Wrist injuries in snowboarding –
Simulation of a worst case scenario of snowboard falls

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Abstract

Snowboarding is one of the most popular winter sports, particularly among adolescents and younger adults. The risk of injuries while snowboarding is higher compared with alpine skiing, with the wrist as the dominant injury region. In contrast to increasing numbers regarding helmet usage, acceptance for wearing wrist protectors is decreasing. To date the market offers a variety of wrist protection products for snowboarding which feature different protective elements. However, there are no minimum performance standards for snowboarding wrist protectors worldwide. Currently a harmonized international standard is under preparation to provide guidelines for minimum safety performance for all wrist protectors used in snowboarding.

In the course of this aim, a multi body system (MBS) was developed to acquire further knowledge about the functional requirements of wrist protectors. To evaluate a worst case scenario different falling scenarios of snowboarders were simulated to calculate the resulting loads in the upper extremity. The simulations were carried out using the multi body dynamics software package SIMPACK 9.0 (SIMPACK AG, Wessling, Germany). The comprehensive model contains a human model, a model of a ski slope and a model of a snowboard. The parameterized models adapt to the body height, the body weight and the shoe size of the snowboarder. In this study a model of a 50 percentile adult (1.80 m, 78.4 kg) was used.

To evaluate a worst case scenario well-known falling situations of snowboarders were simulated. The backward fall on outstretched joints of the upper extremity can be evaluated as worst case scenario.

Keywords: Snowboards, Wrist Guards, Computer Simulation.
1. Introduction

There are an estimated 10-15 million snowboard riders worldwide and it is particularly popular among adolescents and younger adults (Dann (2011), Hasler et al. (2010), Kusche et al. (2010)). Overall, the risk of injuries while snowboarding is higher compared with alpine skiing (Hagel et al. (2004), Sasaki et al. (1999)). The most frequently reported injured body region of all snowboarding injuries are the upper extremities especially the wrists (Russel et al. (2007)). Several studies have shown that the risk of sustaining a wrist injury can be reduced by wearing wrist protection (Hagel et al. (2005), Kim et al. (2011), Sasaki and Lee (1999), Machold et al. (2002)).

Currently, there are a wide variety of wrist protection products for snowboarding on the market which offer a range of protective features. However, there are no minimum performance standards for snowboarding wrist protectors worldwide. The International Society for Skiing Safety (ISSS) convened a task force to develop a White Paper to evaluate the importance and necessity of a minimum performance for all wrist protectors used in snowboarding (Michel et al. (2013)). The broader goal of developing and implementing such a standard is to reduce the incidence and the severity of wrist injuries in snowboarding without increasing the risk of adverse events, such as upper arm or shoulder injury. It is hypothesized that implementation of a snowboarding wrist protector standard would result in fewer and less severe wrist injuries in the sport and could translate into more riding days for healthy snowboarders and significant health care costs savings.

In the course of this aim, a multi body system (MBS) was developed to acquire further knowledge about the functional requirements of wrist protectors. In addition to laboratory experiments carried out by Schmitt et al. (2012) forward and backward falls were analysed. To evaluate a worst case scenario different falling situations of snowboarders were simulated to calculate the resulting loads in the upper extremity.

2. Method

The simulations were carried out using the multi body dynamics software package SIMPACK 9.0 (SIMPACK AG, Wessling, Germany). The comprehensive model contains a human model, a model of a ski slope and a model of a snowboard (fig. 1). The parameterized models adapt to the body height, the body weight and the shoe size of the snowboarder. The stiffness in the joints were modeled with rotational spring-damper elements. In this study a model of a 50 percentile adult (1.80 m, 78.4 kg) based on the anthropometrical database of Size Germany was used. For the detailed model of the upper extremity the exact bone geometries, based on CT data 3D surface models, were created using 3D analysis software AMIRA (FEI Visualization Sciences Group, Bordeaux, France). The soft tissues in the upper extremity were connected to the bones as wobbling masses. The contacts in the wrist joint, i.e. in the Articulatio radiocarpalis and the Articulatio ulnocarpalis, were modeled as areal contacts. The Articulatio humeroulnaris in the elbow was modeled as a hinge joint.

![Fig. 1. Left: MBS model including human model, a model of a ski slope and a model of a snowboard. Right: Contact forces in the Articulatio radiocarpalis, the Articulatio ulnocarpalis and the Articulatio humeroulnaris](image-url)
To evaluate a worst case scenario two well-known falling situations of snowboarders were simulated (Idzikowski et al. (2000), Rønning et al. (2001)): a backward fall about the heelside of the snowboard and a forward fall which occur about the toeside of the snowboard (fig. 2).

Fig. 2. Specified angles for backward fall and forward fall of the human model.

Table 1 shows the specified joint angles at the start of the simulation for shoulder, elbow, wrist and finger joints for both scenarios according to the neutral zero method. A variation of the shoulder angles $\beta_S$ and elbow angles $\gamma_E$ were carried out to evaluate the loading situations.

<table>
<thead>
<tr>
<th>Joint</th>
<th>Backward fall</th>
<th>Forward fall</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder</td>
<td>Abduction $\alpha_S = 10^\circ$</td>
<td>Abduction $\alpha_S = 10^\circ$</td>
</tr>
<tr>
<td></td>
<td>Retroversion $\beta_S = 80^\circ$ (varied)</td>
<td>Anteversion $\beta_S = 100^\circ$ (varied)</td>
</tr>
<tr>
<td></td>
<td>External rotation $\gamma_S = 53^\circ$</td>
<td>Internal rotation $\gamma_S = 53^\circ$</td>
</tr>
<tr>
<td>Elbow</td>
<td>Supination $\beta_E = 19^\circ$</td>
<td>Pronation $\beta_E = 19^\circ$</td>
</tr>
<tr>
<td></td>
<td>Flexion $\gamma_E = 10^\circ$ (varied)</td>
<td>Flexion $\gamma_E = 10^\circ$ (varied)</td>
</tr>
<tr>
<td>Wrist</td>
<td>Abduction $\alpha_{HG} = 0^\circ$</td>
<td>Abduction $\alpha_{HG} = 0^\circ$</td>
</tr>
<tr>
<td></td>
<td>Extension $\gamma_{HG} = 45^\circ$</td>
<td>Extension $\gamma_{HG} = 45^\circ$</td>
</tr>
<tr>
<td>Finger</td>
<td>Flexion $\gamma_F = 30^\circ$</td>
<td>Flexion $\gamma_F = 30^\circ$</td>
</tr>
</tbody>
</table>

3. Results

First of all a validation of the model was carried out by using experimental studies described in the literature.
Figure 3 show the test set-up of the experimental tests carried out by Schmitt et al. (2012). The comparison of the joint angles during ground contact showed a good conformity of the simulation and experimental results (table 2).

![Figure 3. Simulation of the experimental tests carried out by Schmitt et al. (2012)](image)

**Table 2. Kinematic data during ground contact**

<table>
<thead>
<tr>
<th>Joint</th>
<th>Left side [°]</th>
<th>Right side [°]</th>
<th>Simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>(from Schmitt et al. (2012))</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Wrist extension</td>
<td>85 ± 7</td>
<td>82 ± 7</td>
<td>74</td>
</tr>
<tr>
<td>Ulnar abduction</td>
<td>6 ± 8</td>
<td>12 ± 11</td>
<td>5</td>
</tr>
</tbody>
</table>

Additionally a comparison of the ground reaction force with experimental data measured by DeGoede & Ashton-Miller (2003) was carried out by simulating a forward fall with an elbow flexion of 174° and a start position of the shoulder at the height of $z_s = 1m$. With the realistic anthropometric data (1.73 m, 73.0 kg) the characteristics of the curve with two force peaks showed the correct function of the model. The first force peaks $F_1$ show a difference of 185 N, the second force peaks $F_2$ are almost equal (fig. 4).

![Fig. 4. Ground reaction forces measured by DeGoede & Ashton-Miller (2003) (left) and calculated in the simulation (right)](image)

With the validated model the simulation of the backward and forward falls were carried out. The contact in the wrist joint was modeled as areal contact. Therefore the force peaks at the different contact position occur not at the same time. The humeroulnar joint was modeled as a hinge joint. For a better illustration of the results the joint forces in the elbow – which are in sum nearly equal to the forces in the wrist joint - are shown in the following figures.

Figure 5 show the dependence of the flexion in the elbow joint in a range from 0° to 35° on the resulting force in the humeroulnar joint for a 50 percentile adult (1.80 m, 78.4 kg). The peak forces were calculated with the outstretched elbow for both during the backward fall ($F_{max} = 3500 N$) and the forward fall ($F_{max} = 1950 N$). With increasing flexion angles a reduction of the joint forces were calculated.

Figure 6 show the dependence of the flexion in the shoulder joint in a range from 55° to 105° for the backward fall and in a range from -70° to -130° for the forward fall on the resulting force in the humeroulnar joint. Overall the maximum peak force of $F = 3200 N$ was calculated for the backward fall with a shoulder retroversion of 75°.
4. Discussion

With the developed MBS sample the backward fall was evaluated as worst case scenario during different falling situations of snowboarders. The peak force during hand impact was calculated for the simulation with an outstretched elbow joint and a retroversion of the shoulder of 80°.

These results are in accordance with the experimental results published by DeGoede & Ashton-Miller (2003) which also measured peak forces with outstretched elbow joint. Also the field study by Greenwald et al. (2011) and the laboratory study by Schmitt et al (2012) showed that backward falls resulted in statistically significantly higher maximum forces than forward falls. According to a literature review backward falls result in twice as many fractures as forward falls (Deady & Salonen (2010)).

5. Conclusion

The backward fall with outstretched joints of the upper extremity can be evaluated as worst case scenario. Therefore, this scenario should be considered for defining input variables as well as performance criteria for the development of a harmonized international standard for wrist protection in snowboarding. Moreover, these findings should be used for product development of wrist protectors.

Independent from standard development and possible product application, the findings regarding the varying elbow and shoulder angle appear to be important with respect to reducing the wrist loading and might thus be an interesting starting point in developing effective measures to prevent wrist injury, such as instructions for proper fall techniques.

Currently the worst case scenario of this falling situation is simulated with MBS models of different anthropometries (boy - 9 years, adolescent – 13 years, adult – 20 years) (fig. 7). To evaluate the fracture risk at the radius the resulting forces in the wrist will be implemented in finite element models. These data will be useful for designing functional graded wrist protectors based on different anthropometrics in terms of injury prevention and comfort aspects.
Fig. 7. Backward fall simulated with MBS models of different anthropometries (boy - 9 years, adolescent – 13 years, adult – 20 years)

Acknowledgements

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References