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The load tolerance of skin during impact on artificial turf using ex-vivo skin as the readout system

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ABSTRACT

Background: Understanding the mechanism of sliding induced skin injuries on artificial turf is important to define preventive measures. Recent findings revealed that high peak loads on the knee and thigh during impact of a sliding tackle are critical for inducing skin injuries. Unfortunately, skin failure data under combined impact load is lacking.

Methods: In this study the load tolerance of skin to impact on both dry and wet artificial turf and natural grass is investigated by developing an ex-vivo model, a biaxial load applicator and a loading protocol.

Results: The critical shear-normal stress combination at which skin breakdown occurred on dry artificial turf was 38 and 20 N cm\textsuperscript{-2}. Skin damage on wet artificial turf firstly was noticed at a shear-normal stress combination of 55 and 20 N cm\textsuperscript{-2}. On natural grass, skin breakdown only occurred at a combined shear and normal stress of 130 and 125 N cm\textsuperscript{-2}. The thickness of the stratum corneum after impact strongly correlated to the applied mechanical load during impact on dry artificial turf.

Conclusion: In contrast to natural grass, skin damage on dry artificial turf is strongly related to the magnitude of the impact load. Wetting of the artificial turf system improves the load tolerance of skin to impact.

Introduction

The benefits of artificial turf over natural grass such as lower maintenance costs per playing hours and playing consistency are generally recognized (FIFA 2013). However, artificial turf is still strongly associated with abrasion-type injuries (Fuller et al. 2007; Ekstrand et al. 2011). Survey studies revealed that players are complaining more about abrasion-type injuries than any other type of injuries (Zanetti 2009; Roberts et al. 2014; van den Eijnde et al. 2014a). Improving the sliding comfort and abandoning unpleasant sport surfaces, related to skin injuries, have therefore been of interest for manufacturers and sport-governing bodies. In order to reach these goals, a better understanding of the mechanisms behind skin injury is needed.

Current test methods for assessing the abrasiveness of artificial turf use skin replacers like silicon or foam as a readout (FIFA 2008b; ASTM 2009). However, these methods are questioned because they do not simulate the real load experienced by a player during a sliding (Sanchis et al. 2008). Further, the applied materials do not mimic the real skin response (van den Eijnde et al. 2014b).

A sliding movement consists of three phases: a free drop phase, an impact phase and a gliding phase (Corzatt et al. 1984). The existing test methods to evaluate the abrasiveness of soccer pitches focus on the gliding phase, where relatively low contact loads and large displacements are applied. Biomechanical fall studies focusing on the impact on the hip during side jumps by soccer goalkeepers showed that during impact, loads of 4.2–8.6 times body weight are common which corresponds with normal stresses ranging from 87 to 187 N cm\textsuperscript{-2} (Schmitt et al. 2010). Recent studies on the landing phase of a sliding movement reveal that the high peak stresses on the knee and thigh during impact are critical in inducing skin injuries. In their exploratory study, Van den Eijnde estimated that a combined shear and normal stress of at least 14 and 24 N cm\textsuperscript{-2} at impact velocities of approximately 3 m s\textsuperscript{-1} can induce a skin abrasion injury on dry artificial turf (van den Eijnde et al. 2016). At the moment, critical skin failure stress data in relation to fall studies on artificial turf or natural grass are missing. The aim of our study is to investigate the load tolerance of skin to impact on both dry and wet artificial turf and natural grass by using an ex-vivo model, a biaxial load applicator and a loading protocol. It is postulated that the skin response and possible skin damage is strongly related to the magnitude of the impact load and its resulting stresses.

Materials and methods

Biaxial load applicator

The biaxial load applicator (Figure 1) was designed to launch an impact body on a surface with both vertical and horizontal velocity component. To accomplish the combined horizontal and vertical translation, the impact body is placed on a horizontal (Figure 1(a)) and a vertical rail (Figure 1(b)). The impact height of the impact body (Figure 1(c)) can be adjusted to a maximum of 0.5 m, which corresponds to a maximum velocity of 3.0 ± 0.2 m s\textsuperscript{-1} which is
Overview of the equations used to derive the mechanical parameters:

\[ a_1 = \text{calibration constant of the vertical accelerometer of the upper mass (cm/s)}; \]
\[ a_2 = \text{vertical acceleration of the upper mass (m/s^2)}; \]
\[ i_1 = \text{impact body mass (kg)}; \]
\[ k_2 = \text{calibration constant of the vertical accelerometer of the lower mass (cm/s)}; \]
\[ a_3 = \text{vertical acceleration of the lower mass (m/s^2)}; \]
\[ a_4 = \text{lower mass (kg)}; \]
\[ k_3 = \text{calibration constant of the horizontal accelerometer of the impact body (cm/s)}; \]
\[ a_5 = \text{horizontal acceleration of the impact body (m/s^2)}; \]
\[ a_6 = \text{contact area of the skin (cm^2)}; \]
\[ t = \text{instantaneous time during free fall (s)}; \]
\[ t_{\text{impact}} = \text{time of impact (s)}; \]
\[ s_{\text{max}} = \text{maximum horizontal displacement during impact (m)}; \]
\[ s = \text{instantaneous horizontal displacement during impact (m)}; \]

The horizontal velocity of the interconnected (Figure 1(e)) impact body can be adjusted by setting the drop height of the deadweight (Figure 1(d)) to a maximum of 0.57 m. The corresponding maximum horizontal velocity is 1.9 ± 0.2 m/s which is derived from the horizontal accelerometer. In reference, the typical initial horizontal velocities measured during sliding tackle experiments are in the range of 3–4.5 m/s (van den Eijnde et al. 2016). The vertical drop is initiated when the impact body passes an electromagnetic switch (Figure 1(g)) that is positioned on the horizontal rail. The position of the switch can be adjusted and depend on the drop height of the deadweight allowing a free fall of the impact body.

The impact body consists of a mass-spring configuration, which resembles the human body impact. The configuration is able to absorb energy during impact, resulting in a longer contact time compared to a single rigid body impact (Nikooyan and Zadpoor 2011). The typical impact time of the knee during a sliding on artificial turf, derived from biomechanical tests, is 30 ± 5 ms (van den Eijnde et al. 2016). The lower mass (Figure 1(h)) has a weight of 2.09 ± 0.02 kg and the upper mass (Figure 1(i)) can be adjusted from 1.15 ± 0.02 kg in steps of 0.47 kg to a maximum of 3.5 ± 0.04 kg. In this study, the upper mass is held constant at 2.09 ± 0.02 kg. The two masses are interconnected with a spring (Figure 1(j)), which has a spring constant of 10 ± 0.5 N/mm.

The clamping system (Figure 1(k)) for the ex vivo skin is similar to the specimen holder used in the Martindale test (ASTM 2004), for testing the abrasion resistance of textile fabrics. The contact area of the skin is 8 cm^2. The tested surfaces (Figure 1(l)) are mounted on to the outer frame of the apparatus.

**Instrumentation (load measurement)**

The biaxial load applicator contained three capacitive spring-mass accelerometers (type B3, Seika.de, Germany) with a range of −50 g to 50 g and corresponding resolution of <2 10^−2 g and a linearity deviation of <0.5% adjusted on the impact body (Figure 1). The accelerometers are connected to a wireless 7 channel analog input sensor node (type V-link-LXRS, Lord MicroStrain, USA). A USB data gateway (Node Commander, Lord MicroStrain, USA) collects the synchronized data from the wireless sensor node. Data logging software (Node Commander, Lord MicroStrain, USA) is used for node programming, data acquisition and data analyses at a sampling rate of 800 Hz. The accelerometer data are used to derive the velocity, reaction forces and corresponding peak stresses in both directions and the kinetic energy before impact according to the equations listed in Table 1. In wear analyses often the Archard wear equation is applied which asserts that the wear volume is directly proportional to the kinetic energy before impact.

### Table 1.

<table>
<thead>
<tr>
<th>Mechanical parameters</th>
<th>Equation</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vertical velocity before impact</td>
<td>( v_z = \int \frac{a_2 \cdot a_1 , dt}{t_{\text{impact}}} )</td>
<td>m/s^−1</td>
</tr>
<tr>
<td>Horizontal velocity before impact</td>
<td>( v_y = \int \frac{a_2 \cdot a_1 , dt}{t_{\text{impact}}} )</td>
<td>m/s^−1</td>
</tr>
<tr>
<td>Normal force</td>
<td>( F_z = a_1 \cdot a_2 \cdot M_1 + a_1 \cdot a_2 \cdot M_2 )</td>
<td>N</td>
</tr>
<tr>
<td>Shear force</td>
<td>( F_y = a_1 \cdot a_2 \cdot (M_1 + M_2) )</td>
<td>N</td>
</tr>
<tr>
<td>Normal stress</td>
<td>( \sigma_z = F_z / A_1 )</td>
<td>N/cm^2</td>
</tr>
<tr>
<td>Shear stress</td>
<td>( \tau_y = F_y / A_1 )</td>
<td>N/cm^2</td>
</tr>
<tr>
<td>Normal work during impact</td>
<td>( W = \int F_z \cdot ds )</td>
<td>Nm</td>
</tr>
<tr>
<td>Kinetic energy before impact</td>
<td>( E_{\text{kin}} = \frac{1}{2} (M_1 + M_2) \cdot (v_z^2 + v_y^2) )</td>
<td>Nm</td>
</tr>
</tbody>
</table>

**Figure 1.** Schematic illustration of the biaxial load applicator during the four sequential phases: (I) starting position of the horizontal translation, (II) start of the free drop phase induced by an electromagnetic switch, (III) free horizontal and vertical translation phase and (IV) the impact phase (left). Photo of the impact body including the positioning of the accelerometers, wireless inputs node and clamping system (right).
product of the normal load and the horizontal sliding distance, the so-called normal work during impact (Williams 1999).

To verify the repeatability of the vertical and horizontal reaction force signals, a calibration set-up was used consisting of a force plate (Kistler Instruments, Switzerland, type 9282E; dimensions 60 × 60 cm²) and a mounted spring. The spring constant of the mounted spring was 26 ± 0.5 N mm⁻¹. Different impact conditions per impact direction were applied. Three drop heights of deadweight (Figure 1(f)), 10, 30 and 47 cm, resulted in three horizontal impact conditions. In addition, three drop heights of the impact body (Figure 1(c)) 3, 10 and 20 cm, resulted in three different vertical impact conditions. These tests were repeated three times to finalize calibration.

Testing surfaces and conditions

Two different playing surfaces were tested, natural grass and a third-generation artificial turf system. The artificial turf system was tested both dry and wet. The wet condition was obtained by manually spraying, in total with 0.4 l water. The artificial turf system, with a size of 120 × 50 cm², had a monofilament type of polyethylene fibre with a length of 50 mm and contained 35 mm commercial thermoplastic elastomer (TPE) infill material with a granulometry from 0.5 to 2 mm and bulk density of 0.38 kg m⁻³. According to the FIFA method 4 (FIFA 2008a), the shock absorption of this artificial turf pitch was determined to be 56 ± 2% and had an energy restitution of 38 ± 3%. Additionally, the artificial turf pitch met all requirements for FIFA-2-STAR. The natural grass with a size of 50 × 50 cm² was cultivated and constructed according to DIN 18035-4. The total thickness of the grass sod was 75 ± 5 mm and the grass length was 40 ± 5 mm. All playing surfaces were conditioned at 20 ± 2°C and relative humidity of 50 ± 5% for 5 days before testing.

Nine different impact conditions were used per surface condition. The load protocol is summarized in Table 2. All tests were performed under laboratory conditions at 50 ± 5% relative humidity and 20 ± 2°C.

Ex-vivo skin model as read-out

In this study, rabbit ears were used as ex-vivo model (Jung and Maibach 2014). The ears were obtained from a slaughterhouse in Gent (Belgium) and were conserved at 5°C for a maximum of 48 h. The skin samples were cut from the ears using a circular cutting punch with a diameter of 38 mm.

To study the skin morphology, samples (15 × 5 mm²) were taken perpendicular to the horizontal load direction immediately after impact. After fixation in 4% formalin, the skin samples were embedded in paraffin. Paraffin sections (6 μm), cut by microtome (Leica, Microsystems SP 1600, Nussloch, Germany), were deparaffinized with histosafe (Adamas, Rhenen, The Netherlands) followed by rehydration in decreasing concentrations of alcohol (100–50%). The sections were hematoxyline-eosin (HE) stained for morphological assessment. In total, two HE-stained sections per test condition were used for analyses.

The HE-stained tissue sections were examined using a microscope (Axioskop2 MOT; Zeiss) equipped with a digital photo camera (AxioCam MRc5; Zeiss) with a resolution of 2584 × 1936 pixels and AxioVision software (Zeiss). With the used magnification, we were able to examine a field of view of 2.8 × 2.1 mm². From each HE-stained section, two images were taken on different locations.

First, the microscopic images were visually assessed on the presence of epidermal skin damage. Furthermore, quantitative analyses were performed using image software (ImageJ 1.49v, National Institutes of Health, USA) to calculate the average thickness of the stratum corneum. Per microscopic image, the average thickness of the stratum corneum was determined on three different locations covering the full length of the skin section.

Statistics

The Pearson correlation coefficient was used to study the relationship between the absolute measured stratum corneum thickness and the following mechanical parameters: vertical (vₓ) and horizontal velocity (vᵧ) before impact, kinetic energy (Ekיני) before impact, normal (σᵧ) and shear peak (τᵧ) stress and normal work during impact (W).

The interpretation of the correlation coefficient (r) was as follows: |r| < 0.1: very weak correlation, 0.1 ≤ |r| < 0.3: weak correlation, 0.3 ≤ |r| < 0.5: moderate correlation, |r| ≥ 0.5: strong correlation (Cohen 1988). To evaluate the influence of the surface conditions upon the relation between the absolute measured stratum corneum thickness and the mechanical parameters, a correlation comparison analysis was performed using Fisher’s Z-transformation (Weaver and Wuensch 2013). The Shapiro–Wilk test was applied to check the normality. All analyses were performed with SPSS 23.0 for Windows.

Results

Validation and calibration of the apparatus

The performance of the biaxial load applicator had acceptable levels of repeatability with respect to the measured peak forces as illustrated in Table 3. Both horizontal and vertical peak forces were properly reproduced. Comparison of the peak forces of both the force plate and accelerometer data of the biaxial load applicator revealed that the force measurements of the biaxial load applicator are within 10% accuracy.

<table>
<thead>
<tr>
<th>Test run</th>
<th>f [cm]</th>
<th>c [cm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-2</td>
<td>57 ± 0.3</td>
<td>20 ± 0.3</td>
</tr>
<tr>
<td>3-4</td>
<td>29 ± 0.3</td>
<td>20 ± 0.3</td>
</tr>
<tr>
<td>5-6</td>
<td>10 ± 0.2</td>
<td>10 ± 0.2</td>
</tr>
<tr>
<td>7-8</td>
<td>57 ± 0.3</td>
<td>10 ± 0.2</td>
</tr>
<tr>
<td>9-10</td>
<td>29 ± 0.3</td>
<td>10 ± 0.2</td>
</tr>
<tr>
<td>11-12</td>
<td>10 ± 0.2</td>
<td>10 ± 0.2</td>
</tr>
<tr>
<td>13-14</td>
<td>57 ± 0.3</td>
<td>3 ± 0.2</td>
</tr>
<tr>
<td>15-16</td>
<td>29 ± 0.3</td>
<td>3 ± 0.2</td>
</tr>
<tr>
<td>17-18</td>
<td>10 ± 0.2</td>
<td>3 ± 0.2</td>
</tr>
</tbody>
</table>
Qualitative observations

Visual inspection of the skin samples tested on both dry and wet artificial turf showed a pattern of sharp grooves and pits (Figure 2). The grooves are mainly orientated parallel to the sliding direction. This pattern was seen after every run. After run 1 and 2 on both wet and dry artificial turf, it was noticed that infill was left behind on the skin. Skin samples tested on natural grass showed a pattern of grooves after run 1–8. These grooves were less clear compared to the tests on artificial turf. After run 1, 2, 7 and 8, pieces of grass were present on the skin and the skin was stained green.

Assessment of the microscopic images revealed that the stratum corneum of the skin samples subjected to dry and wet artificial grass were disrupted and damaged after test run 1–10 and 1–7, respectively. Skin samples subjected to natural grass showed no signs of damage or disruption of the stratum corneum, except for test run 2.

Quantitative observations

The results of the visual microscopic assessment and both the measured impact velocities and peak stresses are shown in Figure 3. It is seen that the nine conditions resulted in nine different reproducible combinations of horizontal and vertical impact velocities.

The different impact conditions resulted in combined peak shear and normal stress levels ranging from 18 up to 150 N·cm⁻². The shear-normal peak stresses as well as the measured impact velocities on all test surface conditions are comparable.
The results show that skin breakdown on dry artificial turf occurs at lower impact velocity combinations compared to the wet condition and natural grass. Skin breakdown on dry artificial turf occurred at a shear-normal stress combination of 38 and 20 N·cm⁻². In the wet condition, skin breakdown firstly occurred at a shear-normal stress combination of 55 and 20 N·cm⁻². On natural grass, skin breakdown only occurred at a shear-normal stress combination of 130 and 125 N·cm⁻².

The stratum corneum thickness measurements on skin subjected to dry artificial turf revealed that the stratum corneum thickness significantly strongly correlated (Table 4) with the horizontal velocity component \( r = -0.649, P < 0.01 \), the kinetic energy before impact \( r = -0.665, P < 0.01 \), normal work during impact \( r = -0.663, P < 0.01 \), the normal peak stress \( r = -0.555, P = 0.017 \) and shear peak stress \( r = -0.678, P < 0.01 \). In the wet condition, only a significantly moderate

Table 4. Results of the Pearson correlation (\( r \)) and correlation comparison analysis (\( Z \)) between the absolute measured stratum corneum thickness after impact and the mechanical parameters: vertical \( \nu_z \) and horizontal \( \nu_y \) velocity before impact, kinetic energy before impact \( E_{\text{kin}} \), normal \( \sigma_z \) and shear \( \tau_y \) peak stress during impact and work during impact \( W \) of dry artificial turf \( (n = 18) \), wet artificial turf \( (n = 18) \) and dry natural grass \( (n = 18) \).

<table>
<thead>
<tr>
<th></th>
<th>Artificial turf (dry)</th>
<th>Artificial turf (wet)</th>
<th>Natural grass (dry)</th>
<th>Correlation comparison</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \nu_z ) (m/s⁻¹)</td>
<td>-0.299</td>
<td>-0.164</td>
<td>-0.257</td>
<td>-0.393</td>
</tr>
<tr>
<td>( \nu_y ) (m/s⁻¹)</td>
<td>-0.649**</td>
<td>-0.064</td>
<td>-0.063</td>
<td>-1.085</td>
</tr>
<tr>
<td>( E_{\text{kin}} ) (Nm)</td>
<td>-0.665*</td>
<td>-0.054</td>
<td>-0.663</td>
<td>-1.125</td>
</tr>
<tr>
<td>( W ) (Nm)</td>
<td>-0.663**</td>
<td>-0.054</td>
<td>-0.663</td>
<td>-1.125</td>
</tr>
<tr>
<td>( \sigma_z ) (N·cm⁻²)</td>
<td>-0.555*</td>
<td>-0.372</td>
<td>-0.337</td>
<td>-1.104</td>
</tr>
<tr>
<td>( \tau_y ) (N·cm⁻²)</td>
<td>-0.678**</td>
<td>-0.372</td>
<td>-0.337</td>
<td>-1.104</td>
</tr>
</tbody>
</table>

*Significant at the 0.05 level (2-tailed).
**Significant at the 0.01 level (2-tailed).
correlation was found with the shear peak stress \( (r = -0.479, P = 0.044) \). No significant correlations were found with respect to the thickness of the stratum corneum and the mechanical parameters for skin tested on natural grass.

When comparing the correlations coefficients of the different mechanical parameters and the measured stratum corneum thickness after impact between the three different surface conditions, no significant differences were found.

**Discussion**

Essential biomechanical data of the tolerance of skin to impact with sport surfaces are lacking at the moment. With the aid of a newly developed biaxial load applicator and using rabbit ears as a model for human skin, it was possible to systematically evaluate skin damage over a range of impact loads and stresses by measuring the thickness of the stratum corneum after impact and microscopic assessment.

From the correlation study, it was found that the thickness of the stratum corneum when tested on dry artificial turf is strongly related to the magnitude of the horizontal velocity and kinetic energy before impact and resulting in normal and shear peak stress and normal work during impact. Only a weak to moderate relationship was observed for natural grass and wet artificial turf. The correlation comparison analysis between the three surface conditions showed no significant differences. This indicates that although differences in correlation strength were found with respect to tested surface conditions, the stratum corneum thickness after impact is reciprocal to the magnitude of the applied mechanical load conditions. It must be noted that the measured stratum corneum thickness is an average value and can be seen as an indication for skin damage.

The actual skin damage was qualitatively evaluated. It was observed that skin damage during impact on dry artificial turf occurred at a lower shear-normal stress combination (38 and 20 N·cm\(^{-2}\)) compared to a wet artificial turf condition (55 and 20 N·cm\(^{-2}\)) and natural grass (130 and 125 N·cm\(^{-2}\)). These qualitative findings are in line with the results of the comparison correlation study. Both show that skin breakdown and resulting reduction of the stratum corneum occurred on all tested surface conditions only at higher mechanical loading conditions with respect to wet artificial turf and natural grass. The restricted loading range explains the differences in correlation strength found between the tested surface conditions. This is in accordance with player perception studies, where players complain more about the abrasiveness of artificial turf than natural grass (Roberts et al. 2014). The results are also in line with the clinical findings described by Peppelman et al., where soccer slidings performed on artificial turf resulted in more abrasions than on natural grass (Peppelman et al. 2013).

By macroscopic visual assessment, we have found a specific pattern of sharp grooves in the sliding direction and pits on the skin samples tested on artificial turf. It is plausible that the pits are caused by the indentation by the infill material during contact. The grooves have the same width and depth as the pits. From a health perspective, not only skin damage but also the intensive contact with infill material is of interest. In particular, the contact with recycled rubber granules where a number of hazardous substances have been reported (Cheng et al. 2014; Pavilonis et al. 2014). Based on the current evidence available, the European Chemicals Agency has concluded that there is a very low level of concern from exposure to these substances (ECHA 2017).

Zanetti et al. have already shown that the perceived abrasiveness is influenced by the type of infill: with regard to abrasion, players prefer Styrene Butadiene Rubber over TPE infill (Zanetti 2009). For future research, it would be interesting to compare different types of infill and even non-infill systems to determine the influence on the load tolerance of skin by using the developed biaxial load applicator.

At higher loads, the skin samples tested against natural grass stained green and showed a less sharp, less dense and less deep groove like pattern compared to skin tested against artificial turf. The green staining most likely comes from the chlorophyll pigment present in the plant chloroplasts. The local failure of grass blades together with the plant chloroplasts functioning as a sort lubricant are important beneficial properties of natural grass over artificial turf in reducing the risk of abrasion injuries.

The possibility to control many variables, which cannot be controlled in humans in vivo, is a major advantage of the proposed experimental set-up using ex-vivo skin as readout. Although an ex-vivo rabbit ear mimics the human skin very well, it has a limited storage time and histological evaluation using skin samples is tedious and time consuming. In this perspective, the development of multilayered synthetic epidermal skin equivalents is looking promising, also because automated roughness measurements can be used instead of histological analyses as read-out of the surface damage (Morales-Hurtado et al. 2015).

The correlation study showed that the reduction of the stratum corneum is strongly related to the shear loading conditions when tested on dry artificial turf in contrast to a wet surface condition or dry natural grass. This indicates that the degree of skin damage is not only related to the level of stress but also to the characteristics of the counter surface. From an engineering point of view, skin abrasion injuries are a result of a wear process, and wear is typically defined as the loss of material from a surface by the contact and relative motion with a solid, liquid or gaseous counter body (Masen 2004; van Der Heide et al. 2013). One of the wear mechanisms of interest is classified as abrasive wear and occurs when a solid object is in sliding contact with a harder rough counter material. In general, abrasive wear results from scratching and/or micro-cutting. Although hardness and/or roughness are not considered in this study, the observed differences in tolerance of the skin between artificial turf and natural grass can probably be explained by these wear parameters. This means that tribological investigations on component level of yarns and infill can contribute to a better knowledge of the skin-turf friction when surface roughness and possibly the hardness are also taken into account (Hurtado et al. 2016; Tay et al. 2016).

Unfortunately, current research in the field of skin-friendly artificial turf surfaces is concentrated on friction coefficient measurements. It is assumed that the level of friction is correlated to the skin abrasion (Sanchis et al. 2008; Zanetti et al.
However, there is no simple correlation between friction and skin damage (wear). In a qualitative way, it seems reasonable to expect relatively more skin damage in case of high frictional forces but it is quite possible for material combinations to produce very similar frictional forces but very different wear behaviour (Williams 1999). Sanchis et al. showed that there was no strong correlation between the coefficient of friction and damage when using a silicone rubber skin replacer. When tested on different artificial turf surfaces, similar coefficients of friction resulted in different roughness values of the worn rubber (Sanchis et al. 2008). It was suggested that other mechanisms than friction are responsible for the observed damage.

In the interaction between skin and artificial turf, a number of phenomena occur simultaneously, both on a macroscopic and a more localized microscopic scale (Masen 2011). These mechanisms include adhesion between the two surfaces, lubrication, deformation of the skin, the fibres as well as the infill material and micro-ploughing and scratching. The combination of these mechanisms results in friction in the contact as well as in damage to the skin. Whilst that means that both the experienced friction and the resulting skin damage both originate from these basic mechanisms, there is no causal relationship between the level of friction in the contact and the damage to the skin and no obvious quantitative correlation exists.

With more than 1500 installed artificial soccer pitches and over 1.2 million active soccer players in the Netherlands alone, the social relevance of skin injury prevention research is obvious. Specially, when taking in account that artificial pitches for recreational use are seldom watered. Setting minimum standards for sliding friendliness by sport-governing bodies will not only improve the pleasance of playing but will also reduce related health-care costs. In addition, it will create business opportunities for innovative sliding-friendly concepts.

In conclusion, this study provides unique biomechanical data of the load tolerance of skin to impact on dry and wet artificial turf and natural grass. The developed insights are valuable for manufacturers of artificial turf in defining the design space. Additionally, it helps governing bodies in setting standards regarding the sliding friendliness of artificial turf.

**Disclosure statement**

No potential conflict of interest was reported by the authors.

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**References**


**Practical implications**

Wetting of artificial turf system improves the load tolerance of skin to impact on dry and wet artificial turf and natural grass. The developed insights are valuable for manufacturers of artificial turf in defining the design space. Additionally, it helps governing bodies in setting standards regarding the sliding friendliness of artificial turf.

**Acknowledgment**

This study was conducted with financial contribution from the Dutch Ministry of Economic Affairs, Agriculture and Innovation and the Provinces of Gelderland and Overijssel. Further, we would like to thank Wouter Bijnens of the Faculty of Health Medicine and Life Sciences, of the Maastricht University Medical Centre for his contribution during the repeatability experiments.


