Rapid publication

Etiology of Head Injuries due to Falls in Clinical Situations, and Nursing Care to Preventing Injuries

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Abstract

The purpose of this study is to examine the causes behind falling accidents that are directly encountered by nurses, which are associated with a high rate of injury. Falls often result in physical trauma, and this type of accident is common in medical care facilities.

The present research uses crash test dummies as models for investigating human posture during falls and verifying predictions of the severity of the resulting head injury.

Fall simulations were based on preliminary experiments conducted at the Japan Automobile Research Institute and performed under conditions commonly seen in medical care facilities.

The fall was directed rearward without defensive posturing onto a tile floor with head injury criterion (HIC) 100.4 at 42.1 G collision acceleration and onto a linoleum floor with HIC 80.2 at 37.2 G collision acceleration. In both cases, we found an extremely high risk of cranial fracture or other direct injury. Angular acceleration of the head during the fall reached 4,371 rad/s^2, respectively. This indicates a high risk of concussive trauma such as acute epidural hematoma.

These results indicate that falls pose a serious risk in medical care facilities. To reduce the incidence of head injury due to falls by patients for whom adopting a defensive posture is difficult, such as those who are awake at night or have dementia, both direct head trauma and rotation of the head and neck should be reduced.

These results reveal that falling is associated with a very high risk of direct injury, including head fractures. Maximum angular acceleration of the head during a postural change accompanying falling was estimated at 4,371 rad/sec^2. These values indicate a very high risk of concussion injury. Therefore, protecting not only potential contact surfaces as well as the neck is necessary to prevent head injury due to a fall.

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—Key words—

fall accidents, direct injuries, concussion injuries

Introduction

Accidents due to falling are a major problem in many medical institutions, and each institution adopts various measures to prevent and attend to such incidents. The issue of falls is also considered an important topic in the field of education relating to medicine and medical insurance. The number of original articles published in medical journals over the last 5 years that mention the keyword "fall" is as high as 3,553. Thus, the issue of falling accidents is an important subject not only in clinical nursing but also in educational research.

The Japan Council for Quality Health Care reported 2,181 medical accidents during the last fiscal year, of which 492 were falling accidents; in general, medical accidents due to falls represent approximately 40% of all medical accidents. Serious injuries that are commonly reported as a result of falls include trans cervical fractures and acute extradural hematomas[1]. According to medical accident surveys, most of the 10 deaths caused by falling accidents during the last fiscal year in Japan were due to head injuries. However, the initial conditions and circumstances pertaining to the head injuries caused by falls remain unclear. Moreover, no previous studies have clearly evaluated the risk of head injury that is associated with falling accidents. Thus, the mechanism by which death occurs due to falls is not clear. The issue of falling accidents has been attracting significant social interest, and the number of trials that are related to death due to falling has been increasing every year. Furthermore, 4 guilty verdicts were handed down in such trials between 2000 and 2009[2]. Therefore, falling accidents are a serious issue for which healthcare professionals are also held socially responsible.

According to previous studies, the administration of a psychotropic or hypnotic drug has, in many cases, decreased the level of awareness, which resulted in the patient's fall[3)~5)]. In addition, it has been observed that, even among healthy individuals, a person who falls unintentionally is generally unable to assume a defensive posture[6]. Hence, a reduced level of awareness or dementia due to waking at night or due to the use of hypnotic drugs is expected to delay postural defenses and increase the risk of a direct, strong blow to the head. Moreover, head injuries due to falling backwards have a high risk of developing into a serious injury. Therefore, in order to avoid falling down, it is important to have an accurate understanding of the risk of head injury resulting from falls, when providing nursing care.

Head injuries due to falls may be classified as direct injuries due to direct impulse, and as concussion injuries due to shear stress-induced deformation of the brain parenchyma as a result of twisting the neck forward, backward, or sideways. Reported direct injuries include cerebral contusions and acute extradural hematoma (ASDH), and reported concussion injuries include diffuse axonal injury (DAI)[7,8].

Therefore, in this study, we have examined conditions associated with the most serious head injuries, i.e., head injuries resulting from falling backward. We measured the risk of direct head injury and concussion injury due to falls, through calculating changes in the resultant acceleration, angular acceleration at the dummy head's center of gravity.

Overview of the experiment

We used Japan Automobile Research Institute (JARI) facilities to conduct preliminary fall experiments using crash test dummies[9]. These dummies model the human body well and are used for injury evaluation in automobile crash tests. We created a dummy based on the results of these experiments and used this dummy in the fall simulations. The results of those tests were used to evaluate the risk of head injury in falls. We investigated tile and linoleum flooring, both commonly found in medical facilities, and measured head velocity and posture changes when the dummy was allowed to fall in a rearward direction. From this, we calculated the collision acceleration (G), head injury criterion (HIC), and angular acceleration of the intracranial area of the dummy.

Equipment used

▪ Model dummy: We created a human-shaped dummy that faithfully reproduces the balance and weight of body parts; this dummy was based on skeletal preparations. The dummy is portable, allowing experimentation in actual medical care environments. Measured values related to head injury agreed well with previous studies, indicating a high degree of reproducibility for head injury measurement in an actual human body. Fig. 1 and Table 1 are created a dummy's parameters.

▪ High speed camera: The motion of the model dummy at the time of a fall impact was photographed using a Phantom miro-ex of Nobby-tech. Ltd., with a frame rate of 1/1,000 sec.

▪ Motion analysis software: The measured value acquired from an accelerometer inside the dummy was analyzed using PCC software Ver.2.14.727.110 of Nobby-tech.Ltd.. and the HIC value at the time of fall was obtained from the result.

▪ Three-axis accelerometer: The measured value acquired from an accelerometer inside the dummy was analyzed using CXL-25 G-P3 of Global Leader. Ltd.

Fig. 1 shows the model dummy and weight rates of each body part are shown in Table 1.

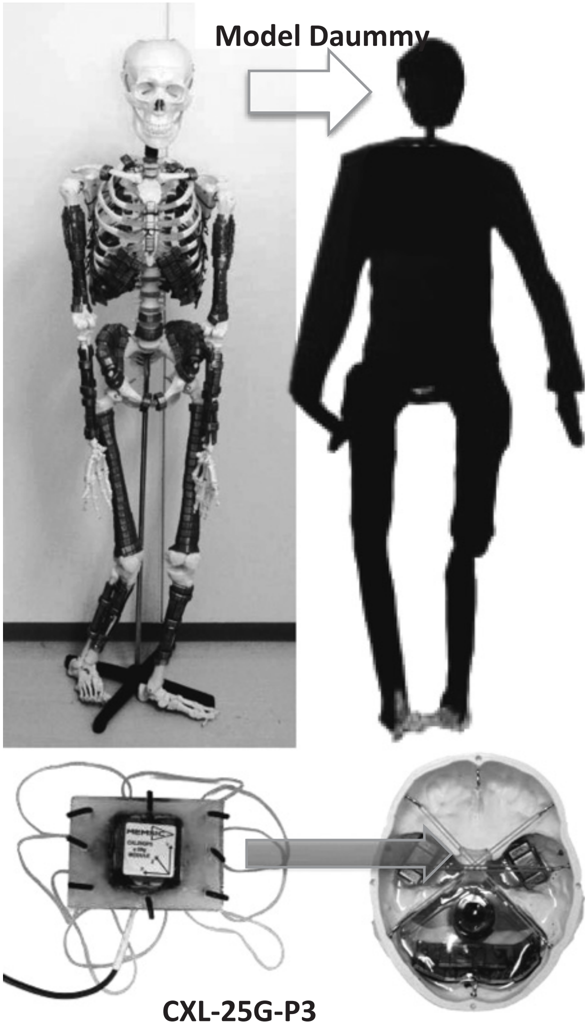

Fig. 1 Model Dummy Equipment

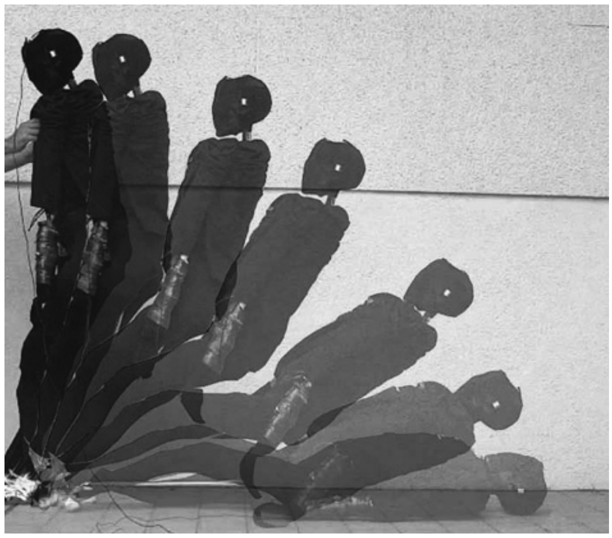

Fig. 2 Fall on Tile Floor

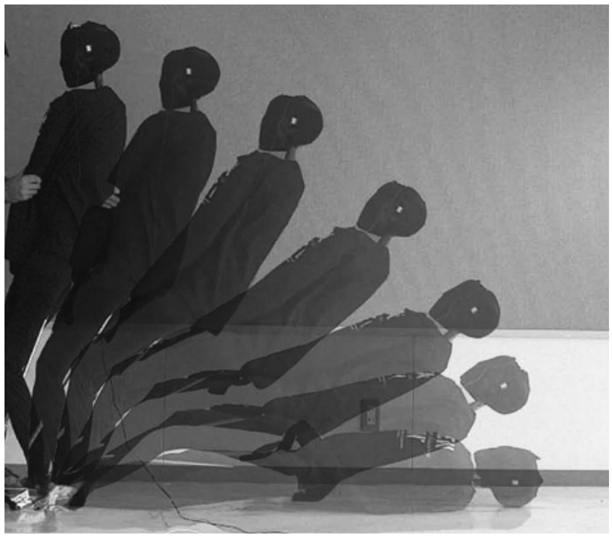

Fig. 3 Fall on Linoleum Floor

Table 1 Body segment weight data (Dummy)

Body part	weight rate (Female)	dummy weight (kg)
Head	0.082	3.9
Thorax	0.17	8.2
Abdomen	0.122	5.8
Pelvis	0.159	7.6
Upper Arm	0.058	2.8
Foream and hand	0.041	2
Thigh	0.235	11.3
Leg and foot	0.133	6.4
Total	1	48

Experimental conditions

Preliminary experiments were performed at the JARI between September 13, 2011 and September 18, 2011. This experiment was performed at the Junshin-Gakuen University between August 12, 2013 and August 16, 2013.

The experimental conditions consisted of falling backward from a standing position without any protection such as protective headgear, as shown in Fig. 2 and 3. The postural change and resultant acceleration at the intracranial of gravity were measured at the time of the fall. The falling model used here assumed that no defensive posture was adopted prior to the fall due to a decreased level of awareness. The contact surfaces of the fall included 2 flooring materials, a tile floor, and a linoleum floor, both of which are currently commonly used as flooring material in hospitals and institutions. HIC measurements are generally calculated from acceleration response measurements made using a three-axis accelerometer affixed to the interior surface of a dummy head. These values may not accurately reflect acceleration changes within the brain, however, because the accelerometer is positioned on the skull's interior surface. In this experiment, therefore, we filled the dummy head with tofu of approximately the same characteristics and material strength as brain parenchyma and positioned the three-axis accelerometer at the center of volume. This allowed accurate measurement of changes in collision acceleration (G) and of HIC values within the brain. The head angular acceleration was calculated from the head displacement measured using tracking head markers and photographic images from a high-speed camera.

Methods

Direct injury evaluation (change in acceleration at head collision, HIC)

As an indicator of direct head injury, this report uses the change in acceleration at head collision acceleration and the HIC.

Acceleration at head impact has long been used internationally as an index for head injury. The Japan Head Tolerance Curve[10] defines impacts of 300 G or greater in less than 5 ms as likely to cause permanent damage. There have also been many other reports of brain damage resulting from intracranial acceleration change. These reports calculate the collision acceleration (G) according to the acceleration experienced by the brain parenchyma; it is likely that the cranium and brain parenchyma greatly reduce the impact due to falls. In an investigation of attenuation of impact force by body tissue, Matsuura et al.[11] reported that a 1-cm-thick layer of adipose tissue or muscle reduces impact force by a factor of 10. We therefore suppose an attenuation factor due to bone and brain tissue of approximately 15. From this, we define falls exceeding 20 G as extremely dangerous. Therefore, here we define a dangerous fall as a fall with a measured acceleration of 20 G or greater in less than 5 ms.

Impacts exceeding HIC 1,000 generally pose a risk of serious cranial damage. This report calculates the HIC according to the acceleration experienced by the brain parenchyma; it is likely that the cranium and brain parenchyma greatly reduce the impact due to falls. We therefore suppose an attenuation factor due to bone and brain tissue of approximately 15. From this, we define falls exceeding HIC 65 as extremely dangerous.

HIC is an index of head injury used in compliance with the American Society for Testing and Materials (ASTM) F1292-04 and European EN1177 standard specifications, and is calculated using the following formula (1):

$$HIC = \left[\left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)\, dt \right\} (t_2 - t_1) \right]_{Max}^{2.5} \quad \cdots\cdots (1)$$

Where t_1, t_2 = start and end time (in sec) of interval when HIC is at maximum;

$a(t)$ = resultant acceleration at the head's center of gravity;

max = maximum acceleration value in t_1–t_2 interval.

Concussion injury evaluation (change in angular acceleration at head collision)

In this study, head angular acceleration was used to evaluate concussion injury. The brain's ability to withstand injury due to a strong rotational movement of the neck can be evaluated by assessing the head's angular acceleration $\ddot{\theta}_{Max}$[12] or changes in the angular velocity. According to many studies, Maximum angular acceleration $\ddot{\theta}_{Max}$ was the index that can most evaluate of DAI due to shear deformation of the brain[13][15]. Hence, in the present study, maximum angular acceleration ($\ddot{\theta}_{Max}$) was used to evaluate the risk of DAI.

Values were calculated at each instant during the fall, such that postural change θ each 1.0 msec was converted to an image, and the head displacement angle was calculated using a head marker and the heel as origin. Taylor expansion was used to increase the precision of values obtained through the fourth-order finite-difference method. $\ddot{\theta}_{Max}$ was calculated using equations (2) and (3).

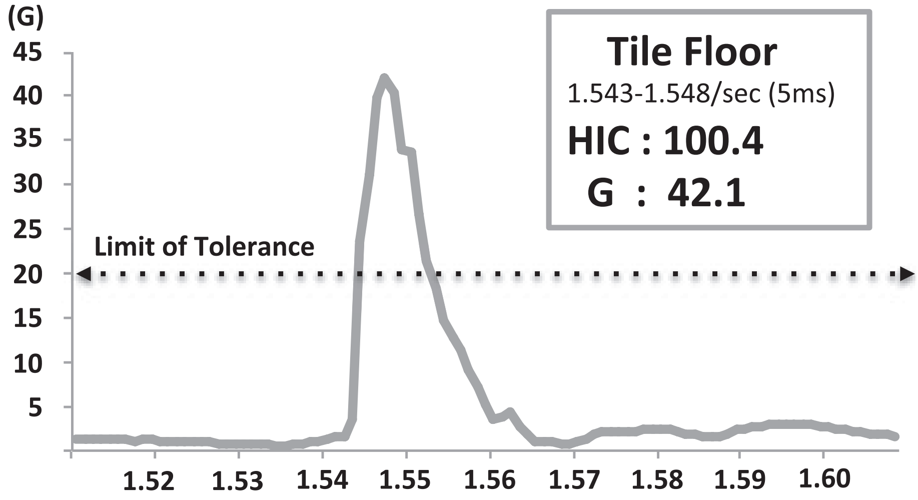

Fig. 4 Fall Acceleration on Tile Floor

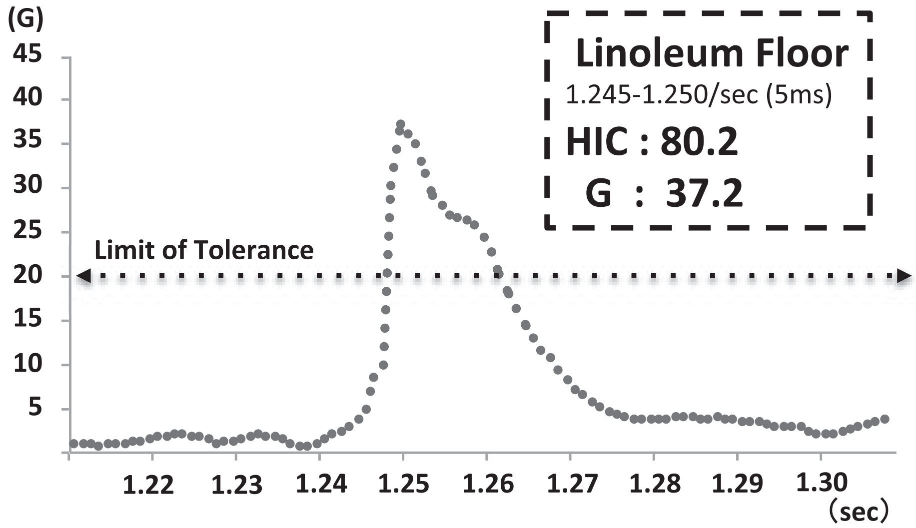

Fig. 5 Fall Acceleration on Linoleum Floor

$$\dot{\theta} = \left[\frac{\theta_2 - \theta_1}{(t_2 - t_1)} \right] \quad \cdots (2)$$

where $\dot{\theta}$ (rad/sec) is the angular velocity during the time interval from t_1 to t_2.

$$\ddot{\theta}_{Max} = \left[\frac{\dot{\theta}_2 - \dot{\theta}_1}{(t_2 - t_1)} \right] \quad \cdots (3)$$

$\ddot{\theta}_{Max}$ is the time derivative of the angular velocity obtained in equation (2).

A head angular acceleration ($\ddot{\theta}_{Max}$) value of ≥4,500–6,500 rad/sec^2 is usually considered a risk value for DAI[12)13)]. In addition, an acceleration value of ≥8,000–10,000 rad/sec^2 has been reported as the threshold risk value[14)15)]. In this study, we defined a fatal fall as a head angular acceleration of ≥4,500 rad/sec^2.

Results

Direct injury evaluation (change in acceleration at head collision, HIC)

Fig. 4 and 5 show the test results. In those figures, the vertical axis indicates acceleration (m/s^2), and the horizontal axis shows time(s). By substituting into Eq. (1) the start and end times of the highest peaks from the test results, the collision acceleration and HIC values are calculated as 42.1 G within 5 ms and HIC 100.4 for the

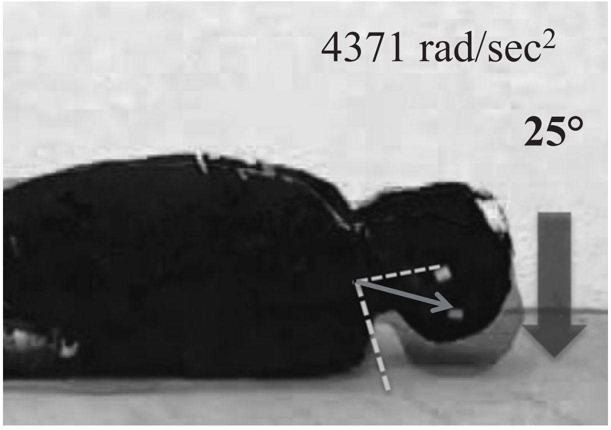

Fig. 6 Maximum Angular Acceleration After Impact on Tile Floor

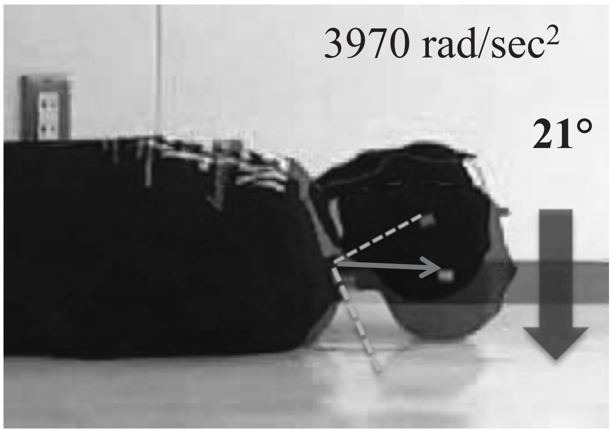

Fig. 7 Maximum Angular Acceleration After Impact on Linoleum Floor

tile floor and 37.2 G within 5 ms and HIC 80.2 for the linoleum floor. Thus, both cases represent extremely dangerous falls.

Concussion injury evaluation (change in angular acceleration at head collision)

Fig. 6 shows the head's angular displacement at an angular acceleration of 4,371 rad/sec^2 on impact with a tile floor. Fig. 6 shows the head's angular displacement when the head was displaced by 25° from the impact angle, which was the maximum displacement angle reached after impact with a tile floor at an angular acceleration of 4,371 rad/sec^2. Fig. 7 shows its angular displacement at an angular acceleration of 3,970 rad/sec^2 on impact with a linoleum floor. Fig. 7 shows the angular acceleration displacement when the head was displaced up to 21° from the impact angle, which was the highest displacement angle reached, after impact with a linoleum floor at an angular acceleration of 3,970 rad/sec^2.

Discussion

Assessment of direct injury due to falling backwards

With the goal of evaluating head injury in falling accidents, we installed an accelerometer within a simulated brain and evaluated direct injury from collision acceleration change and intracranial HIC. On the basis of previous research, we assumed an attenuation factor of 15 for the impact force on living tissue and therefore performed an evaluation in which a measured collision acceleration (G) and HIC values exceeding 20 and 65 corresponded to a standard collision acceleration (G) and HIC values exceeding 300 and 1,000. The resulting collision acceleration and HIC values were 42.1 G within 5 ms and HIC 100.4 on a tile floor. The resulting collision acceleration and HIC values were 37.2 G within 5 ms and HIC 80.2 on a linoleum floor. These values indicate that in both cases a falling accident would almost certainly result in serious cranial injury.

However, the calculated HIC and collision acceleration (G) values were slightly lower for the linoleum floor. This could be explained by the softer quality of linoleum (vinyl chloride) compared to tile, which allows for a longer contact time due to deformation of the material. This observation suggests there are possible to reduce the collision acceleration (G) and HIC values by choosing a softer material for flooring. Linoleum flooring, which is quite inexpensive and easy to install, is used in many buildings. It is particularly used in medical facilities, since it is impermeable to blood and other body fluids, and is very hygienic. However, our results showed that falling in a facility with linoleum flooring still carries a high risk of suffering a serious direct head injury. Therefore, it is necessary to take additional preventive measures against direct injury due to falling.

The risk of direct injury caused by falling is significantly affected by the acceleration upon impact with the floor surface. Therefore, increasing the contact time at the moment of impact is most effective for reducing the HIC and collision acceleration values. Other protective measures such as using a headgear are also effective in preventing injury to the head.

In this study, we assumed the worst conditions for falling is when the individual falls backward without adopting a defensive posture due to a decreased level of awareness. However, it is believed that in a normal falling situation, people generally protect themselves with their arms or fall on their hips, and few cases involve a direct fall on the head.

However, people do not generally fall on a flat surface only. In a three-dimensional environment, many reported cases involve people sustaining a strong blow to the head against bed railings, a bedside lampstand, or bedside table without assuming any defensive posture as they fall.

Therefore, we consider that it is very important to acknowledge the risk associated with direct head injury due to falling.

Assessment of concussion injury due to falling backwards

We evaluated the risk of concussion head injury by calculating the change in angular acceleration,($\ddot{\theta}_{Max}$), of the head. As a result, the change in angular acceleration was estimated at 4,371 rad/sec^2 for a fall on a tile floor, and at 3,970 rad/sec^2 for a fall on a linoleum floor. This value is very near the DAI tolerance limit reported by Lowenhielm and Ommaya[12)14)] and suggests that shear stress-induced deformation in the brain parenchyma results from the fall, which indicates a high risk of sustaining a serious injury. This value is lower than that considered to be the threshold DAI risk value in this study. However, in elderly people, the elasticity and compliance of blood vessels are lost due to ageing, and blood vessels become stiffer. Consequently, blood vessels at the surface of the brain parenchyma could easily rupture when compressed or extended due to the impact from the fall. This would lead to a high risk of cerebral hemorrhage, including DAI. Therefore, a fall on either type of flooring surface could be just as fatal.

In this experimental result, we observed that concussion injury factor. Concussion injuries have one type. This type showed a specific trend of instantaneous change in angular acceleration on contact with the floor surface, as in the case of a concussion injury.

Interestingly, our results also showed that the risk of concussion injury is not significantly different between falling on a tile floor and falling on a linoleum floor.

The origin of concussion injury is the sudden movement of the head forward and backwards following a rear impact mainly to the neck. Therefore, although the use of headgear extends the contact time at the moment of impact and is probably effective in preventing direct injury, it seems to be insufficient for preventing concussion injury. Thus, we believe that additional protective headgear that is similar to a neck collar should be developed to prevent concussion injury, fix the neck's range of motion, and decrease the head's forward and backward motion.

A limitation of the present study is that the initial velocity at the time of the fall was not considered. Many of the falling accidents that actually occur in hospitals are due to slipping or falling while moving. If the initial velocity at the time of the fall is taken into account, a higher angular acceleration would be measured than those obtained under the current experimental conditions. Moreover, the possibility of assuming defensive posture at the time of the fall was not considered by the present study.

However, this study demonstrated that, during a fall, there is risk of direct injury as well as of concussion injury in the skull, prior to and following impact. This is very likely to effectively help in devising preventive measures against injuries caused by falling in the future.

Conclusion

In this study, we have evaluated head injuries sustained due to falling by measuring the head angular acceleration and collision acceleration (G) and head injury criteria (HIC). These measurements were obtained using an impact dummy model, and were based on the change in posture during a fall and on the resultant acceleration at the head's center of gravity.

We observed that the risk of sustaining a direct injury due to a fall is very high in most medical institutions, i.e., whether tile or linoleum was used as flooring, as demonstrated by the collision acceleration (G) values of 42.1 G and 37.2 G that correspond to both floorings, respectively.

Maximum change in angular acceleration was estimated at 4,371 rad/sec^2 on a tile floor, and 3,970 rad/sec^2 on a linoleum floor.

The above findings indicate that protective gears should have leveled surfaces and that the use of protective gears such as headgears is effective in providing protection against direct injuries; however, such equipment does not provide sufficient protection against concussion injuries.

Till date, protective gears such as headgear and mats are only used in some facilities such as psychiatric wards or nursing homes. The general use of such equipment is probably avoided because of the discomfort experienced by patients while wearing them and the resultant hindrance in nursing.

During the course of this study, the use of protective gears successfully prevented direct head injuries, suggesting that it is important to implement preventive measures involving the use of protective gears and promote awareness programs about these equipment.

In addition, conventional protective gears may not be effective in preventing concussion injuries; this may be attributable to the fact that concussion injuries do not occur from direct contact with the floor; rather, they occur because of torsion in an anteroposterior direction.

The above findings indicate the necessity to manufacture protective equipment in a way that anteroposterior movements at the time of collision are inhibited, thus preventing concussion injuries. Moreover, practicing defensive postures may be useful for this purpose, e.g., correct falling techniques in judo, wherein movements that cause neck torsion are controlled by pulling the chin in.

Compared with that of femoral fractures, the incidence of head injuries caused by falls is low and the extent of attention given to these injuries is less; therefore, preventive measures based on the mechanism of onset are yet to be elucidated.

It is extremely important to promote preventive measures against head injuries, which are based on the mechanism of onset, and for this purpose, nurses who commonly come in direct contact with patients play a major role.

In future, it will be necessary to conduct studies for verifying the efficacy of conventional protective gears, effective preventive measures based on the mechanism of onset, and prevention practices.

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臨床現場での転倒による頭部外傷発生機序と外傷予防に向けた看護ケアの検討

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―キーワード―
転倒事故，直達外傷，震盪外傷

本研究の目的は，臨床現場において直接的に看護師が関わることが多く，事故発生による外傷発生率の高い転倒事故の予防に向けて，衝突モデルダミーを用いた転倒実験を行い，転倒時の姿勢変化と衝突加速度から転倒によって生じる頭部の外傷危険度を予測検証することである．その結果，後方へ防御姿勢を取ることなく転倒するケースにおいて，床面がタイル床と浴室での転倒を想定した場合HIC＝100.4 衝突加速度＝42.1Gであり，病室を想定したリノリュウム床での転倒ではHIC＝80.2 衝突加速度＝37.2Gと転倒による頭部骨折など頭部における直達外傷の危険性が極めて高いことや，転倒姿勢変化に伴う，頭部角加速度変化から，角加速度は最大で4,371rad/sec^{2}とびまん性脳軸索損傷（DAI）などの震盪外傷の危険性も高いことを明らかとした．

これらのことから，医療施設において転倒は極めて危険性の高い事故であり，夜間覚醒時や認知症患者など転倒時における防御姿勢を取ることが困難な患者において転倒による頭部外傷を軽減する為には，頭部への直接的な外傷を軽減するだけでなく，頸部も含めた頭部の回転を軽減することで頭蓋内部の損傷を軽減できる予防具の使用が必要であることを示した．

（日職災医誌，62：65―73，2014）