Protection against Impact with the Ground Using Wearable Airbags

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Abstract: Incidental falls from heights, falls on the same level caused by slipping or tripping, and falls from wheelchair overturns are commonplace phenomena, associated with serious injuries from impact with the ground. A wearable airbag device is a countermeasure applicable to all these types of incidents. Three types of wearable airbag systems were developed and evaluated: for protection against falls from heights (Type-1), against wheelchair overturns (Type-2), and against falls on the same level (Type-3). The systems consist of an airbag, sensor, inflator, and jacket. The sensor detects the fall and the airbag inflates to protect the user. Fall tests using dummies with/without the airbags demonstrated the effectiveness of these devices. In the experiments with system Type-1, for fall heights of less than 2m, the airbags reduced the impact acceleration, and the Head Injury Criterion (HIC) values were under 1,000, the auto-crash test requirement. However, there are limits to the amount of protection afforded: in Type-1, the airbag can protect only the back of the head.; in Types-2 and 3, the fall height of the center of gravity is lower than 2m, and there is some margin of extra protective resource, which can be used to extend the protected area.

Key words: Wearable airbag, Protective devices, Fall, Slip, Trip, Wheelchair overturn

Introduction

In the construction industry and in many other industries, incidental falls from heights are still frequent despite fall protection measures such as protective railings, and safety belt systems. Incidental falls on the same level are also commonplace, caused by slipping or tripping in industrial and daily life settings. The most dangerous outcome from a fall incident is injury to the head and the back from impact with the ground. Therefore, a wearable airbag system which can cushion the impact may be an appropriate countermeasure applicable for different fall situations. The authors have participated in the development of protective devices using wearable airbags and have evaluated the shock absorption performance of airbags. These airbag systems are designed for protection against falls from heights (Type 1)\textsuperscript{1}, for protection against falls from wheelchair overturns (Type 2)\textsuperscript{2}, and for protection against falls on the same level (Type 3)\textsuperscript{3}.

The Type 3 airbag system is mainly targeted toward the elderly. There are two major tasks in the development of the airbag system: one is the design of the airbag itself, and the other is the design of the sensor system. This paper introduces three wearable airbag systems for fall protection applications, describes experiments for evaluation of their shock absorption performance, and provides discussion on the effectiveness and limitations of their potential applications.

The Components of a Wearable Airbag System

The wearable airbag system consists of airbags (see Figs. 1–3), a sensor system, an inflator (see Fig. 4), and a jacket. The user of the airbag system wears the jacket, which contains the folded airbag, sensor system, and inflator. When the sensor system detects a fall incident, the inflator expands the airbags, which protect the user against impact with the ground.

The sensor system consists of a processor unit and a
sensor system, which is a combination of an accelerometer, rotation sensor, and/or distance sensor. In airbag system Type-1 (for falls from heights), a 3-axis accelerometer detects zero gravity, which indicates a free fall. In airbag system Type-2 (for wheelchair overturns), distance sensors on both sides of the wheelchair detect the distance between the bottom of the wheelchair and the ground, and the processor unit determines the inclination. In airbag system Type-3 (for falls on the same level), a 3-axis accelerometer and a rotation sensor detect a fall initiation. Both sensors are necessary to distinguish sudden falls from daily non-risky movements such as sitting, bending down, etc. The sensor system could detect simulated falls when attached to a dummy. The sensor system could also distinguish between intentional falls and other types of daily non-risky movements when attached to a human. However, as there are numerous kinds of ordinary non-risky movements, testing of the sensor system under real-life conditions is being continued.

Fig. 1. Airbag for protection against falls from heights (Type-1 airbag).

Fig. 2. Airbag for protection against wheelchair overturns (Type-2 airbag).

Fig. 3. Airbags for protection against same level falls (Type-3 airbag).

Fig. 4. Airbag inflator.
The inflator is a steel cylinder with pressurized gas; it inflates the airbag with high-pressure gas in case an incident is detected. The Type-1 and Type-2 airbags systems use one large inflator (30 liter gas capacity). The Type-3 airbag system uses two small inflators (10 liter gas capacity) — one for each airbag.

Test Method

Measurements

There were many studies on impact injuries and injury tolerances relating to the automobile accidents. Many impact experiments using cadavers, animals, volunteer and dummies were conducted and many injury tolerance criteria, such as Severity Index (SI), Head Injury Criterion (HIC), were proposed. And in USA automobile impact tests are required in law to evaluate passenger protection, such as air bags. In the tests impact accelerations of the dummy are measured, and the impact accelerations must be reduced to fulfill the HIC.

We also conducted impact tests and acceleration measurements. Fall incident simulation tests, i.e., of falls from heights, falls from wheelchairs overturns, and falls on the same level were conducted using dummies equipped with 3-axial accelerometers. The difference between impact accelerations in conditions with and without airbags were measured. Small-size 3-axial accelerometers (NEC SANEI 9G3102S; 7 × 8 × 5.5 mm) were used. One accelerometer was attached to the dummy’s face or placed inside the dummy’s head (see Fig. 5) — preliminary tests showed that there was no difference between the waveforms of accelerometers on the face and in the head. A second accelerometer was attached to the body. The acceleration data were recorded through an A/D converter at a sampling frequency 20-50 kHz, and were later analyzed with a computer. The tests were also recorded with a normal and a high-speed video, except for the experiments with the Type-3 airbag system. The records were used to understand the impact process — for example, a contact of the head with the ground can be ascertained using the videos. For the tests, the airbags were pre-inflated using a compressor.

Drop test

Drop tests using dummies were conducted to measure the shock absorption performance of Type-1 airbag (see Fig. 6). Measurements were carried out under several different fall conditions: with airbag and without airbag, falls in the horizontal position and falls in the inclined position, and falls from various heights. Two dummies were used in the tests. One dummy (dummy A) was made for rescue training (weight 76 kg, height 163 cm). The weight of the dummy was set to 85 kg with a weight of 9 kg in the jacket and the head. The weight of 85 kg is the same as that of the dummies used for safety belt drop tests in Japan and represents the 95th percentile of the male Japanese. The other dummy (dummy B) was an 85 kg steel torso equipped with 3 axial accelerometers (Kyowa AS-THH & AS-THC; for automobile crash test); only the neck was bendable. This dummy was used for the drop tests with an inclined posture.

Wheelchair overturn test

In the tests on the Type-2 airbag, a wheelchair with a dummy was inclined and overturned by pushing the wheelchair from a rectangular platform about 10 cm high. The overturn of the wheelchair was conducted in three ways: one with a dummy wearing a jacket airbag, another with an airbag incorporated into the wheelchair, and a third without an airbag. Several airbag shapes (Long
guard, Face guard, Oval, C Large. see Fig.11) were tested. The overturn of the wheelchair was also conducted in different directions: lateral overturn and backward overturn. For some airbag shapes, forward over turns were also conducted.

The dummy used for the tests (dummy C), which was developed for car crash experiments based on the Japanese physical constitution, was 166.7 cm in height and 60.6 kg in weight. The aluminum structure of the dummy was covered with hard sponge. The dummy had multiple joints and could assume a variety of postures; moreover, changes in the constriction force acting on the joints altered the flexibility of the posture.

**Same-level fall test**

A test method and test apparatus\(^9\), which is developed by Nagata et al, was used.

Dummy C was again used in the tests on the Type-3 airbag. Large constriction forces on all the joints (fixed joint) made the dummy rigid, and a small constriction force (non-fixed joint) on a joint made the joint free. In case of fixed joints, the falling conditions were similar to those of falling rigid bodies (rigid-body-type fall); and in case of non-fixed joints, the falling conditions were set as falling onto the buttocks (buttock-type fall).

Since the dummy could not stand by itself it was hung from a frame and positioned on a moving platform. A synchronized release of the dummy with motion of the platform simulated a fall on the level ground (see Fig. 7).

Using this experimental setup, tests on backward falls with/without airbags were conducted. Experiments on forward falls were also conducted because the elderly sometimes cannot protect their heads when falling forward. In the forward fall experiments, two types of airbags were tested: airbags for backward fall protection and airbags for forward fall protection. These airbags had the same shape but were worn in reversed orientation.

Preliminary tests of backward falls with dummies demonstrated that the maximum head acceleration was greater in a rigid-body-type fall as compared to a buttocks-type fall; therefore, rigid-body-type falls were conducted in the study.

**Test Results**

**Drop test**

Figure 8 shows the relation between the maximum acceleration of the head with/without an airbag and the fall height for falls in the horizontal posture.

Drop tests without an airbag from 2-m height were carried out twice. One of the data sets exhibited a low accel-
eration value; here, the dummy was slightly inclined to the left or right, which made the shoulder touch the ground first. This reduced the maximum acceleration.

From the head acceleration data, the HIC values were calculated in order to understand the absolute level of the shock absorption performance. Figure 9 shows the relation between the HIC values and the fall height. The HIC is defined as follows:

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HIC = \left( \frac{\int_{t_1}^{t_2} a \, dt}{(t_2 - t_1)} \right)^{2.5} \cdot (t_2 - t_1)
\]

where \(a\) is the acceleration, and \(t_1\) and \(t_2\) represent time. This criterion is used in impact tests on automobiles, and for HIC values under 1,000, it is considered that the probability of severe injury is low. In the current drop tests without airbag, the HIC values were over 1,000, while in the tests with airbag the HIC values reached over 1,000 only at fall heights of more than 2 m.

Figure 10 shows the relation between the maximum acceleration of the head and the large inclination in drop tests for 1 m fall heights with an airbag. A positive value indicates a head-down inclination, and a negative value a head-up inclination.

Wheelchair overturn test

The results for maximum acceleration in wheelchair overturn tests of various airbags are summarized for three different overturn directions in Fig. 11.

When the wheelchair is overturned, the dummy is thrown out of the wheelchair. For adequate protection in such incidents, it is important to identify the most appropriate shape and position of the airbag: a jacket-type or a wheelchair-incorporated-type airbag. When an airbag is attached to the wheelchair, in case of an overturn, the person may miss the airbag and hit the ground. However, this problem is not as serious with a jacket-type airbag that moves with the person.

Same level fall test

Figure 12 shows the maximum accelerations both with and without an airbag. There is a small acceleration in case of no airbag. Videos show that the left side of the dummy hits the ground before the right side. This generates a buffering effect. As mentioned previously, when a fall-down process is divided into two or more steps such an impact on the left shoulder and the right shoulder, or an impact on the buttocks and head, the maximum acceleration is smaller as compared to the case wherein the entire dummy body hits the ground simultaneously.

Discussion

These experiments confirmed the effectiveness of different airbag systems to decrease the maximum acceleration in a simulated fall tests. Thus, wearable airbags could reduce the degree of injury in many types of fall incidents. However, the tested wearable airbags were efficient only within a range of conditions, and could not provide complete protection from injury to the user.

There are two approaches to devising incident countermeasures: one is prevention of incidents, and the other is prevention or reduction of the injury that can be caused by incidents. The prevention strategy is always preferable because no damage is incurred. However, there are cases in which prevention is not possible, such as during a natural disaster. It is usually extremely difficult to prevent incidents completely. Therefore, injury reduction strategies are also needed: falls from heights and falls on same level are some of the applicable areas. In this sense,
Fig. 11. Maximum impact acceleration in wheelchair overturn tests of airbag system Type-2.

Fig. 12. Maximum impact acceleration in fall-on-the-same-level tests with different Type-3 airbag systems and joint motion restriction conditions.
the wearable airbag system, albeit not the best countermeasure, is very necessary.

Drop tests show that when the fall height is below 2 m, the HIC value of the impact acceleration is less than 1,000. This demonstrates the limit of the protection offered by the system. This limit arises from the fact that the wearable airbag system must be light and can have only a small amount of gas to inflate the airbag. Thus, as it cannot protect the entire body, prioritizing protection is necessary. The most important region that must be protected is the head; head injury is the leading cause of death in 60% of fatal fall accidents. In addition, protection for the back of the head is more important than that for the front because the front can be protected with the hands, but not the back. Table 1 shows the level of impact energy that the airbags must dissipate during the incidents. For airbag systems Types-2 and 3, the energy is smaller than that for Type-1; therefore, there is some room to extend protection to regions other than the back of the head in these systems.

In the wheelchair overturn tests, the maximum impact acceleration was relatively low in some cases, and higher in other cases. The lower values were obtained when the dummy fell on the airbag, while the higher values resulted when the head missed the airbag and directly hit the ground. The reasons why the head misses the airbag could be as follows: an overturn produces not only a downward velocity but also a horizontal velocity. This velocity causes the difference in the impact positions on the ground between the airbag and the head. In the tests, the motions of the wheelchair were small, but in real-life use of a wheelchair, the motion of the wheelchair in the time of the overturn may be significant; it may exacerbate the difference between the head motion and the airbag motion. This factor may necessitate greater protection resources.

The design of a fall protection device for the elderly, should provide protection not only against fatal injury but include also a more broad protection against bone fractures in general, since a bone fracture may decrease the quality of life of an elderly for a long period of time. However, the realization of this requirement necessitates greater protection resources.

### References


