation error in the horizontal direction was greater than in the vertical direction. It was related to the displacement vector of COG during standing up motion. In our experiment, the average displacement vector in the vertical direction was about 25 cm, which was greater than in horizontal direction vector of about 11 cm.

In three link model which used in this study, the trunk was defined as the thoracic spine, the lumbar spine, and the pelvis as one rigid body. Body trunk bending angle is treated as the hip flexion angle, because alignment change of the trunk (changes in bending of the thoracolumbar region) has not been taken into consideration. So it would cause error for COG position estimation.

As we know, body sway exhibited in daily life activities is defined as the slight postural movement made by an individual in order to maintain a balanced position. Typically, the term “body sway” is used to describe the extent of the center point of pressure (COP) or the COG (same as the center of mass COM) excursions11). A previous study recorded the measurement of body sway at quiet standing posture by using a force plate, and the results showed that the maximum amplitude described by the COP in medio-lateral is 2.01 cm, and the maximum amplitude in the antero-posterior directions is 4.32 cm.10) Fang Wang et al. estimated body sway in terms of centroid trajectory by using an inexpensive webcam system. The results demonstrated that the average sway amplitude for anterior-posterior sway and lateral sway, respectively, were 14.04 cm and 13.8 cm (peak-peak)13). In this study, body sway exists in order to maintain a balanced position during standing up motion, and it can be captured by a motion capture system accurately, though it was difficult to measure using our system. Therefore, we hold that a part of the difference in COG position between our system and the motion capture system may be caused by body sway.

According to this study, we can provide the parameters of joint angles for extension phase analysis by using the proposed

Table 1. Average RMSE values and Pearson correlation coefficient of knee and ankle joint angle of our system and motion capture system during extension phase

| Speed  | Subject | RMSE (degrees) | PCC  | RMSE (degrees) | PCC  |
|-------|--------|----------------|------|----------------|------|
|       |        | Knee joint angle |      | Ankle joint angle |      |
| Fast  | A      | 2.65 ± 0.69     | 0.989** | 2.50 ± 0.29     | 0.930** |
|       | B      | 2.68 ± 0.61     | 0.990** | 3.10 ± 0.54     | 0.924** |
|       | C      | 2.99 ± 0.37     | 0.988** | 2.35 ± 0.41     | 0.938** |
|       | D      | 2.57 ± 0.59     | 0.998** | 2.95 ± 0.39     | 0.896** |
|       | E      | 3.33 ± 0.68     | 0.965** | 2.45 ± 0.43     | 0.971** |
|       | F      | 3.54 ± 0.78     | 0.988** | 3.30 ± 0.89     | 0.819** |
|       | G      | 3.67 ± 0.38     | 0.901** | 2.97 ± 0.24     | 0.891** |
| Normal| A      | 3.53 ± 0.50     | 0.832** | 2.38 ± 0.19     | 0.815** |
|       | B      | 2.45 ± 0.70     | 0.819** | 2.38 ± 0.64     | 0.798** |
|       | C      | 2.71 ± 0.13     | 0.968** | 2.32 ± 0.40     | 0.826** |
|       | D      | 2.40 ± 0.61     | 0.946** | 3.56 ± 0.72     | 0.903** |
|       | E      | 3.25 ± 0.78     | 0.864** | 2.98 ± 0.21     | 0.895** |
|       | F      | 3.89 ± 0.89     | 0.841** | 3.03 ± 0.85     | 0.911** |
|       | G      | 3.56 ± 0.57     | 0.902** | 2.75 ± 0.84     | 0.847** |
| Slowly| A      | 4.58 ± 1.31     | 0.901** | 3.10 ± 0.66     | 0.801** |
|       | B      | 3.79 ± 0.96     | 0.856** | 3.33 ± 0.86     | 0.797** |
|       | C      | 3.81 ± 0.30     | 0.900** | 3.81 ± 0.32     | 0.796** |
|       | D      | 4.51 ± 1.72     | 0.860** | 3.55 ± 0.93     | 0.812** |
|       | E      | 4.58 ± 0.78     | 0.885** | 3.71 ± 0.45     | 0.832** |
|       | F      | 3.89 ± 0.78     | 0.954** | 4.52 ± 1.82     | 0.821** |
|       | G      | 3.68 ± 1.89     | 0.924** | 4.50 ± 1.33     | 0.901** |

RMSE: root-mean-square error; PCC: Pearson Correlation Coefficient
RMSE data are means ± SD.
**Correlation is significant at the 0.01 level

Table 2. Average RMSE values and Pearson correlation coefficient of COG position in horizontal and vertical directions of our system and motion capture system during extension phase

| Speed  | Direction | RMSE (cm) | PCC  |
|-------|-----------|-----------|------|
| Fast  | Horizontal | 4.02 ± 0.65 | 0.852** |
|       | Vertical  | 2.86 ± 0.94 | 0.864** |
| Normal| Horizontal | 3.47 ± 0.44 | 0.925** |
|       | Vertical  | 2.98 ± 0.75 | 0.896** |
| Slowly| Horizontal | 4.48 ± 0.56 | 0.786** |
|       | Vertical  | 3.19 ± 0.49 | 0.806** |

**Correlation is significant at the 0.01 level

10) Fang Wang et al. recorded the measurement of body sway in quiet standing posture by using a force plate, and the results showed that the maximum amplitude described by the COP in medio-lateral is 2.01 cm, and the maximum amplitude in the antero-posterior directions is 4.32 cm. Fang Wang et al. estimated body sway in terms of centroid trajectory by using an inexpensive webcam system. The results demonstrated that the average sway amplitude for anterior-posterior sway and lateral sway, respectively, were 14.04 cm and 13.8 cm (peak-peak). In this study, body sway exists in order to maintain a balanced position during standing up motion, and it can be captured by a motion capture system accurately, though it was difficult to measure using our system. Therefore, we hold that a part of the difference in COG position between our system and the motion capture system may be caused by body sway. According to this study, we can provide the parameters of joint angles for extension phase analysis by using the proposed...
system instead of a motion capture system. The proposed system for COG estimation is not recommended due to the lack of accuracy compared with motion capture system. Furthermore, the proposed system can be used in different applications that no need high level of accuracy but the advantage in this system is the usability compare to the motion capture system. For instance, the system valid to be used in the comparison and analysis of the difference of COG position before and after treatment, rather than using a motion capture system for standing up motion.

Further development of our system is needed in order to decrease errors caused by integration of acceleration data at different speeds, and improve the rigid link model which can separate the thoracolumbar region from trunk in order to increase the accuracy of COG estimation.

In clinical application, the initial angle of knee joint, ankle joint during standing up varies depending on the individual subject. Therefore, in the future study, the estimation system should be improved so that it can be used in different clinical cases because of different initial knee and ankle joint angle.

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