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Original Article

Examination of a practical method to quantitatively evaluate the rolling over movement in a clinical setting

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Abstract. [Purpose] We investigated the utility of wearable inertial and magnetic sensing modules for analyzing neck and trunk movements during the rolling over movement. [Participants and Methods] The participants were instructed to roll over from the supine to the side-lying position with three sensor units attached to their forehead, xiphoid process of the sternum, and abdomen. Experiments were conducted on two prescribed patterns: one emphasizing hip joint flexion and adduction, and the other focusing on scapular protraction and horizontal shoulder joint adduction in two healthy participants (one male and one female). The flexion and rotation angles of the neck and trunk were calculated using conventional spreadsheet software with data obtained from the sensors. The obtained values were compared for agreement with those derived from a three-dimensional (3D) motion analysis device. [Results] The cross-correlation coefficient for the flexion and rotation angles of the neck and trunk between the two measurement methods was approximately 0.85, and the root mean square (RMS) angle difference was approximately 5.0°. [Conclusion] Wearable inertial and magnetic sensors can be used to quantitatively evaluate neck and trunk movements during the rolling over movement.

Key words: Rolling over, Wearable inertial sensor, 3-dimensional motion analysis system

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INTRODUCTION

Rolling over from a supine position is a movement that requires posture change from a stable position to a mechanically unstable position[1)]. This movement poses challenges for many older individuals and patients because it requires coordinated motor control of the whole body[2)]. Thus, in treatment or movement guidance, it is important to determine the feasibility of performing this movement and compare it against a standard or norm[1)]. Compared to walking, where motor learning prioritizes either speed or energy conservation, rolling over allows a range of motion patterns as long as the movement is reasonably smooth[3)]. A previous study attempting to categorize rolling over in healthy adults captured 32 distinct combinations through video analysis, with the top four combinations occurring at a rate of 10%[4)]. Consequently, quantitative analysis often focuses on specific motion patterns[1)], making extracting comprehensive kinematic characteristics of normal operation challenging.

Miki and Nitta[5)] attempted to classify rolling-over patterns using cluster analysis based on trunk flexion and rotation measurements from a three-dimensional (3D) movement analysis device, revealing its kinematic properties. This classification, focusing on trunk motion, is expected to elucidate the consistent features of rolling over, which manifests diverse motion patterns[1, 5)]. However, the availability of 3D movement analysis devices necessary for evaluation is limited to only a few facilities. Moreover, the difficulty of measurement, stemming from markers positioned beneath the body, poses a significant

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challenge. Therefore, the benefits of evaluation must outweigh the associated costs to encourage widespread data collection. Therefore, a method has been suggested where motion is tracked using wearable inertial sensors[6], enabling the analysis of rolling over without constraining the measurement environment and at a lower cost. However, effectively analyzing 3D motion through data obtained from inertial sensors requires specialized knowledge. Consequently, using inertial sensors in clinical practice has not advanced significantly.

This study presents a method for estimating neck and trunk angles during rolling over using conventional spreadsheet software with data gathered from wearable inertial and magnetic sensors. Given reported disparities in neck rotational range of motion among elderly individuals who can and cannot roll over[7], we included the neck movement in our analysis. We then confirm the validity of wearable sensors in rolling-over analysis by comparing their results with those from a 3D motion analysis device to examine whether this method can be used as a easily estimation of kinematics of rolling over movement in the clinical setting.

PARTICIPANTS AND METHODS

Two volunteers participated (male: height 170.0 cm, female: height 155.0 cm) without orthopedic and neurologic disorders participated in the experiment. This study is an experiment of healthy participants with standard body type to verify the accuracy of the wearable inertial and magnetic sensors by referring to a similar study[8]. The volunteers tasked with executing rolling-over movements from the supine to the side-lying position in two prescribed patterns: one emphasizing hip joint flexion and adduction (rolling over from lower limb), and the other focusing on scapular protraction and horizontal shoulder joint adduction (rolling over from upper limb). Wearable inertial and geomagnetic sensors were employed to calculate neck and trunk angles and trunk velocity, with the obtained values compared in agreement with those derived from a 3D motion analysis device. This study has been approved by the Himeji Dokkyo University Bioethics Committee (#17-13). All participants provided their informed consent prior to participation[9].

A 3D motion analysis system (MAC 3D system, Motion Analysis Corp., Santa Rosa, CA, USA), comprising eight infrared cameras, was utilized to track marker trajectories during the rolling-over movements, operating at a sampling frequency of 200 Hz. Infrared-reflecting markers were placed at 42 locations on various body surfaces, including the top of the head, forehead, back of the head, xiphoid process of the sternum, anterior aspect of the abdomen, midpoint of the posterior superior iliac spines (PSIS), and lateral aspects of the head, acromioclavicular joints, front of cubital fossas, front of forearms, midpoint of the distal ends of the radius and ulna, palmar and dorsal surfaces of the hands, anterior superior iliac spines (ASIS), greater trochanters, the midpoint between ASIS and greater trochanters, lateral and medial epicondyles of the femur, front of patellas, lateral and medial malleoli, medial side of the first metatarsophalangeal joints, medial side of the five metatarsophalangeal joints, and dorsum of the foot. To ensure minimal interference with the rolling-over movement, custom-made flat infrared reflective markers were employed specifically for the lower part of the body.

Utilizing motion analysis software (Cortex-64 1.1.4, Motion Analysis Corp.), 3D kinematic calculations were conducted to determine neck and trunk angles, defined as the angles formed by the head, trunk, and pelvic segments. These segments were established by connecting four coordinates: the midpoint between the lateral parts of the heads, the midpoint between the clavicular heads, the midpoint between ASISs with PSIS, and the PSIS. Additionally, the estimation of body center of mass (COM) coordinates was achieved by referencing a prior study's defined mass and the location of the center of mass for each segment[10]. The composite component of the COM velocity across three axes was derived from the velocity obtained through the time-differentiation of these coordinates. The resulting data was filtered via a fourth-order zero lag, low-pass filter with a cutoff frequency set at 4 Hz.

Three wearable inertial and magnetic field sensors (LP-WS0942, Oisaka Electronic Equipment Ltd., Hiroshima, Japan) were employed to examine neck and trunk movement during a rollover. Each sensor, measuring 40 mm in length, 20 mm in width, 30 mm in height, and weighing 35 g, captured 3-axis acceleration (measurement range ± 5G), angular velocity (measurement range ± 300 dps), and magnetic field, following a right-hand orthogonal coordinate system. These sensor units were securely attached using a swimming cap and belt to three locations on the body surfaces: the forehead, the xiphoid process of the sternum, and the front of the abdomen at the level of the left and right iliac crests. Acceleration and angular velocity were sampled at 200 Hz, while magnetic field measurements were obtained at 50 Hz. All data streams were synchronized using an EMG logger (Oisaka Electronic Equipment Ltd.).

The positive direction for acceleration and magnetic field was established: the x-axis denotes the cranial direction, the y-axis represents the right-hand direction, and the z-axis indicates the vertical direction. Angular velocity around each axis is considered positive in the clockwise direction. When evaluating exercises using wearable inertial sensors, it is essential to correct the moving coordinate system, which is linked to sensor motion at each time point, to a static reference coordinate system[6]. In this study, we employed quaternions, denoted as $q=q_1i + q_2j + q_3k + q_4$ (with i, j, k representing imaginary axes), comprising one real number and three imaginary numbers to synchronize the three-axis angular velocities measured during motion[11]. Quaternions adeptly represent rotation in 3D space with four components, depicting rotation axis and angle. They offer an advantage over Euler angles by avoiding singularities, where errors amplify at specific rotation angles[12]. Converting from angular velocity to quaternions can be performed similarly to Euler angles, and the reverse conversion is relatively straightforward. The angles were calculated by expressing posture with quaternions and converting them back to Euler angles.

The conversion formula (1.1) from the Euler angle value to the quaternion was utilized to establish initial quaternion values, while equation (1.2) was employed for the initial Euler angle value[6, 11]. The orientation was determined by rotating the sensor sequentially along the z-axis → y-axis → x-axis, reflecting the sensor's installation position, expressed as roll, pitch, and angular rates (φ, θ, and ψ)[6]. Initial Euler angles φ_0 and θ_0 were derived from the fact that the sensor's acceleration (A_x, A_y, and A_z) in the rest state aligns with the absolute coordinate system's gravity of 1G in the vertical direction when stationary, with Coriolis acceleration owing to Earth's rotation disregarded. Since ψ_0 cannot be derived from gravity, it was inferred from the magnetic field, leveraging the equivalence between the yaw and azimuth angles in inertial sensor attitude measurement[6]. A correction formula based on the rotation matrix composed of φ and θ was applied to rectify tilt errors arising when the sensor is inclined. M_{xi}, M_{yi}, and M_{zi} represent magnetic fields corrected for tilt errors, while M_x, M_y, and M_z denote geomagnetism (formula (1.3))[6].

$$\begin{pmatrix} q_1 \\ q_2 \\ q_3 \\ q_4 \end{pmatrix} = \begin{pmatrix} \cos\left(\frac{\phi}{2}\right)\cdot\cos\left(\frac{\theta}{2}\right)\cdot\cos\left(\frac{\psi}{2}\right)+\sin\left(\frac{\phi}{2}\right)\cdot\sin\left(\frac{\theta}{2}\right)\cdot\sin\left(\frac{\psi}{2}\right) \\ \sin\left(\frac{\phi}{2}\right)\cdot\cos\left(\frac{\theta}{2}\right)\cdot\cos\left(\frac{\psi}{2}\right)-\cos\left(\frac{\phi}{2}\right)\cdot\sin\left(\frac{\theta}{2}\right)\cdot\sin\left(\frac{\psi}{2}\right) \\ \cos\left(\frac{\phi}{2}\right)\cdot\sin\left(\frac{\theta}{2}\right)\cdot\cos\left(\frac{\psi}{2}\right)+\sin\left(\frac{\phi}{2}\right)\cdot\cos\left(\frac{\theta}{2}\right)\cdot\sin\left(\frac{\psi}{2}\right) \\ \cos\left(\frac{\phi}{2}\right)\cdot\cos\left(\frac{\theta}{2}\right)\cdot\sin\left(\frac{\psi}{2}\right)-\sin\left(\frac{\phi}{2}\right)\cdot\sin\left(\frac{\theta}{2}\right)\cdot\cos\left(\frac{\psi}{2}\right) \end{pmatrix} \quad (1.1)$$

$$\phi = \tan^{-1}\frac{A_y}{A_z}, \quad \theta = \tan^{-1}\frac{-A_x}{\sqrt{A_y^{\,2}+A_z^{\,2}}}, \quad \psi = \tan^{-1}\frac{-M_{yi}}{M_{xi}} \quad (1.2)$$

$$\begin{pmatrix} M_{xi} \\ M_{yi} \\ M_{zi} \end{pmatrix} = \begin{pmatrix} \cos\theta & \sin\phi\sin\theta & \cos\phi\sin\theta \\ 0 & \cos\phi & -\sin\phi \\ -\sin\theta & \sin\phi\cos\theta & \cos\phi\cos\theta \end{pmatrix} \begin{pmatrix} M_x \\ M_y \\ M_z \end{pmatrix} \quad (1.3)$$

To measure quaternion attitude during motion, we applied the conversion formula (2.1) to derive the time derivative of the quaternion from angular velocities ω_x, ω_y, and ω_z. During the process of sequentially adding the obtained rotational changes to update the 3D attitude, there is a risk of deviating from the normal condition of quaternion $q_1^2 + q_2^2 + q_3^2 + q_4^2=1$ owing to numerical integration[11]. Thus, normalization was conducted post-integration. Subsequently, the acquired data were converted back to Euler angles utilizing formula (2.2), from which φ, θ, and ψ were calculated[10, 11].

$$\begin{pmatrix} \dot{q}_{1(t)} \\ \dot{q}_{2(t)} \\ \dot{q}_{3(t)} \\ \dot{q}_{4(t)} \end{pmatrix} = \frac{1}{2} \begin{pmatrix} -q_{4(t)} & -q_{3(t)} & q_{2(t)} \\ q_{3(t)} & -q_{4(t)} & q_{1(t)} \\ -q_{2(t)} & q_{1(t)} & -q_{4(t)} \\ q_{1(t)} & q_{2(t)} & q_{3(t)} \end{pmatrix} \begin{pmatrix} \omega_{x(t)} \\ \omega_{y(t)} \\ \omega_{z(t)} \end{pmatrix} \quad (2.1)$$

$$\phi = \tan^{-1}\left(\frac{2(q_1q_2+q_3q_4)}{1-2(q_2^{\,2}+q_3^{\,2})}\right), \quad \theta = \sin^{-1}\big(2(q_1q_3-q_4q_2)\big), \quad \psi = \tan^{-1}\left(\frac{2(q_1q_4+q_2q_3)}{1-2(q_3^{\,2}+q_4^{\,2})}\right) \quad (2.2)$$

The neck angle was determined by the difference between angles from the inertial sensor positioned on the forehead and the xiphoid process of the sternum. The trunk angle was deduced from differences in angles from inertial sensors placed on the xiphoid process of the sternum and the front of the abdomen, aligned with the left and right iliac crests. Furthermore, trunk velocity was calculated by integrating acceleration measured by the sensor affixed to the front of the abdomen because the body's COM typically lies slightly in front of the sacrum at approximately 55% of adult body height, and rolling over entails limb movements initiated from the supine position in front of the body. The resulting data was filtered using a fourth-order zero lag, low-pass filter with a cutoff frequency set at 4 Hz.

Data analysis interval ranged from the beginning to the conclusion of the motion, with movement time normalized to 100%. The start of a movement was determined when any of the limbs began to move, and the end was determined when the trunk movement ended. The alignment between time-series data of joint angles and COM velocity, derived from the inertial sensor and the 3D motion analysis device, was assessed via a cross-correlation function (CCF). It is a function expressing to what extent arbitrary two time series data are similar. If the results of the array of functions are all ± 1, there is a correlation and if all the results are zero, there is no correlation. Angle differences were quantified using the root mean square (RMS). Data processing was performed using Microsoft Office Excel 2010 (Microsoft Japan Inc., Tokyo, Japan), 3D graphs were generated using Origin Pro 2023 (Light stone Inc., Tokyo, Japan), and statistical analysis was performed using IBM SPSS Statistics 21 (IBM Japan Ltd., Tokyo, Japan), with significance set at $p<0.05$.

RESULTS

Table 1 shows the results for the agreement in the flexion and rotation angles of the neck and trunk, and COM velocity between the two measurement methods. Figure 1 illustrates the representative example time series of the neck and trunk angles, COM coordinates and COM velocity. The mean CCF ± standard deviation (SD) for the eight joint angles between the two measurement methods was 0.83 ± 0.14 for male and 0.85 ± 0.11 for female. The mean RMS ± SD for the eight joint angles between the two measurement methods was 4.9 ± 2.0° for male and 5.3 ± 2.1° for female. Meanwhile, the mean value ± SD of the maximum COM speed was 0.26 ± 0.05 m/s as determined by the inertial and magnetic sensors and 0.26 ± 0.05 m/s using the 3D motion analysis device. Furthermore, the mean CCF ± SD was 0.89 ± 0.04.

Table 1. CCF and RMS of the neck and trunk angles and CCF of COM velocity between the two measurement methods

Participant	Leading limb	Neck				Trunk				COM velocity
		Flexion		Rotation		Flexion		Rotation		
		CCF	RMS	CCF	RMS	CCF	RMS	CCF	RMS	CCF
Male	Lower	0.98	6.7°	0.81	4.0°	0.98	4.2°	0.93	4.2°	0.87
	Upper	0.76	8.4°	0.78	6.4°	0.88	2.4°	0.57	3.5°	0.96
Female	Lower	0.78	4.4°	0.71	9.4°	0.95	2.0°	0.94	6.8°	0.88
	Upper	0.98	5.4°	0.83	5.5°	0.69	4.4°	0.91	4.8°	0.87

CCF: cross-correlation function; RMS: root mean square of the angle differences; COM: center of mass.

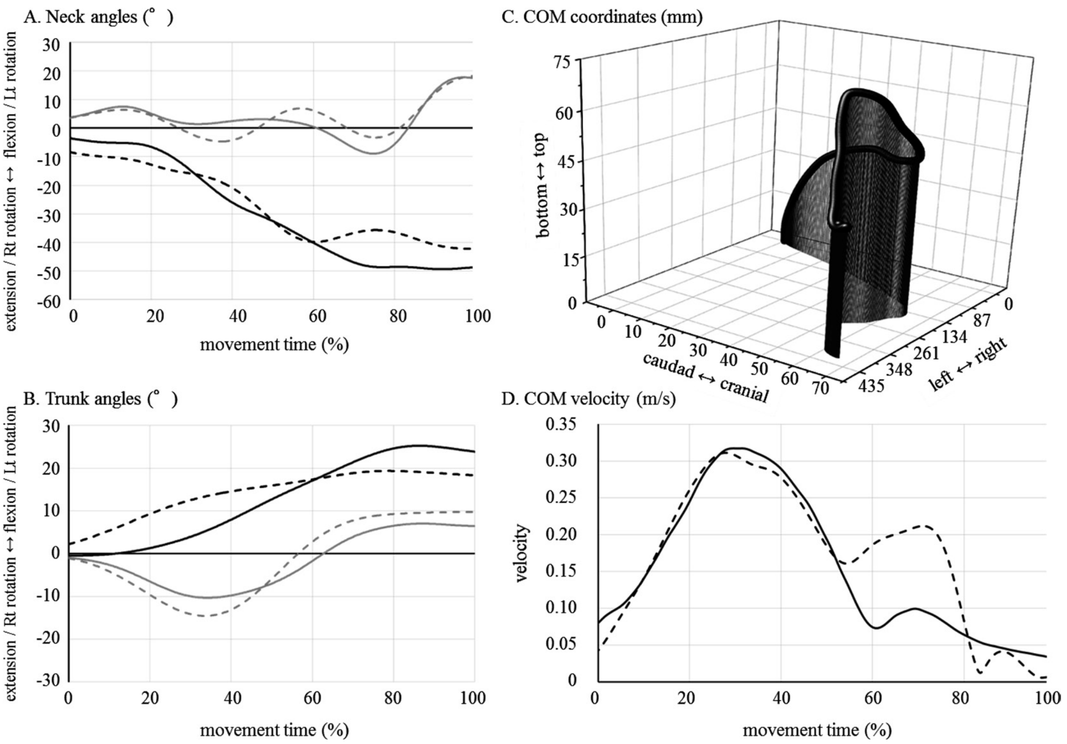

Fig. 1. Representative example time series of the neck and trunk angles, COM coordinates and COM velocity.
The figure shows the neck and trunk angle, COM coordinates and COM velocity during rolling over from lower limbs for a man. A is the neck angles (solid line=inertial sensor; dashed line=3D motion analysis device; dark line=flexion; light line=rotation). B is the trunk angles (solid line=inertial sensor; dashed line=3D motion analysis device; dark line=flexion; light line=rotation). C is the 3D coordinate of the COM estimated by the 3D motion analysis device. D is the 3-axis composite component of the COM velocity (solid line=inertial sensor; dashed line=3D motion analysis device). COM: center of mass.

DISCUSSION

In this study, we introduced a methodology for estimating neck and trunk angles and COM velocity during rolling-over movements, employing wearable inertial and magnetic sensors positioned on the forehead, sternum, and front of the abdomen. Furthermore, we assessed the sensor's efficacy compared to the 3D motion analysis device. Our findings revealed a high degree of agreement in the flexion and rotation angles of the neck and trunk between the two measurement methods, approximately 0.85 in CCF and approximately 5.0° in RMS. In a previous study, the agreement level of flexion angles of hip and knee joints during walking was determined by inertial sensors attached to the pelvis and lower legs, and the 3D motion analysis device was approximately 0.90 in CCF and 8.0° in RMS[8)]. Despite differences in movement and joint dynamics compared to our study, commonalities exist. Both studies calculated joint angles using sensors placed across the joints, with agreement assessed against the 3D motion analysis device. While some degree of angle discrepancy between the inertial sensor and 3D motion analysis is inevitable owing to sensor placement errors or sensor movement[13)], our findings support the potential use of inertial sensors for quantitatively evaluating neck and trunk movements during rolling over. Our results demonstrate a comparable level of accuracy to the previous study[8)].

Furthermore, the agreement level of the COM velocity between the two analysis methods was approximately 0.90 in CCF. During the process of rolling over on the floor, rotation of the pelvis—connecting the upper and lower body—is essential to complete the postural shift, facilitating trunk rotation while coordinating the movement of the head and limbs in front of the body. Substantial force is required to initiate movement and change the support base from a mechanically stable supine position. Our study demonstrates the feasibility of estimating COM velocity generated by the rotational force of the body, utilizing trunk velocity data acquired from a sensor affixed to the front of the abdomen. However, as illustrated in the latter half of movement in Fig. 1D, when motion transitions from lower limbs and pelvis to upper limbs and head, there may be limitations in capturing the untwisting rotation of the upper trunk relative to the initially moved lower trunk using an inertial sensor positioned on the abdomen. This issue warrants consideration when estimating COM velocity based on trunk velocity.

When analyzing rolling over, quantitative assessment of neck and trunk angles and COM velocity is instrumental in understanding the range of motion and required force for executing the movement, elucidating factors contributing to its complexity. However, the challenge lies in mitigating the burdens associated with data collection. Wearable inertial sensors offer a cost-effective solution, facilitating continuous movement tracking across varied environments. As equipment capabilities advance, the labor and complexity of data analysis are expected to decrease further. While our study demonstrates the utility of wearable inertial sensors in analyzing neck and trunk motion during rolling over, it is important to note that these findings are based on typical movement patterns of standard body types among male and female. In the future, we anticipate that comprehensive data collection across diverse participants will enhance the efficacy of quantitative analysis using inertial sensors, ultimately benefiting treatment and movement guidance.

Conflict of interest

The authors have no conflicts of interest to declare.

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