Loss of Mechanical Energy Efficiency in the Sit-to-stand Motion of Acute Stroke Patients

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Abstract The purpose of this study was to demonstrate the usefulness of a small inertia sensor for quantitative classification of movement disorders based on the change in mechanical energy in patients following a stroke. We measured the sit-to-stand motion in acute stroke patients using inertial sensors in a small clinic. Three acute stroke patients and three healthy adults performed the sit-to-stand paradigm. The three-dimensional angle in the global coordinate system of the inertial sensor attached to the participant’s body was then calculated. The movements of healthy adults were measured using inertial sensors and a camera motion capture system simultaneously, and only sagittal plane angles were used for the analysis, which were similar in the two devices. Subsequently, link segment models were created, and the mechanical work until seat-off was calculated. In stroke patients, the thoracic potential energy was not converted to kinetic energy, and deceleration of the thorax was greater in stroke patients than in healthy adults. Furthermore, the mean pelvic kinetic energy in stroke patients was approximately one tenth of that in healthy adults. In healthy adults, the waveforms of the angular velocities of the thorax and pelvis were synchronized. Such synchronization was not observed in the waveforms of stroke patients. A reason for the low pelvic kinetic energy in stroke patients is the fact that deceleration of the thorax by lumbar muscles does not lead to acceleration of the pelvis. The lack of synchronization of thoracic and pelvic angular velocities reduced the energy transfer efficiency. The usefulness of a small inertial sensor was demonstrated based on the evaluation of energy change efficiency during the sit-to-stand motion performed by an individual following a stroke.

Keywords: inertia sensor, motion analysis, mechanical energy, stroke.


1. Introduction

Movement disorder caused by a stroke is the leading reason for nursing care [1]. Quantitative classification of this movement disorder is strongly demanded by the society. Therapists often use observational evaluation for the diagnosis of stroke patients [2]. However, it is difficult to quantify movement disorder just by observing the patient. Furthermore, only observing movement does not reveal the disorder of the kinetic parameter causing the resultant movement.

Researchers often evaluate the motion of healthy adults using a camera motion capture system [3, 4]. Such motion-measuring devices are often expensive and require a large measuring space. A small inertial sensor is a motion-measuring device that can be used regardless of the measurement space. The development of small inertial sensors is extensive. Many studies have verified the measurement accuracy of inertial sensors by comparing with a camera motion capture system [5, 6]. However, few previous studies measured the motion of severe acute stroke patients using inertial sensors in clinics.

In daily motion, sit-to-stand is important as a habitual action prior to walking. One of the reasons for people with disabilities not able to stand up by themselves is the existence of the instantaneous joint moment [7]. The joint moment reaches its maximum value immediately
after seat-off [4]. Therefore, the sit-to-stand motion often fails with sit-back at seat-off [8]. Previous studies have revealed that the pelvic movement supplies mechanical energy to assist this joint moment [9]. We hypothesized that lack of this mechanical energy largely reflects movement disorders caused by stroke.

In this study, we measured the sit-to-stand motion in acute stroke patients using inertial sensors in a small clinic. The purpose of this study was to demonstrate the usefulness of small inertial sensors for quantitative classification of movement disorders by clarifying how the mechanical energy is changed by stroke.

2. Methods
2.1 Subjects
This study evaluated three acute stroke patients with hemiparesis and three healthy adults. All patients were male. The mean (standard deviation) time at which these patients were studied after the stroke was 8.2 (4.2) days. Other patient characteristics are listed in Table 1. The data are expressed as mean (standard deviation). The inclusion criteria were as follows: (1) first ischemic/hemorrhagic stroke, (2) unilateral weakness, and (3) < 2 weeks post-stroke. The exclusion criteria were as follows: (1) other neurological or orthopedic conditions, (2) major cognitive deficits as assessed by standard tests, and (3) sitting balance or trunk stability deficits.

This study was conducted in accordance with the ethical principles of the Helsinki Declaration after obtaining written informed consent from each subject. The study was approved by the ethics review committee in Higashi Saitama General Hospital and the ethics review committee in Saitama Prefectural University.

2.2 Tasks
The subjects performed the sit-to-stand task from a sitting posture on a chair. We tried to standardize the start posture for all subjects. The seat height was adjusted so that the thigh was tilted 25° forward from the floor and the shank was tilted 25° forward from the perpendicular axis to the floor. The thorax and pelvis were adjusted, attempting to keep in alignment with the perpendicular axis to the floor, and these were set as the initial angles of the inertial sensor. After measurements, the mean (standard deviation) initial angles obtained from the three-dimensional motion analysis system in healthy subjects were: thorax, 4.08 (5.90)°; pelvis, 4.33 (6.57)°; thigh, 27.67 (2.73)°; and shank, 26.74 (3.16)°. Although we tried to standardize the start posture of stroke patients, standardization of their thorax posture was difficult because of disabilities. The mean (standard deviation) initial angles measured by the inertia sensors in stroke patients were: pelvis, 1.78 (2.59)° and thorax, 10.09 (4.49)°. In stroke patients, the initial angle of the pelvis was close to the perpendicular axis to the floor as in healthy adults, but the initial angle of the thorax was tilted forward compared to healthy adults. During the sit-to-stand motion, the upper limbs were beside the body, and the paradigm was conducted to avoid contact with the support surface. The subjects were asked to stand at a comfortable speed after receiving a signal from the experimenter. Each subject performed three trials. The data of each trial were included in the analysis without averaging.

2.3 Kinematic data acquisition
To acquire kinematic data, we used small nine-axis inertial sensors, including a three-axis gyroscope, three-axis accelerometer, and three-axis magnetometer (Oisaka Electronics Device Ltd., Hiroshima, Japan). The product name was "Small 9-axis wireless motion sensor", and the model number was OE–WS0905. The measurement range was 16 G for the acceleration sensor, 1500°/s for the gyro sensor, and 8 G for the geomagnetic sensor. The size was 40 mm × 30 mm × 20 mm, and the weight was 30 g. These sensors were wireless-controlled. Each sensor was attached to six sites of the subject's body: thorax, pelvis, both shanks, and both thighs (Fig. 1). Data were sampled at 100 Hz.

Then, the rotation matrix in the absolute space coordinate system of each sensor was obtained using commercial software (Pose estimation application; LP–WSAP03, Oisaka Electronics Device Ltd., Hiroshima,
Japan. Conversion from each rotation matrix to segment angle and creation of link segment models were performed using custom programs of MATLAB (MathWorks, Inc., Massachusetts, USA).

The movement of healthy adults was also studied using a camera motion capture system at 100 Hz (Vicon Motion Systems, Inc., Oxford, UK). The data were used to calculate the error for the inertial sensor.

We extracted the measurement data during the sit-to-stand motion. The starting point was identified when any of the joint angular velocities started to change continually. The end point was identified when all the joint angular velocities became equal to 0 rad/s. The seat-off timing was identified as the time when the thigh angular velocity started to change continually. The definition of seat-off timing was in accordance to a previous study [9].

We compared the thoracic and pelvic angles calculated from the camera motion capture system with those obtained from the inertial sensor in healthy adults. The mean (standard deviation) Pearson’s correlation coefficient was 0.94 (0.12) on the sagittal plane, 0.34 (0.46) on the frontal plane, and 0.23 (0.36) on the horizontal plane. Therefore, we used only sagittal plane angles that were well matched in both instruments.

2.4 Mechanical energy calculation

The potential energy $E_p$ of segment $i$ at data point $j$ was calculated from Equation (1).

$$E_p_{ij} = m_i g h_{ij}$$

where $m$ is the segment mass, $g$ is the gravitational acceleration, and $h$ is the height of the segment center of mass.

The kinetic energy $E_k$ of segment $i$ at data point $j$ was calculated from Equation (2).

$$E_k_{ij} = 1/2 m_i v_{ij}^2 + 1/2 I_i \omega_{ij}^2$$

where $v$ is the translation velocity of the segment center of mass, $I$ is the inertia moment of the segment, and $\omega$ is the angular velocity of the segment.

In this study, we used body measurement values, i.e., the mass, position of the center of mass; and the inertial moment of each segment as reported by Kodama et al. [10]. These authors modified the link parts defined in the previous research conducted by Ae et al. [11], and performed inverse dynamics calculations using an accelerometer to guarantee increased accuracy.

2.5 Analysis

First, we calculated the general parameters of the sit-to-stand motion, including the total duration, and thoracic and pelvic angles at seat-off. We calculated these parameters because they are related to the total amount of energy.

Then, we calculated the following parameters from kinematic data and mechanical energy.

1. $E_p - E_k_{\text{thorax}}$: Difference between the potential energy and kinetic energy change from the onset of the motion to seat-off in the thorax.

2. $E_k_{\text{pelvis}}$: Change in the kinetic energy from the onset of the motion to seat-off in the pelvis.

3. $CCF_{\text{thorax-pelvis}}$: Cross-correlation coefficient between the thorax and pelvis angular velocities from the onset of the motion to seat-off.

For these, we applied the independent $t$-test for statistical comparison between groups.

We quantified the amount by which the antagonist inhibited the conversion of potential to kinetic energy in the thorax, and the kinetic energy obtained by the pelvis, including its muscle activity. This suggests how the energy transfer from the thorax to the pelvis differs between stroke patients and healthy adults. Because this energy transfer is maximized when the angular velocity is synchronous; i.e., when isometric contraction is executed, the temporal correlation between the thoracic and pelvic angular velocities was also quantified.

3. Results

3.1 Comparison of parameters

The results of general parameters are listed in Table 2.

The total duration for the sit-to-stand motion was longer in stroke patients than in healthy adults. Despite the fact that we subtracted the initial angle of the thorax at the start posture, the stroke patients showed larger amount of thoracic angular displacement until seat-off.

The results of parameters of mechanical energy are listed in Table 3. The data are expressed as mean (standard deviation).

Figure 2 shows a representative trial of the temporal pattern of potential energy. The upper graph represents a

### Table 2 Results for general parameter.

<table>
<thead>
<tr>
<th></th>
<th>Healthy</th>
<th>Stroke</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total duration [sec]</td>
<td>3.29 (0.83)</td>
<td>6.11 (0.15)</td>
</tr>
<tr>
<td>$\theta_{\text{thorax}}$ [°]</td>
<td>38.51 (6.97)</td>
<td>87.86 (27.45)</td>
</tr>
<tr>
<td>$\theta_{\text{pelvis}}$ [°]</td>
<td>22.38 (0.04)</td>
<td>27.95 (6.40)</td>
</tr>
</tbody>
</table>

*p < 0.001

### Table 3 Results for mechanical energy.

<table>
<thead>
<tr>
<th></th>
<th>Healthy</th>
<th>Stroke</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_p - E_k_{\text{thorax}}$ [J/kg]</td>
<td>0.22 (0.11)</td>
<td>0.81 (0.21)</td>
</tr>
<tr>
<td>$E_k_{\text{pelvis}}$ [J/kg]</td>
<td>0.05 (0.01)</td>
<td>0.002 (0.002)</td>
</tr>
<tr>
<td>$CCF_{\text{thorax-pelvis}}$</td>
<td>0.95 (0.04)</td>
<td>0.30 (0.37)</td>
</tr>
</tbody>
</table>

*p < 0.001
healthy adult, and the lower graph represents a stroke patient. Figure 3 shows the group mean (standard deviation is depicted as shaded area) for potential energy of the thorax and pelvis. Because the focus of this study was seat-off, we normalized and averaged the movement time for seat-off as 50% time (the same hereinafter). Both healthy adults and stroke patients had the lowest potential energy at the time of seat-off. Note that the normalized potential energy of the thorax was smaller in stroke patients than in healthy adults, as opposed to the normalized potential energy of the pelvis which was larger in stroke patients than in healthy adults. This occurred because the start posture of the thorax of stroke patients was tilted forward compared to healthy adults.

Figure 4 shows a representative example (the same subject and trial as in Fig. 2) of the temporal pattern of kinetic energy. The upper graph represents a healthy adult, and the lower graph represents a stroke patient. Figure 5 shows the group mean (standard deviation is depicted as shaded area) for kinetic energy of the thorax and pelvis. The vertical axis for stroke patients (lower graph) is one-tenth of that for healthy subjects (upper graph).

Figure 6 shows a representative example (same subject and trial as in Fig. 2 and Fig. 4) of the temporal pattern of angular velocity. The upper graph represents a healthy adult, and the lower graph represents a stroke patient. Figure 7 shows the group mean (standard deviation is depicted as shaded area) for angular velocity of thorax and pelvis. For healthy adults, the waveforms of the thorax and pelvis are synchronized. In contrast, no such synchronization is observed for stroke patients.
4. Discussion

First, we discuss the loss of energy transfer from the thorax to pelvis in stroke patients. The trunk flexion during the sit-to-stand motion resembled a passive inverted pendulum. However, even in healthy adults, the potential energy and kinetic energy were not transformed equally. That is, the change in height of the thorax due to gravity was not completely converted to acceleration of the thorax itself. This deceleration may be caused by contraction of the antagonist—the lumbar muscle. On the other hand, contraction of the lumbar muscles leads to pelvic movement, resulting in increased kinetic energy in the pelvis. This energy transfer is most efficient when the angular velocities of the thorax and the pelvis are synchronized (because the muscle length is kept constant [12, 13]). In stroke patients, the thoracic potential energy was not converted to kinetic energy in the same manner as healthy adults. The thorax was decelerated more compared to healthy adults. Furthermore, the mean pelvic kinetic energy in stroke patients was approximately one-tenth of that in healthy adults. Part of the reason for this phenomenon is that deceleration of the thorax by the lumbar muscles did not lead to acceleration of the pelvis in stroke patients. The lack of synchronization between the thoracic and pelvic angular velocities reduced the energy transfer efficiency.

Second, we must mention the loss of the total amount of energy. The start posture was different between healthy adults and stroke patients. Because the thorax of stroke patients was tilted forward in the start posture, both the potential energy and kinetic energy could decrease. Not only kinematics of sit-to-stand motion but also start posture were related to the loss of total amount of energy, and both of which might reflect movement disorders of stroke. However, in addition to this fact, we must mention that the reason for the reduced pelvic kinetic energy cannot be explained only by the greater forward tilting of the thorax in the start posture (i.e., decrease in thoracic potential energy) in stroke patients. The thoracic potential energy in stroke patients was smaller, but the difference between potential and kinetic energy was larger than that in healthy adults. Of course, the loss of energy transfer due to lack of synchronization between the thoracic and pelvic angular velocities contributed to this difference. In addition, the total duration of the sit-to-stand motion was longer in stroke patients. Even if the amount of angular displacement is large, the kinetic energy is small unless it moves at a sufficient speed (see also Fig. 7). Because stroke patients performed the sit-to-stand motion slowly, kinetic energy was considered to be smaller than that in healthy adults.

Third, we describe how the loss of pelvic kinetic energy makes it difficult to perform the sit-to-stand motion. Kinetic energy in the pelvis is important because it supplies the energy for thigh-lift that occurs at seat-off. In a previous study [9], energy transfer from the pelvis to the thigh was verified by negative and positive work done by the joint moment on each segment. Healthy subjects performed sit-to-stand motion under normal comfortable condition and the condition with the trunk tilted forward during the motion. As a result, the negative and positive work performed by the hip joint moment to the distal pelvis and proximal thigh was reduced to a greater extent under the condition with the trunk tilted forward during the motion, despite greater increase in the hip joint moment. This means that a decrease in pelvic kinetic energy leads to a decrease in kinetic energy in the thigh, making

![Fig. 6 Angular velocity (representative data).](image1)

![Fig. 7 Angular velocity (group mean).](image2)
Last, we summarize the significance of measuring the loss of energy transfer and the loss of total energy caused by stroke. Although the pelvic kinetic energy at seat-off was less than one-tenth of the weight, difference between subjects with disability and normal subjects was considered significant in a previous study [14]. This previous study used magnetic sensor for measurement, which was different from our study. The experimental conditions were similar to our experiment except that the trunk was in the upright position in both groups. They compared healthy adults and low back pain patients. The mean amplitudes were 0.13 J/kg in the control group and 0.17 J/kg in the patient group. Patients with low back pain had increased lumbar muscle activity to suppress the pain caused by hip movements [15]. The difference of 0.04 J/kg was associated with the supply of extra energy from muscle activity. The difference between two groups was 0.048 J/kg in our study, which can be regarded as a considerable difference, as was reported in the previous study. On the other hand, no research has quantified the loss of energy transfer and the loss of total energy during the sit-to-stand motion in stroke patients. Previous study revealed the loss of energy in stroke patients during walking, which is representative habitual action [16]. The investigators used a camera motion capture system to observe chronic stroke patients and revealed the loss of energy from the joint moment work of the whole body (difference between groups: 0.08 J/kg). Because such loss of energy correlated with the walking speed, they regarded the loss of energy as an important indicator that sensitively reflects the rehabilitation process of walking in stroke patients. However, stroke patients, especially patients in the acute phase, often have difficulty with walking due to movement disorder. Our study is important because it clarifies the loss of energy transfer and the loss of total energy during the sit-to-stand motion, which is a habitual action prior to walking. This study suggests that the loss of energy transfer and the loss of total energy are useful as quantitative measures for movement disorders in acute stroke patients who have difficulty with walking. In addition, the previous study that revealed the loss of energy in stroke patients used a camera motion capture system [16]. Our study used small inertial sensors that are not restricted by measurement space. Therefore, the results presented here can be the basic knowledge for constructing a measurement system in small clinics.

However, this study has some limitations. First, the mean age of the control group in this study was younger than that of stroke patients. Because physical function tends to deteriorate with age, even healthy elderly people may choose the movement strategy to compensate for physical function. However, in previous studies, the movement time and segmental angular velocity, which were analyzed in this study, were not different between young and elderly people [3, 17]. For reference, the average (standard deviation) movement time of young people included in this study was 3.18 (0.40) s, which was higher than the time [2.13 (0.36) s] of elderly people reported in a previous study. The age difference found in the previous study was the parameter after seat-off [3], and this difference was not directly related to seat-off. Therefore, the biomarkers associated with the control group in this study may be reliable for comparison of subjects with stroke patients. Furthermore, stroke patients in this study were able to perform the sit-to-stand motion independently. In the future, to standardize the rehabilitation evaluation, it is necessary to examine patients with severe hemiparesis who fail to perform sit-to-stand motion.

5. Conclusion

By measuring the sit-to-stand motion in acute stroke patients using a small inertial sensor, we revealed the loss of mechanical energy efficiency in stroke patients. This result demonstrated the usefulness of the small inertial sensor that can easily measure the movement of patients in a small clinic.

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References


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