Design of Wearable Airbag with Injury Reducing System

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Abstract: Injuries caused by falls has become significant social problem in aging society. Falls could cause fractures which is significant cause of morbidity and mortality. As a result, active protecting devices are being developed to protect body from severe injuries. In this study, simulation test method of falls situation is established and the wearable airbag system for protecting from falls is designed through simulation. Ergonomic design is considered in this wearable airbag system to reduce injury level effectively. It will be possible to establish the reliability of the development of the fall prevention system for the elderly and to be the basis for the future development.

1 INTRODUCTION

Falls are an accident that occurs when a person's balance is lost and the balance cannot be maintained again. This falls occur in all age groups, but especially in elder people. The fall of an elderly person causes a fear of falling and reduces the amount of activity, and acts as a major threat to the independent living and quality of life of the elderly (Vellas et al., 1997). In the injury statistics due to fall, about 68% of injuries caused by falls were fractures, and about 47.1% of injuries occurred in pelvic and thighs (Watson and Mitchell, 2011). 90% of hip fractures in elderly peoples are caused by falls, and 20% of elderly patients hospitalized for hip fracture due to falls die during treatment (Courtney et al., 1995). These problems cause serious problems in the world, which is turning into an aging society. To prevent injury by falls in the elderly, it is necessary to have a device that can relieve the fall impact. Developed equipment is divided into passive and active type to protect the body from falling of elderly person. In the case of passive type, there is a fixed type of protective material such as pad or form, and it shows about 2.5% ~ 50% protection performance in case of impact energy equivalent to fall (Laing et al., 2011; Nabhani and Bamford, 2002; Holzer et al., 2009). However, this makes it less comfortable for the user and makes the elderly person avoid wearing it (Honkanen et al., 2006). Research on active equipment is divided into research on fall detection algorithms and research on wearable air bags. In the case of wearable airbags, there are studies to confirm the performance of the airbag using a dummy (Fukaya and Uchida, 2008). However, most studies have designed the airbag only as a shock absorber between the ground and the human body without going through ergonomic design. In addition, verification is not done through simulation, so the expected performance before device development can not be grasped.

Therefore, in this sutdy, the ergonomic wearable airbag is designed and the effectiveness of the wearable airbag is confirmed thorugh simulation of reducing the fall injury.

2 MATERIALS AND METHODS

2.1 Design of the Ergonomic Wearable Airbag System

For falls simulations, human models, environments (boundary conditions) and algorithms must be applied to simulations. First, in case of human models, pedestrian facet model (mathematical dynamic model) provided by the MADYMO program (Release 7.6, TASS international, Netherlands). Wearable protective equipment has high protection performance when it is made of hard material, but users prefer the soft material because they do not want to wear it (Honkanen et al., 2006). Also, since there

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is a difference in the protection performance according to the wearing method of the wearer or the fall method, it is possible to fix the body to the maximum to overcome it (Forsen et al., 2004). The shape of the wearable airbag is designed to cover all parts except the face part by considering the average body size of the elderly over 65 years old, and parts of the hips are designed as a skirt type to wrap around the side and back direction.

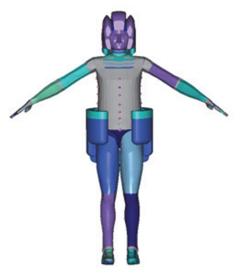

Figure 1: Human model wearing airbag(front(a) back(b) side(c)).

The human model in Figure 1 wear the ergonomically designed airbag. The wearable airbag system use acceleration sensor, gyro sensor and compass sensor to input each sensor data to the CPU. Then, CPU uses the input sensor data to convert them to values such as angles, angular velocity and acceleration. The converted values are substituted into the threshold value determination algorithm to determine whether the wearer's state is falling or not. When it is judged that the falls has not occurred, the continuous voltage is transmitted to the sensor unit so that the current state of the wearer can get feedback. In case of a fall, the CPU sends a signal to the inflator so that the airbag can be deployed.

The airbag is deployed in the order of the hips, thorax and head, which are the order of contact between the human body and the ground at the time of falling. When the airbag touches the ground, the vent hole is actuated at the moment and discharge the gas inside the airbag into the outside.

Figure 2 shows the wearable airbag system as a block diagram. The sensor block consists of gyro, acceleration and geomagnetic sensor to measure the current state of the wearer. The measured data is transmitted to the CPU to calculate the angle, angular velocity, and angular acceleration. The thresholding algorithm input to the CPU has a threshold value for each computed element. Further, when the condition is satisfied, it is judged that a fall occurs. If the CPU judges that a fall has occurred, the CPU sends a signal to the driver block. When the CPU judges that the fall has not occurred, the CPU sends a signal to the sensor block. If a signal is received from the driver block, the inflator is activated because a fall has occurred. The airbag is deployed by the action of the inflator and protects the wearer from collision with the ground. After collision with the ground, the Venthole works to release the air inside the airbag, reducing the impact of collision between the wearer and the ground. If the sensor block receives a signal, it will not fall and will continue to measure the current state of the wearer.

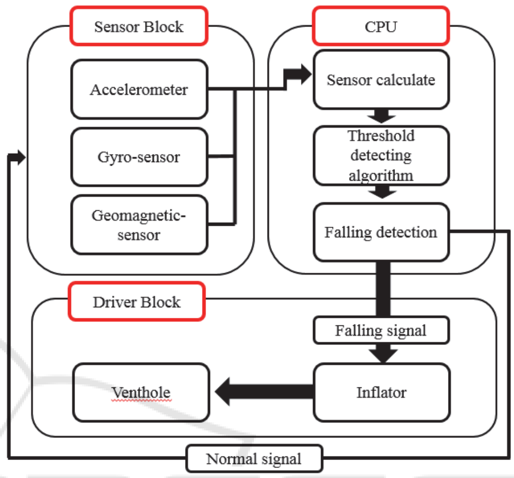

Figure 2: Block diagram of wearable airbag system.

2.2 Simulation of falls situation and reducing the fall injury

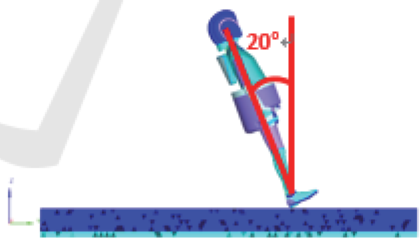

Figure 3: Initial state of human model falling simulation.

The falling simulation of the human model is made on a ground condition of concrete properties (Density: 2300 kg/m^3 , Elastic modulus: 17E+9N/m^2, Poisson's ratio : 0.15) and the angle of 70° between the lower limb and the ground is set, where the elderly person cannot maintain and regain the balance (Hsiao, 2008) (Figure 3). Thereafter, gravity is applied to allow the human body to fall back on the ground.

To determine how much the human body model was injured by the falls, HIC (Head Injury Criterion), CTI (Combined Thoracic index), Impact force (between the hips and the ground) is measured in simulation. At the beginning of the simulation, the human body model is tilted toward the ground by gravity. As the angle of the joint cannot be maintained due to the characteristics of the human body model, the joint collapses due to the gravity, and the human body model impacts to the ground in the order of the hips, thorax and head.

This simulation compared case 1 (before wearing airbag) and case 2 (after wearing airbag), and determine the effectiveness of the designed wearable airbag.

3 RESULTS AND DISCUSSION

(a) (b) (c)

Figure 4: The collision between the human body model and the ground with airbags on the simulation.(Case 2-A(a), Case 2-B(b), Case 2-C(c)).

Simulation results for four cases were measured. Figure 4 shows the moments when the human model with the airbag collided with the ground in case 2 simulations. (a) only wears a pelvic airbag and the head and head collide after a pelvic impact. Case 1, in which no airbag is worn, collides with the ground in the same order as in (a). The collision simulation of case 2-B, like (a), causes the head to collide with the ground after collision between pelvic and ground. Simulation (c) is the last time the head collides with the ground on the volume of the chest airbag after a primary collision between the pelvic and the ground.

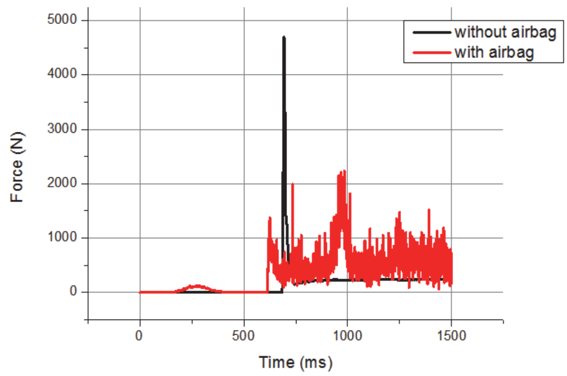

Figure 5: Impact force of before and after wearing pelvic airbag.

The graph of Figure 5 shows the impact force applied to the hips of the human model at the time of the falls in case 1 (before wearing airbag) and case 2 (after wearing airbag).

In case 1, maximum impact force is 4694N, and in case 2, maximum impact force is 2240N, respectively. Case 1 (4694N) exceed the reference hip fracture point of the elderly people (3100N) and case 2 (2240N) not exceed (Kennedy, 1987).

Table 1: Head and spine injuries of before and after wearing airbag.

	HIC	CTI
No Airbag(case 1)	1657	0.2163
Pelvic(case2-A)	4412.2	0.2015
Pelvic+Head(case2-B)	498.06	0.1517
Pelvic+Spine+Head(case2-C)	165.37	0.191

The injuries of the head and thoracic are shown in Table 1. In case 1, the HIC value is 1657, which is above the reference value of the concussion (1000). This can cause more injuries depending on the situation. And the CTI value not exceed the standard value of fatal thoracic injury (1.0) but numerically indicates that the injury occurred to the human body.

Case 2 is subdivided to wearing only pelvic airbag (Case 2-A), wearing pelvic and head airbag (Case 2-B), and wearing pelvic, thoracic and head airbag (Case 2-C). And HIC and CTI were measured for each case. HIC had a value of 4412.2 in case 2-A, which was 2755 higher than that of case 1. This is classified as a very serious injury. The reason for this result is that the sequence of the impact of the human body model with the ground changes in order of hip, head and thoracic, in the process of protecting the human body model. As a result, the head is more impacted than the situation without airbags, and the injury level has increased. However, when the head airbag was worn (Case 2-B), the injury rate was reduced to 498.06, which was about 89% reduction in injury rate when wearing only the hip airbag (Case 2-A). In case 2-C (wearing the entire airbag), it was found that 90% of the injury rate were reduced than case 1. If do not wear a thoracic airbag, head will collide with the ground earlier than thoracic. However, in case of wearing a thoracic airbag, the volume of the airbag causes the chest to collide with the ground earlier than head, and thoracic injury level increase but head injury level decrease. To conclude, when entire airbags were worn, the human body model was found to be able to prevent fatal injuries by reducing injury levels than without airbags.

Based on the results of the simulation, the wearable pelvic airbag system was developed (Figure 5).

The main purpose of previous studies was to determine falls. However, it did not address the injury reduction by airbags. this study showed numerical reduction of injuries caused by fall, which showed that it is applicable to real human fall situation.

(a) (b)

Figure 6: Pelvic airbag(before inflate(a) after inflate(b)).

This hip airbag system is manufactured in the form of a belt as shown Figure 5-a, and can accommodate the airbag. And sensor module (3D accelerometer, gyro-sensor and compass sensor) is used to measure and calculate z-axis acceleration, sum acceleration, angular velocity, tilt angle, obliquity angle, resultant angle. These values is used in double threshold algorithm to determine fall event. If fall event is found, the inside airbag (thermoplastic poly urethane) is unfolded by the gas.

4 CONCLUSIONS

In this study, the ergonomic wearable airbag system is designed and the effectiveness of the airbag is proved by showing the injury value in the simulation. And also, a simulation method that can be used as a basis of safety research for the elderly. Based on simulation results, the actual wearable airbag system was developed. This airbag system is expected to prevent fractures and reduce cost of treatment. In addition, through this study, it is possible to develop wearable airbags in other parts to prevent injuries caused by falls.

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