Development of a test method to evaluate laceration risk of studded footwear

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I dedicate this thesis to my loving family.

Abstract

Studded footwear has previously caused a number of severe laceration injuries in rugby union. Current test methods for assessing the laceration injury risk of rugby stud designs are unrepresentative of the game and are not mandatory for manufacturers to follow. The aim of this project was to develop a new, game-representative test method to assess the laceration injury risk of stud designs used in rugby union.

First, the prevalence of skin and laceration injuries in rugby union was assessed through a systematic literature review of epidemiological studies. It was found that 2.4 skin injuries occurred per 1000 match hours, which could be interpreted as one time-loss injury per team, per season.

A survey study of 191 rugby players was then conducted, indicating that stamping in the ruck was the most prevalent cause of stud laceration injuries. Following this, twelve participants were asked to perform stamping impacts in a simulated rucking scenario. Three-dimensional shoe kinematics and individual stud kinetics were measured for each impact. Two key phases were identified: an initial impact phase, and a subsequent raking phase.

A two-phase mechanical test method was developed based on the results of the stamping study. In the initial impact phase, the stud is attached to a pendulum impacting a skin simulant. The velocity, stud angle and mass of the impact can be adjusted. The stud and skin simulant are then moved to the second phase, performing a controlled rake. In this phase, raking speed, stud angle and stud mass can be changed.

Finally, six studs were compared on their predicted laceration injury risk using the developed method. Four of the tested studs were bespoke designs incorporating different edge radii and top diameters. The developed test method showed an increased laceration injury risk when stud edge radius or top diameter was reduced. Two of the tested studs were commercially available designs which had previously passed rugby union's current studded footwear tests. One of commercial studs showed an increased risk of laceration in the developed test method. Future research should focus on improving the developed test method's validity and investigating the influence of stud material, shape and wear on laceration injury risk.

Acknowledgements

Throughout this project I have been lucky enough to have met and been supported by a great number of people. I would first like to thank my supervisors, Dr David James, Dr Heather Driscoll and Dr Marcus Dunn for their support and patience over the last three years. Dave, thank you for making time for me when I needed it and for helping me focus on the end goal. Heather, without your enthusiasm for this project I would not have written this thesis today. Thank you for your wisdom and honest advice throughout the years. And Marcus, thank you for critically challenging me, which has led me to become a better researcher.

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Publications

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Table of Contents

Nomenclature

Abbreviations

3D	Three-Dimensional
ACL	Anterior Cruciate Ligament
ATD	Anthropomorphic Test Device
CAD	Computer Aided Design
CI	Confidence Interval
FIFA	Fédération Internationale de Football Association
IFAB	International Football Association Board
ITEH	Injuries per Thousand Exposure Hours
PDS	Product Design Specification
PMHS	Post Mortem Human Subject
PRISP	Professional Rugby Injury Surveillance Project
PVA	Polyvinyl Acetate
RLIF	Rugby League International Federation
SD	Standard Deviation
STAR	Skin Tear Audit Research
UK	United Kingdom
USA	United States of America

Terminology

Aetiology	The cause, or manner of causation, of an injury.
Anisotropic	The property of being directionally dependent.
Backs	Referring to playing positions 9 - 15 in the rugby game, players are not involved with the scrum.
Biofidelic	Accurately modelling a biological system.
Effective mass	Proportion of body segment mass that is contributing to an impact.
Epidemiology	Study of factors determining and influencing the frequency of disease, injury, and other health-related events.

Ex-vivo	Tissue from an organism in an external environment.
Football	Unless otherwise specified, refers to association football / soccer.
Forwards	Referring to playing positions 1 - 8 in the rugby game, where players are involved with the scrum.
Frangible	Simulant which breaks under load, opposite of durable.
Impact	The action of one object coming forcibly into contact with another.
Injury	Transfer of energy to the tissue that exceeds the ability to maintain structural and / or functional integrity.
Injury prevalence	Mean frequency of an injury per specified time-interval (i.e. 1000 hours).
In-vivo	In a living organism.
κ	Fleiss' kappa.
Kinematics	Study of motion in terms of angles, positions, velocities and accelerations.
Kinetics	Study of motion taking into account forces and torques.
Laceration	Cut or incision.
p	Probability of correctly rejecting the null-hypothesis in statistics.
Raking	When a player scrapes another player in the ruck with their shoes.
Ruck	A phase of play in rugby where players from each team, who are on their feet are in physical contact with one another, use their feet to try to win or keep possession of the ball.
Rugby	Unless otherwise specified, refers to 15-a-side rugby union.
Skin graft	Transplantation of skin or skin-like tissue in patients.
Speed	The rate at which someone or something moves, one-dimensional.
Stamping	A purposeful movement where the player brings their foot heavily down on another player.

Tribology	Friction, lubrication and wear of interacting surfaces in relative motion.
Velocity	The speed of something in a given direction, multi-dimensional (vector).
Viscoelasticity	Substance with elastic and viscous component, giving the substance a strain rate dependence on time.

Mathematical operators

E_{kin}	Kinetic energy (J)
F	Force (N)
g	Gravitational constant (m/s^2)
h_i	Impact height (m)
h_s	Starting height (m)
k	Spring constant (N/m)
m	Mass (kg)
m_e	Effective mass (kg)
t	Time (s)
v	Velocity (m/s)
v_i	Impact velocity (m/s)
v_s	Starting velocity (m/s)

1 Introduction

The following chapters describe a three-year study on the development of a new, game-relevant test method for stud designs that can be used to evaluate the risk of laceration injuries. The test method is developed for stud designs used in the game of rugby union.

1.1 Motivation for research

Participating in sport has numerous health benefits such as increased quality of life and reduced risk of disease (Haskell *et al.* 2007). Sports injuries can counter these benefits if an athlete is unable to continue to participate (Maffulli *et al.* 2011). Rugby union is one of the fastest growing team sports; it has seen a 19% increase in members between the 2007 and 2011 World Cups (Rugby World 2011). Rugby union is reported to have 7.2 million players in 2014 (World Rugby 2014). The sport has one of the highest injury incidences of all team sports. In one study, rugby union's self-reported injury rate was 96 injuries per 1000 occasions of participation, which was higher than association football (64 injuries per 1000 occasions) and field hockey (62 injuries per 1000 occasions) (Nicholl *et al.* 1995). An estimated 72% of injuries in rugby union are a result of player-to-player contact (Brooks *et al.* 2005b), thus indicating that the full-contact nature of the sport is a major risk factor for injury. High injury rates in sport are a pressure on health services and discourage people from participating (Bazelmans *et al.* 2004). Increasing awareness of the risks involved in rugby union has led its governing body World Rugby to fund research to monitor the effects of rule changes on injury rates (Fuller *et al.* 2009) and to analyse the cause of frequent injuries (Fuller *et al.* 2005).

Rule changes in rugby union often aim to reduce injury rates among players. In 1996, a note was supposedly added to rugby union's lawbook at law 21 -16.3(f), stating that a player must ruck for the ball, and not intentionally ruck players on the ground (Moore 2010; Rugby referee forum 2015). This suggests a growing concern about injuries sustained by raking and stamping in the ruck. More recently, a trend towards using artificial playing surfaces has increased the abrasion injury prevalence in rugby union (Williams *et al.* 2015). This highlights the shortcomings of current injury prevalence recording methods for recording skin injuries (van den Eijnde *et al*. 2014b).

As in most field-based sports, rugby players wear studded footwear to increase traction. Traction is defined as the grip between the shoe and playing surface. The amount of traction is dependent on the ground surface as well as the soleplate of the shoe. Hence, sports played on different surfaces will require differing outsole designs to attain ideal traction. Generally, softer surfaces require longer studs to increase traction levels through penetration. With harder surfaces it can be better to increase contact area instead of stud length and use the properties of adhesive friction between the studs and the surface (Barry & Milburn 2013). In rugby union natural grass is still the dominant playing surface, though artificial turf is becoming increasingly common. Both natural grass and artificial turf surfaces allow for penetration of the studs. Figure 1.1 shows a variety of studs that are commonly used in rugby union. The specific type of stud utilised depends on playing position, pitch conditions and personal preference.

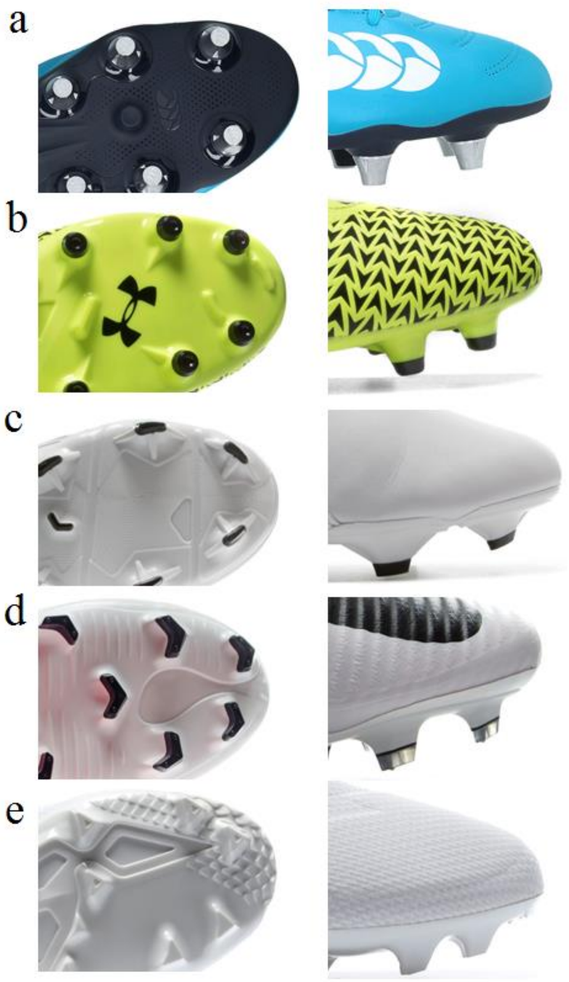


Figure 1.1: Common stud designs found on rugby shoes as per October 2017; a) aluminium screw-in studs, b) moulded rounded studs, c) blades, d) chevrons and e) triangular studs. Photos adapted from Lovell Rugby (2017).

The influence of these widely variable stud designs on laceration injury risk has not previously been investigated. Studded footwear injury research has traditionally focussed on injuries to the wearer, such as metatarsal injuries and anterior cruciate ligament (ACL) injuries (Ford *et al.* 2006; Grund & Senner 2010; Milburn & Barry 1998). Media articles have suggested that specific stud designs are more likely to cause laceration injuries (Aarons 2013; Hills 2013; McGeady 2017). Blades, an oblong shaped stud design (type C, Figure 1.1) known for increasing players' turning speed, have been available to players for over two decades (Briant 1997). There is a widely held belief that this particular stud design has an increased laceration injury risk. Hall & Riou (2004) raised a cause for concern, citing two case-studies showing severe laceration injuries resulting from bladed studs. The head coach of Manchester United Football Club called for a ban on blades in 2005 (BBC Sport 2005). A small number of rugby clubs decided to forbid their players to wear bladed studs following the media attention around this ban, especially in youth divisions (Old Elthamians RFC 2011; Alton RFC 2012). In 2008, a laceration injury to the head measuring 13 cm in length led a football player to sue a major shoe manufacturer over their 'dangerous stud design' (Dennehy 2008; Heylin 2016). Again, the stud design in question was a blade.

These events have raised concerns on the injury risk associated with modern stud designs and have highlighted the shortcomings of World Rugby's current test methods for studded footwear. The test method regarding safety aspects of rugby shoe outsole design is described in Regulation 12, Schedule 2 (World Rugby 2015). It is currently not mandatory for stud designs to pass the studded footwear tests described in Regulation 12, even in professional rugby.

In the studded footwear tests of Regulation 12, the damage on a skin simulant material caused by the tested stud design is benchmarked against the damage caused by a 'comparator stud'. The comparator stud is a standardised stud that has been identified as possessing an acceptable injury risk by World Rugby. The test method that is currently in place has not been updated since its original formation in 1990 despite a rapid increase in the number of new stud designs. The test is based on the heel-striking impact of an 80 kg man running over a force plate (M Douglas, pers. comm., 21 January 2015). Given the nature of the rugby game, a heel-striking running impact is unlikely to be representative of stud laceration injury scenarios in rugby union. Further, the description of the test parameters described in Regulation 12 is incomplete. The regulation fails to specify what skin simulant material should be used and does not provide a full specification of the test methodology.

This research project focusses on developing a game-representative test method for assessing the laceration injury risk of studded footwear in rugby union. Such a method could equally be applied to other field sports using similar footwear, although the test parameters could differ.

In this work, the sports injury prevention model by van Mechelen (1997) was used as a theoretical framework. The model describes four phases to successful injury prevention which are shown in Figure 1.2. This thesis considers phase one (establishing the extent of the injury problem) and phase two (establishing the aetiology and mechanisms of sports injury) of van Mechelen's model. The groundwork of phase three (introducing a preventive measure) will be undertaken with the development of a new test method. The final implementation of phase three will remain dependent on the governing body of the sport introducing a rule or regulation change, or on outsole manufacturers adjusting their stud designs with the aim to lower injury risk. According to van Mechelen's model, after the introduction of a preventive measure, its effectiveness should be assessed by re-evaluating the extent of the injury risk (i.e. repeating phase 1).

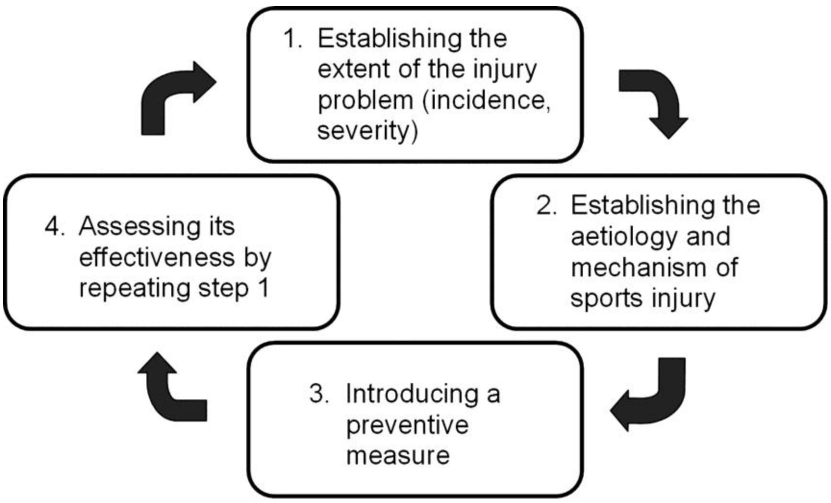


Figure 1.2: The sequence of sports injury prevention (adapted from van Mechelen 1997).

1.2 Aim and objectives

Aim:

To develop a test method to assess the laceration injury risk of stud designs used in rugby union.

Objectives:

1. Investigate the current prevalence of skin laceration injuries in rugby union.

2. Identify the most prevalent game scenario in rugby union that leads to laceration injuries caused by studded footwear.

3. Determine the kinetics and kinematics of stud-skin impacts during the game scenario identified in Objective 2.

4. Design and build a mechanical test for assessing stud laceration risk based on the kinetics and kinematics measured.

5. Compare the laceration injury risk of stud designs using the developed method, thus providing a proof-of-concept.

1.3 Research structure

Figure 1.3 describes the six stages (A - F) that were identified for this research project. The first stage (A) reviews current literature associated with laceration injuries resulting from studded footwear. Stage A involves five separate sub-stages:

1. Injuries linked to stud design.
2. Current regulations for studded footwear in rugby union, rugby league and association football.
3. Measurement methods for obtaining kinetic and kinematic data on sports injury events.
4. Anatomy and mechanical failure of human skin.
5. Materials frequently used to substitute in-vivo human skin.

The second stage (B) establishes the prevalence of laceration injuries resulting from stud-skin interactions in rugby union. This stage is divided into a systematic review and meta-analysis of skin and laceration injury prevalence based on the existing literature, and a retrospective, self-reported injury prevalence based on primary data collected in a survey. Stage C identifies common play scenarios causing stud laceration injuries in rugby, through the same survey. Stage D investigates the kinetics and kinematics of the most common play scenario associated with these injuries. In stage E, a new test method is developed based upon the impact parameters identified in stage D. Stage F provides a proof-of-concept for the newly developed test method.

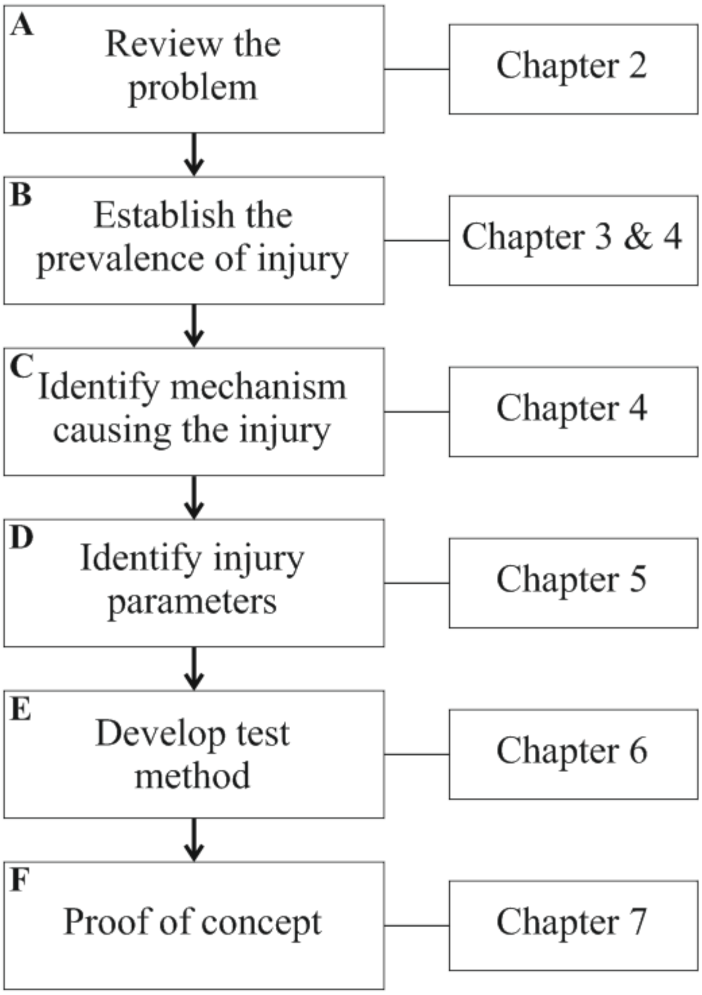


Figure 1.3: The six stages (A - F) of the stud laceration risk research project.

2 Literature review

2.1 Introduction

This chapter provides an overview of published literature on studded footwear injury mechanisms, studded footwear regulations, measurement methods of injury biomechanics, human skin properties and skin simulants. Footwear used in rugby union is often similar to footwear used in rugby league and association football due to comparable traction requirements in the sports. Therefore, relevant literature for rugby union, rugby league and football are included in this chapter.

2.2 Studded footwear injuries

Rugby union is a sport with one of the highest number of injuries per playing hours (Nicholl *et al.* 1995). As a consequence, there have been a number of studies investigating the prevalence of injuries in rugby and their potential causes. These studies have often focussed on injuries in professional rugby, such as the longitudinal Professional Rugby Injury Surveillance Project (PRISP, Rugby Football Union 2017), which have been assessing injury trends in professional rugby union since 2002. A systematic review and meta-analysis of 10 epidemiological studies showed that the average match injury prevalence of professional players was 81 injuries per thousand exposure hours (ITEH) (Williams *et al.* 2013). In comparison, professional football's injury prevalence during match play is estimated at 27 ITEH (Ekstrand *et al.* 2011). The PRISP report showed that match injury prevalence in professional English rugby over the last 15 years had stabilised at 84 injuries per 1000 player hours, which relates to 1.6 injuries per club, per match (Rugby Football Union 2017). The PRISP report does not report detailed information on skin injuries, injuries caused by studded footwear and injury trends in amateur rugby.

With relation to studded footwear, three potential causes of injury can be identified: 1) shoe-surface interactions, where excessive traction has the potential to cause knee- or ankle ligament injuries; 2) shoe-foot interactions, where stud placement and shape can lead to pressure distributions which contribute to metatarsal injuries; 3) shoe-skin interactions, in which a stud comes into contact with another player, causing a skin or laceration injury. Each of these studded footwear injuries are explained in the following sections.

2.2.1 Shoe-surface interactions

The frictional force, or resistance against slipping, between the shoe outsole and ground surface is known as traction. In field sports such as rugby, football and hockey, studs on the outsole are used to obtain 'optimal traction'. For optimal shoe performance, traction should be high enough to avoid slipping, though excessive traction can lead to injury (Wannop *et al.* 2010). Translational traction relates to the linear resistance to slipping of the outsole, whilst rotational traction refers to the resisting of pivoting and turning movements. It is generally accepted that high translational traction increases performance by avoiding slippage, whilst high rotational traction can be a risk factor for knee and ankle injuries (Torg & Quedenfeld 1971; Milburn & Barry 1998; Shorten *et al.* 2003; Thomson *et al.* 2015). Stud design has been thought to influence both translational and rotational traction (Villwock *et al.* 2009), though it is unclear whether translational and rotational traction can be independently controlled (Shorten *et al.* 2003; Wannop *et al.* 2010).

The amount of traction between the shoe and the ground surface can be determined through mechanical testing, player testing and computational modelling. Mechanical testing provides repeatable results which allow direct comparisons between the performances of outsole designs. The validity of mechanical test methods for traction has previously been criticised for being constricted to laboratory measurements and / or for using unrepresentative test parameters (Nigg 1990). In 2007, Grund *et al.* addressed both these issues by developing the TrakTester, a portable device allowing field-based measurements (Figure 2.1). The device was developed based on estimates of the kinetics and kinematics of real-life ACL injuries taken from broadcast footage. Despite potential validity issues with mechanical test methods, in-field player testing is not often used to measure traction. Player testing produces higher variability than mechanical testing, and this variability is likely to mask any potential effect of subtle changes in outsole design. Player testing is further restricted by ethical considerations: traction cannot be tested to injurious levels. Computational models on shoe-surface interactions are challenging to develop and rely on accurate simulation of complex soil mechanics

(Driscoll *et al.* 2012; Barry & Milburn 2013). However, when implemented successfully, they can be used to predict the traction of prototype studs without the need of physically manufacturing them.

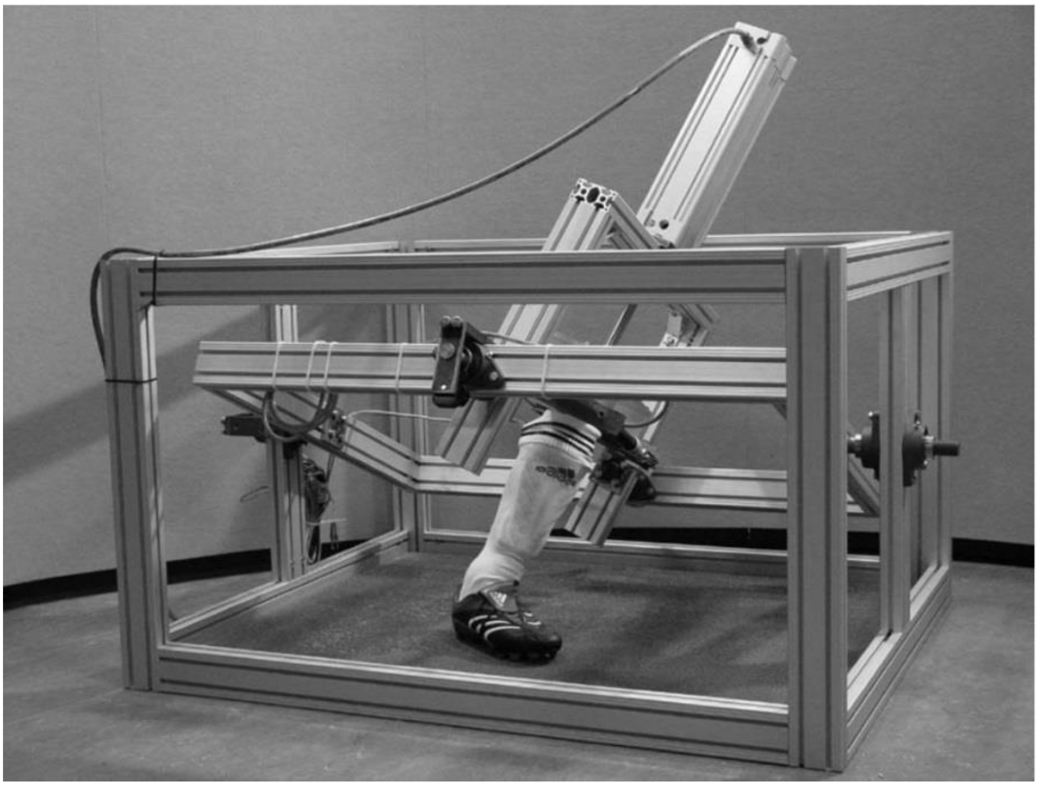


Figure 2.1: The 'TrakTester' was developed to quantify traction at realistic test conditions (adapted from Grund *et al.* 2007).

2.2.2 Shoe-foot interactions

Stud shape and configuration can have an influence on the pressure distribution inside the shoe. Continued, localised pressure on the foot is thought to increase the risk of metatarsal stress fracture in runners (Gross & Bunch 1989). A case study of 23 metatarsal fractures showed 35% of the injuries were sustained by athletes wearing studded footwear (Porter *et al.* 2005). Queen *et al.* (2008) tested four different stud patterns shown in Figure 2.2 on their plantar pressure distribution during agility courses. The tested outsoles included a 'bladed' stud type. The findings of this study did not conclusively link one of the stud types to a higher risk of metatarsal injuries. The authors suggested that increased cushioning in the forefoot region could help decrease loading on the metatarsal bones.

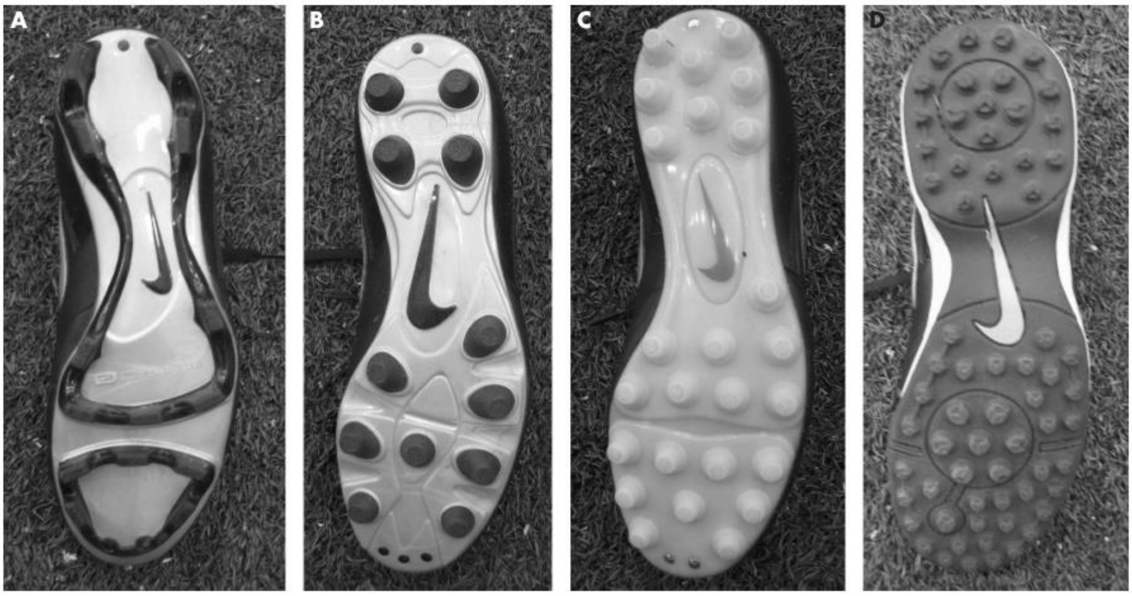


Figure 2.2: Four stud configurations used by Queen *et al.* (2008): a) bladed, b) firm ground, c) hard ground and d) artificial turf stud.

In a later study, Bentley *et al.* (2011) investigated the peak plantar pressure of conventional (i.e. conical) studs versus bladed studs in 29 male football players, running a straight line and a slalom section. It was found that the peak pressure on the medial side of the foot was higher when wearing conventional studs, whilst the pressure on the lateral side of the foot was increased when wearing the bladed shoes. The authors concluded that this increased lateral loading of the foot in bladed shoes is potentially more hazardous to the player than increased medial loading, thus predisposing the player to metatarsal injuries when wearing bladed shoes.

2.2.3 Shoe-skin interactions

A third type of injury caused by studded footwear is skin injuries. Though minor skin injuries such as chafes or abrasions resulting from shoe-skin interaction in rugby union are common and an accepted part of the game, more severe lacerations can lead to time-loss, infection, and disfiguration (Figure 2.3). Injuries resulting from shoe-skin interactions have not been well documented. One case study has called for attention to the potential of increased laceration injury risk of modern stud designs (Hall & Riou 2004), but no further research has been published. A test method to assess the skin injury risk of studded footwear is present in rugby union, and this method is further discussed in Section 2.3. Other field sports using studded footwear, such as football and rugby league, do not have test methods in place to evaluate stud laceration risk.

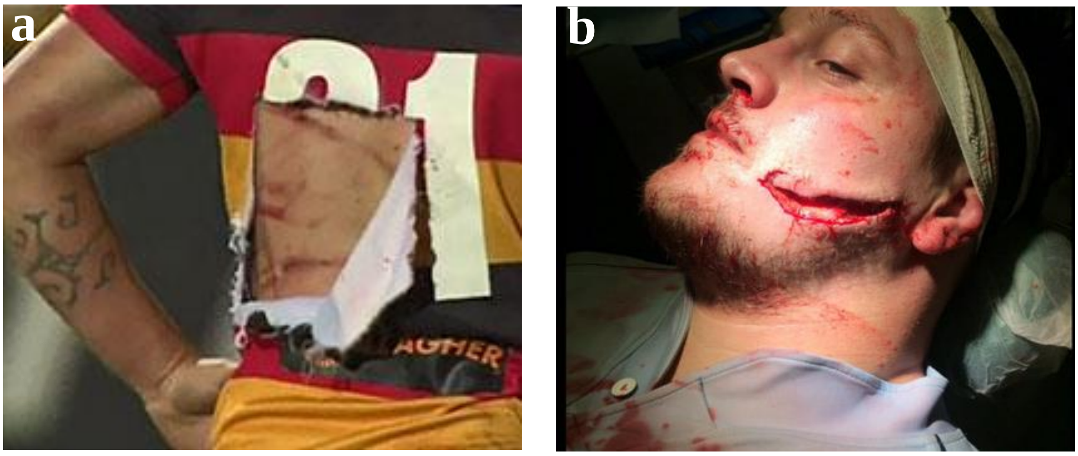


Figure 2.3: Skin injuries resulting from studded footwear. a) A graze or abrasion is usually not defined as an injury; b) Example of laceration requiring medical attention.

2.2.4 Summary

Rugby and football are field sports with a high injury prevalence. Although the overall injury prevalence in rugby and football has been well researched, no studies have specifically focussed on skin or laceration injury prevalence. Three types of injuries related to studded footwear in rugby and football have been identified: 1) shoe-surface interactions, 2) shoe-foot interactions, and 3) stud-skin interactions. The injury mechanism of knee and ankle ligament injuries are thought to be related to shoe-surface interactions and metatarsal injuries are commonly associated with shoe-foot interactions. These injury mechanisms have been better understood than laceration injuries in rugby and football, which are potentially related to shoe-skin interactions.

2.3 Studded footwear regulations

Governing bodies enforce regulations relating to sports equipment to maintain the integrity of the sport, avoid unwanted competitive advantage and to safeguard the players. The first known stud regulations in rugby date back to 1889, stating *"No one wearing projecting nails or iron plates on any part of his boots shall be allowed to play in a match"* (Rugby Football History 2007). The laws of the game of rugby union are determined by World Rugby, and in rugby league the regulations are the responsibility of Rugby League International Federation (RLIF). In association football, the

International Football Association Board (IFAB), this has representatives from the English, Northern Irish, Welsh and Scottish Football Associations and the Fédération Internationale de Football Association (FIFA), controls the rules of the game. Rules regarding outsole design differ between these three governing bodies. The relevant regulations on the design of studded footwear are described in the following section.

2.3.1 Rugby union

The design of studs worn in rugby union is regulated by the sport's international governing body, World Rugby (Dublin, Ireland). World Rugby has published their equipment rules in Regulation 12: Provisions relating to players' dress (World Rugby 2015). In this regulation, two performance tests (Test A and Test B, pp. 219 - 220) are used to evaluate the laceration injury risk of individual studs. Although published in World Rugby's regulations, these tests are currently optional for manufacturers to follow. The current test parameters have not been validated to replicate injurious scenarios of rugby play. In both tests, damage caused by the tested stud design is compared to damage from a standardised comparator stud (Figure 2.4). This comparator stud has been identified as the benchmark for 'acceptable injury risk' by World Rugby. The shape and dimension of the tested stud design should be so that it presents no greater risk of injury to another player than this defined comparator stud. Both tests require a 'suitable human flesh simulant'; a 1.5mm thick poromeric shoe upper material combined with gelatine is suggested in the protocol. In addition to passing the optional test protocol, studs used in rugby union should be no longer than 21 mm, with all edges finished smooth and rounded to a minimum radius of 1 mm. British Standard 6366: 2011 (British Standards 2011) is based on World Rugby's stud performance tests and can be interpreted as the same test protocol.

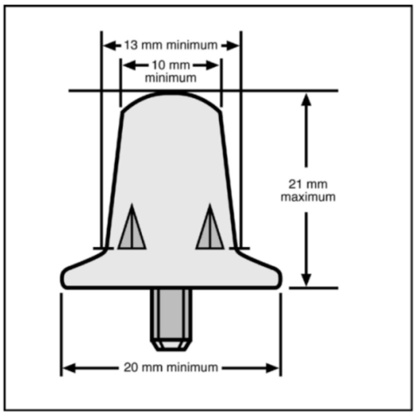


Figure 2.4: Acceptable dimensions for conical or cylindrical studs. A stud within the limits of the dimensions shown can act as a comparator stud in testing protocols (adapted from World Rugby 2015).

Performance test A: Skin glancing or raking

The first performance test relating to the laceration injury risk of studded footwear is a pendulum impact (Test A, World Rugby 2015, p219). The test aims to simulate a glancing or sliding blow of a stud on skin. An example of such a situation in the game of rugby is during 'raking' in the ruck, a movement in rugby where a player aims to get the ball out of a ruck by kicking it backwards. In Test A, a single stud is connected to a pendulum device with a spring-like feature in the attachment of the stud (Figure 2.5). The spring is needed to achieve the 70 mm minimal sliding distance of the stud. In the test, the stud is dropped and dragged onto a suitable skin and soft tissue simulant combination. The simulant materials are not proscribed. The inbound velocity and impact mass are not specified for this test. For a stud design to pass Test A, the damage inflicted by the stud on the skin and soft tissue simulant should be compared to the damage inflicted by a comparator stud (Figure 2.4). It is not further specified how this damage should be defined or measured.

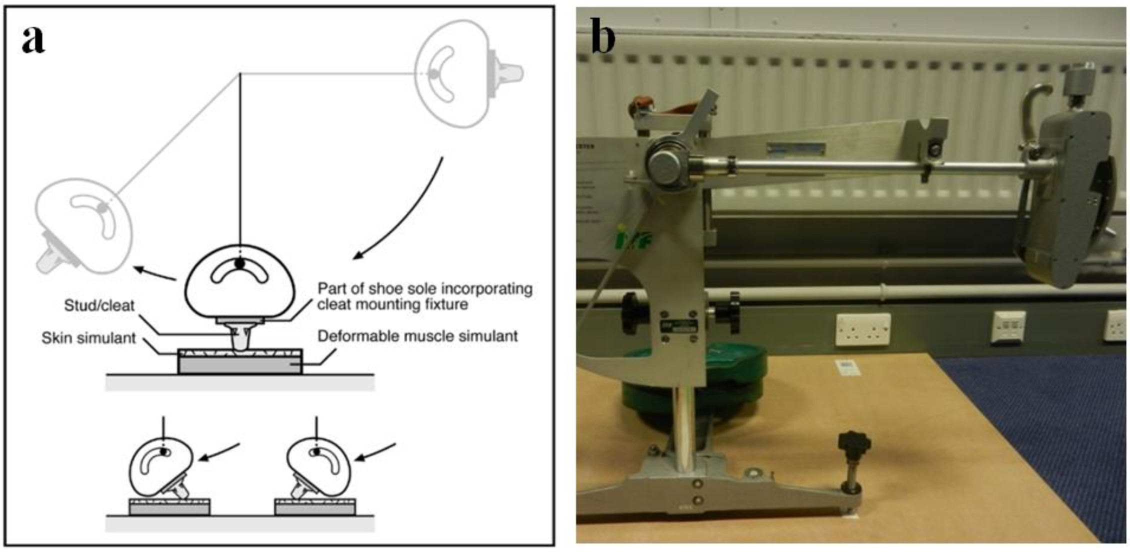


Figure 2.5: a) Schematic representation of the simulated glancing and raking test (adapted from World Rugby 2015); b) Test set-up for glancing and raking test, using a modified skid-testing pendulum.

Performance test B: Skin stamping

The second performance test relating to the laceration injury risk of studded footwear is a drop test (Test B, World Rugby 2015, p220). The test aims to simulate a stamping impact. Stamping is a purposeful movement where the player brings their foot heavily down on another player who is lying on the ground. In Test B, stamping is simulated using a linear drop hammer device, in which a stud is attached to a mass and dropped onto a skin and deformable tissue simulant (Figure 2.6). The drop height and drop mass is not prescribed, though a height of 50 mm and weight of 8.5 kg is suggested to be suitable. Again, a suitable skin and tissue simulant needs to be placed underneath the stud. The penetration depth of the tested stud design should be compared to the penetration depth inflicted by the comparator stud. Furthermore, a visual comparison of the difference in damage to the skin simulants should be made.

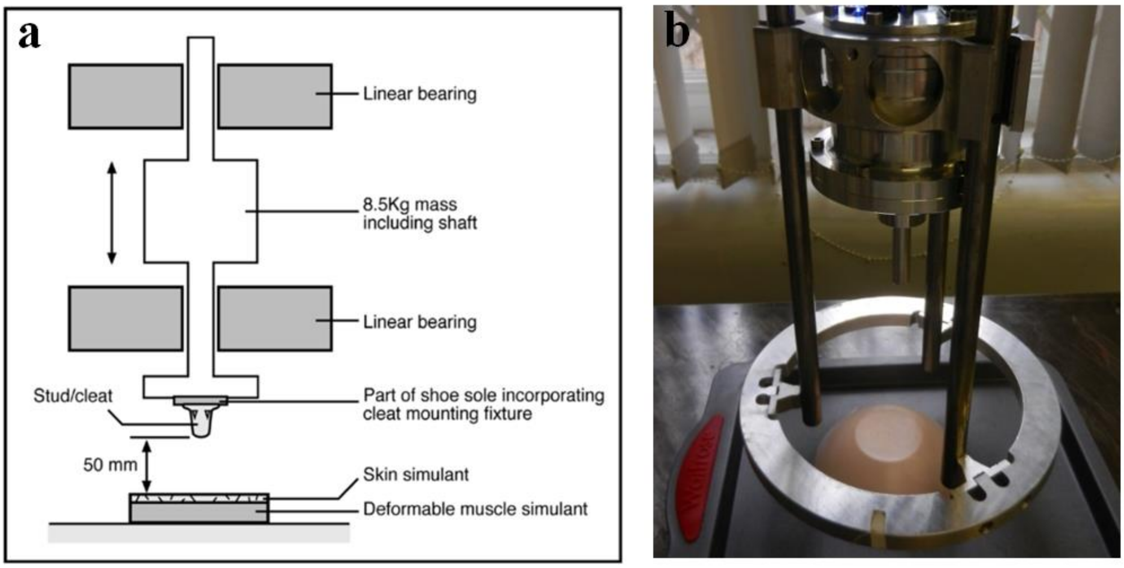


Figure 2.6: a) Schematic representation of the simulated stamping test (adapted from World Rugby 2015); b) Test set-up of the simulated stamping test using a custom designed drop hammer and silicone soft tissue simulant.

2.3.2 Rugby league

Traction demands of footwear used in rugby league is similar to rugby union, especially for the backs. In Law 4(f) of the Rugby League International Laws of the Game (Rugby League International Federation 2013) regarding studded footwear, it is stated that studs cannot be less than 8 mm diameter at the apex and that if the studs are made out of metal, they should have rounded edges. The regulations do not further specify what the minimum edge radius of 'a rounded edge' is. Law 4(f) would exclude most modern stud designs, such as bladed or triangulated studs, since their diameter at the apex is usually less than 8 mm (Figure 2.7).

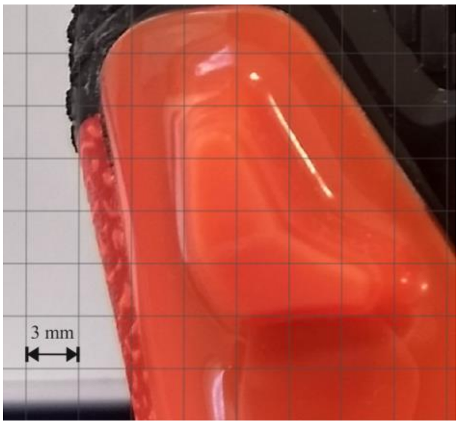


Figure 2.7: Modern stud design, measuring approximately 3 mm at the apex.

2.3.3 Association football

Many rugby shoes that are marketed as appropriate for firm ground surfaces are originally designed as football shoes (Lovell Rugby 2017). Rugby and football players are therefore likely to use similar footwear; both sports are played on natural grass as well as artificial turf and in both cases the shoes need to provide traction to its wearer. The 'Laws of the game 2014/2015' are authorised by the International Football Association Board (IFAB) and published by the Fédération Internationale de Football Association (FIFA). This document states in Law 4: Regarding Players' Equipment, Safety (p21) that players must not use equipment or wear anything that would be dangerous to themselves or another player. This law is not specific to studs, the outsole or the shoe as a whole, and allows for interpretation of 'dangerous' by players, manufacturers and referees. There are no other specific guidelines stating quantifiable measures for stud shape or design internationally accepted in football.

However, there are guidelines for testing football shin guards which include a stud impact. Shin guards are mandatory when playing competitive football in the United Kingdom (UK). The main aim of a shin guard is to protect the tibia against impact. In the UK, shin guards must adhere to BS EN:13061:2009 (British Standards 2009). The stud impact test in BS EN:13061:2009 aims to simulate a slide tackle in football. The test uses a metal stud with a diameter of 10 mm and an edge radius of 0.5 mm. The test set-up, consisting of a horizontal and a vertical cone test, is shown in Figure 2.8. The horizontal cone test (Figure 2.8a) mounts the shin guard on a freely suspended cone weighing 5 kg. The stud is attached to an impact mass of 1 kg, and subsequently dropped onto the shin guard-cone combination with an inbound speed of 5.4 m/s. In the vertically inclined cone test (Figure 2.8b), the shin guard is mounted on a rigid cone, angled at 10° from the vertical. The same impact mass and inbound velocity is required. A shin guard fails when the inner surface of the guard has torn or perforated during either of the tests.

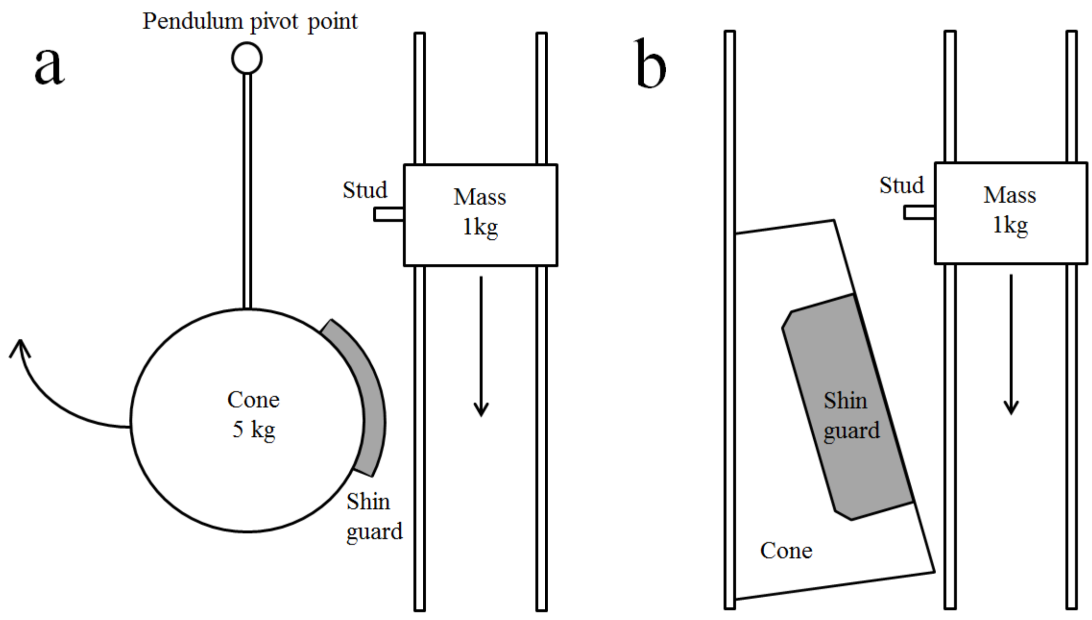


Figure 2.8: Test set-up of stud impact test for shin guards in BS EN: 13061:2009. Arrows indicate movement directions during test. a) Horizontal cone test; b) Vertically inclined cone test.

The impact parameters used in BS EN:13061:2009 have not been informed by biomechanical parameters of injury scenarios. Ankrah & Mills (2003) used a different set of impact to test the efficacy of shin guards for protecting against stud-shin impacts. In their study, the effective mass was estimated at 0.1 kg based on the segmental mass of a foot, and inbound velocity of the impact was taken similar to speeds seen in a placed ball football kick. An earlier comparison of different shin protectors by Francisco *et al.* (2000) aimed to simulate the foot-player contact that happens during a slide tackle. The slide tackle was identified as the most common shin injury mechanism. This injury scenario is associated with lower inbound velocities than the study of Ankrah & Mills, and shin guards were tested at 1.7 - 2.7 m/s in combination with an impactor mass of 4.2 kg. Both the inbound velocity and the impact mass were based on estimates of slide tackle impacts without biomechanical measurements. The limited justification for test parameters, in both regulations and in shin guard comparison studies, suggests insufficient knowledge about the kinetics and kinematics of stud-player impacts.

2.3.4 Summary

Prior to a rugby match, stud checks are performed by the referee to determine if the studs are safe enough to play with. Rugby union has a testing protocol in place for assessing stud laceration injury risk, outlined in Regulation 12. The test parameters in the laceration injury risk tests of Regulation 12 have either not been informed by biomechanical measurements of laceration injury scenarios or are altogether missing. The validity of the current test method is therefore questionable. Other field sports such as rugby league and association football use similar footwear to rugby union. Rugby league regulations stipulate that studs with a diameter smaller than 8 mm at the apex are not allowed, but do not have any further testing protocols in place for stud designs. Football does not have any specific regulations for safety of stud design. Football shin guards need to comply with BS EN: 13061:2009, which incorporates a stud impact test. This test does not assess the injury risk of a stud design but measures the ability of the shin guard to protect against generic stud impacts. Again, the impact parameters used in this test were not based on biomechanical measurements. It can be concluded that the test methods in regulations that are currently in place for studded footwear have insufficient justification for their test parameters.

2.4 Injury mechanics

2.4.1 Methodological approaches

The second step in the injury prevention sequence by van Mechelen (1997) (Figure 1.2) describes the need to establish the aetiology and mechanism of the sports injury. The type of play scenario causing laceration injuries in rugby union had not previously been investigated. Krosshaug *et al.* (2005) identified eight methods to assess injury mechanics in sport: athlete interviews, clinical studies, video analysis, laboratory motion analysis, in-vivo force measurements, accidental injuries during experiments, cadaver and dummy studies, and mathematical modelling. Krosshaug *et al.* recommend using multiple methods to in order to describe all aspects of the injury situation. When determining the test parameters of a new test method to assess laceration injury risk of studs, injuries during experiments are unethical. Cadaver and dummy studies will not be able to inform the mechanism of injury, and mathematical models need real-life experimental data for their validation. Clinical studies predominantly evaluate wound size and severity and are unlikely to determine the impact kinetics and kinematics causing the injury. Analysing match footage of injury events captures data of real-life injury situations and can be used to calculate kinematic information, for example by using a model-based image-matching technique for uncalibrated video footage (Krosshaug & Bahr 2005). This type of approach is valuable for non-contact injuries where kinetic estimates can be made through inverse mechanics (Grund *et al.* 2007). Athlete interviews and questionnaires in isolation are unlikely to provide reliable biomechanical information on the injury event, though this method has been found effective for determining an accurate description of the injury event (Arnold *et al.* 1979; Kobayashi *et al.* 2010). The identified event can subsequently be simulated in laboratory studies, where kinetic and kinematic data on the injury event is then captured.

2.4.2 Kinematic measurement methods

As described in the previous section, laboratory measurements of the kinetics and kinematics of laceration injury events can be used to inform the design of a new test method. Optical, electromagnetic, ultrasonic or inertial systems are available to measure the kinematics of the absolute motion of participants (Chris Kirtley 2006). The most optimal system for measuring kinematic data depends on many factors, such as the measurement frequency needed, the type of movement, the accuracy that is required and the physical area that the movement will take place in.

Optical systems such as high speed cameras and motion capture systems have been used to obtain displacement, velocity, angles and angular velocity during scrummaging (Preatoni *et al.* 2012). The advantage of optical systems is that there is minimal interference for participants, with systems usually being wireless. One of the disadvantages is the risk of marker occlusions, especially when multiple participants are performing a relatively uncontrolled motion (e.g. a tackle).

Inertial systems use accelerometers to obtain kinematic data from sports injury scenarios (Hendricks *et al.* 2012; Withnall *et al.* 2005). Inertial systems can be used wireless and these systems do not have marker occlusions (Morris 1973). Their output is an acceleration signal which needs to be integrated over time to obtain velocity or double integrated to obtain displacement. Integrating a signal means it accumulates error, so velocity or position measures can become increasingly unreliable over time within a trial.

Ultrasonic and electromagnetic systems are used less often than optical and inertial systems; electromagnetic systems can suffer from disturbances caused by metal which is difficult to eliminate in a research setting (C Kirtley 2006). Although susceptible to occlusions, optical systems can provide high accuracy combined with minimal interference to the participant. Using high speed cameras as the optical measurement system can further add qualitative information on the injury event, if necessary.

2.4.3 Kinetic measurement methods

The force during sport impact events is commonly measured with or derived from force plates (Lieberman *et al.* 2010), load cells (Peterson *et al.* 2008), strain gauges (Preatoni *et al.* 2012) or pressure sensors (Halkon *et al.* 2014). Although load cells and force plates can give accurate overall force readings at a high sampling frequency, they do not provide information about the distribution of force, e.g. the force per stud when multiple studs are in contact. Force plates are heavy, rigid measurement tools which are not suitable for placing on curved surfaces such as a person.

Lightweight and flexible piezo-resistive pressure sensors (Tekscan F-scan 3000E, Boston, USA) have previously been used to measure impact forces of potential injury scenarios in basketball and rugby players (Pain *et al.* 2008; Halkon *et al.* 2012; Halkon *et al.* 2014). The thin, flexible sensors can be used safely in impact situations where a rigid measurement device could harm the participants. This type of sensor is also able to provide a spatial distribution profile of the impact force by giving a reading of the pressure per sensing element. On the downside, pressure sensors do not provide the same accuracy of total force that can be achieved using force plates or load cells (Wirz *et al.* 2002). The accuracy of pressure sensors has been shown to improve when bespoke calibration methods for the application method were developed (Cazzola *et al.* 2013; Brimacombe *et al.* 2009; Halkon *et al.* 2012; Oudshoorn *et al.* 2016b). The calibration method needs to be specific to the loading rate and expected force range for the most accurate results (Cazzola *et al.* 2013; Wirz *et al.* 2002; Oudshoorn *et al.* 2016b).

2.4.4 Calculating effective mass

Examining the kinetics of impacts is challenging due to the segmental and muscular complexity of the human body (Lenetsky *et al.* 2015). The term 'effective mass' is often used in sports science literature to describe the proportion of body (segment) mass that is contributing to the impact (Chi & Schmitt 2005; Lenetsky *et al.* 2015; Rousseau & Hoshizaki 2015). Effective mass is used because athletes cannot accurately be described in terms of rigid blocks of mass. Upon impact, our bodies can deform, reducing the impact forces (Gruber *et al.* 1998). Spring-damper models have been used to describe the combination of rigid structures (bone) and deformable structures (soft tissue) in the human body (Derrick *et al.* 2000).

There is no direct way to measure the effective mass of an impact; however, a variety of approaches have been used to calculate effective mass from biomechanical measurements (Lenetsky *et al.* 2015). Following Newton's second law, the effective mass of hand striking impacts in combat sports was defined through Equation (2.1), (Neto *et al.* 2012);

$$m_e = \frac{\int_{t1}^{t2} F(t)}{\Delta v} \tag{2.1}$$

With m_e being effective mass of the impact, $F(t)$ being force as a function of time, t_1 the time of first impact, t_2 the time that the hand stops momentarily during collision and Δv the change in velocity of the striking object during this time. Equation (2.1) has also been found effective at determining the mass of a known impact mass (Kessler *et al.* 2003).

2.4.5 Summary

The test parameters for a sport injury prevention test method should be derived from biomechanical measurements of the injury scenario. Identifying this injury scenario and measuring its kinetics and kinematics is an important step in the development of a new test method for studded footwear. Previous studies have used a range of optical and inertial measurement methods to obtain kinematics of sports impacts. Optical systems provide an accurate and easy to use solution if marker occlusions can be avoided.

Impact kinetics is most commonly measured with: force plates, load cells, strain gauges and pressure sensors. Pressure sensors output a pressure distribution per sensing element which is required if one wants to measure the force of individual studs. The effective mass is a common term used to describe the proportion of segmental body mass that contributes to an impact. To calculate the effective mass of an impact, a combination of kinetic and kinematic data is used.

2.5 Human skin

2.5.1 Skin properties

The human skin is the largest organ in our body, accounting for approximately one-eighth of our body weight (Wood & Bladon 1985). Its functions include providing a barrier against infections and damage from the outside world, restricting fluid loss, regulating body temperature and providing a sensory surface. Figure 2.9 shows the three main layers of human skin: the epidermis, the dermis and the hypodermis (Stafleu van Loghum 2009; Wood & Bladon 1985). The outermost layer, the epidermis, consists predominantly of dead skin cells. This layer can be damaged without causing bleeding or a pain response. The middle layer, the dermis, supplies blood flow to the skin cells and contains nerve endings. The deeper hypodermis consists mainly of fat cells, which have a different indentation response than the dermal layer. It contains larger blood vessels and works as a protective layer for underlying tissue. Damaging the skin as far as the hypodermis is likely to cause severe bleeding and produce scarring tissue (Stafleu van Loghum 2009; Wood & Bladon 1985).

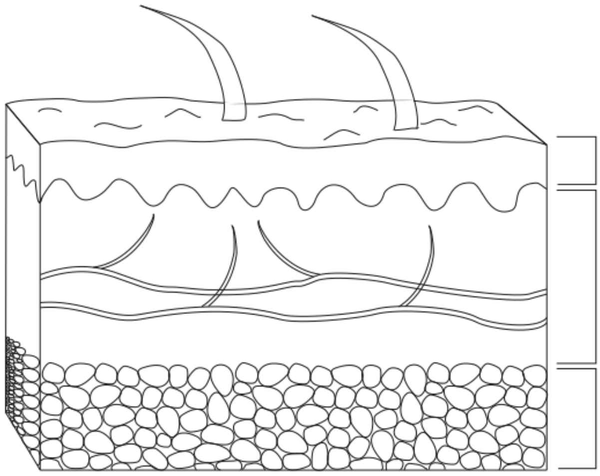


Figure 2.9: Human skin can be divided up into three layers, the epidermis, dermis and hypodermis.

The dermal layer is most important for protecting the body from injury. It consists of collagen fibres which provide mechanical strength, mixed with elastin to create tension and elasticity in the skin (Wood & Bladon 1985). The mechanical properties of human skin vary from person to person and are specific to the body location (Wood & Bladon 1985; Zhang & Mak 1999). However, estimates of five mechanical properties frequently used to quantify skin behaviour are summarised in Table 2.1.

Table 2.1: Estimates of mechanical properties of human skin.

Property	**Human skin**	**Reference**
Thickness (dermis and epidermis)	2 - 5 mm	(Wood & Bladon 1985)
Young's modulus	0.1 - 0.3 MPa 0.3 - 1.0 MPa	(Shergold & Fleck 2004) (Liang & Boppart 2013)
Tensile strength	3 - 14 MPa 10 - 20 MPa	(Jansen & Rottier 1958) (Shergold & Fleck 2004)
Elongation at break	100% 65% 64 - 102%	(Shergold & Fleck 2004) (Ankersen *et al.* 1999) (Jansen & Rottier 1958)
Tear strength	2 - 20 kN/m	(Shergold & Fleck 2004)
Density	1176 kg/m^3	(Shergold & Fleck 2004)

2.5.2 Wound types

Mechanical damage to the skin can result in different wound types. An overview of four commonly seen wound types and their definition is given in Table 2.2.

Table 2.2: Overview of four commonly used skin injury categories.

Category	**Definition**	**Reference**
Abrasion	Superficial removal of epidermal and / or dermal layer cells, produced by contact of exposed skin with the environment.	(Basler *et al.* 2001)
Laceration	A wound produced by the tearing of body tissue, as distinguished from a cut or incision.	(Miller-Keane & O'Toole 2005)
Contusion	Injury to tissues with skin discoloration and without breakage of skin; called also bruise.	(Miller-Keane & O'Toole 2005)
Blister	A fluid-filled, thin-walled structure under the epidermis or within the epidermis.	(Farlex Partner Medical Dictionary 2012)

Skin-surface contact in rugby and football often leads to abrasive injuries, a common problem especially for pitches with artificial turf (van den Eijnde *et al.* 2014a; van den Eijnde *et al.* 2014b; Twomey *et al.* 2014). The systematic review by *van den Eijnde et al.* (2014b) describes both the incidence and mechanism of abrasive injuries sustained in football, and gives recommendations on improving the reporting of skin injuries in field sports. Lacerations can be caused by two types of tissue damage: tensile strain injuries and shear strain injuries (Figure 2.10) (Viano *et al.* 1989). The stretch that both tensile and shear strain puts on blood vessels can cause them to rupture. When a blood vessel ruptures and the protecting skin layer breaks with them, loss of blood occurs causing a laceration. When the skin layer stays intact, the blood will stay contained in the skin and will cause a bruise or contusion.

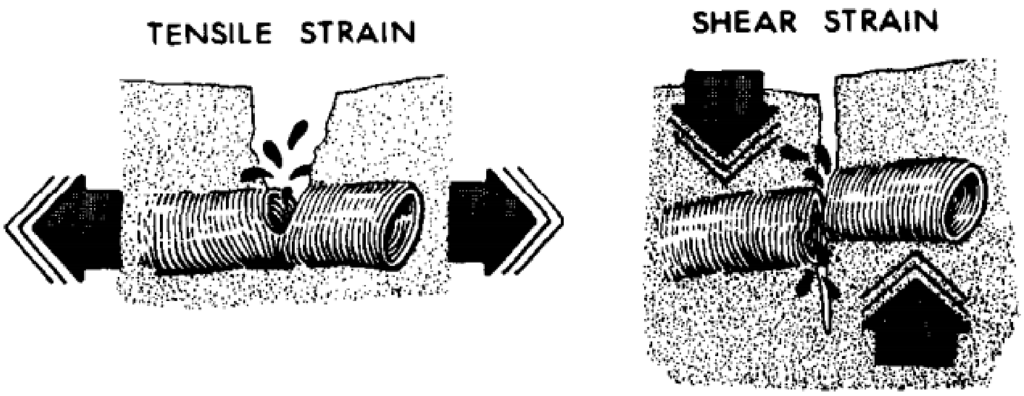


Figure 2.10: Tensile and shear strain can both cause the blood vessels and protecting skin layer to tear, causing a laceration injury (adapted from Viano *et al.* 1989).

A model on the resistance of skin again tearing has previously been presented by Yang *et al.* (2015), which carefully presents the mechanics of skin tearing. Once the skin is broken, the collagen fibres inside the dermal layer hinder wound propagation, or further 'tearing' of the skin (Yang *et al.* 2015). The relatively high tear resistance of skin comes from a process in which collagen fibres and their fibrils straighten and stretch along the tension axis, as shown in Figure 2.11. This resistance against tearing remains even when an initial puncture is made; if this is the case, the wound tends to yawn rather than propagate (Figure 2.12). This re-aligning of collagen fibrils can explain the viscoelasticity and anisotropy of mammalian skin.

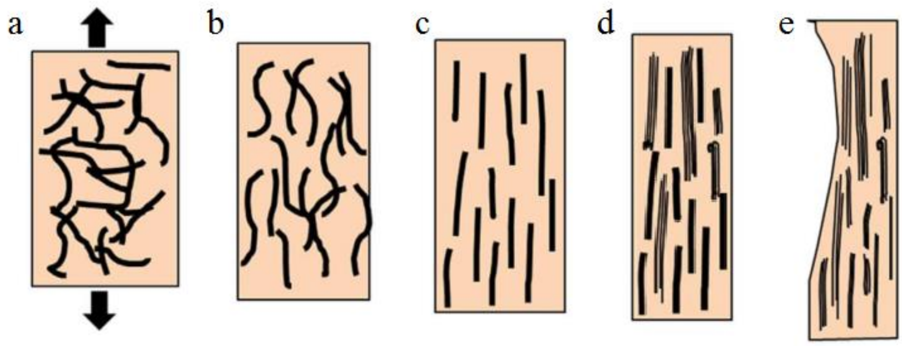


Figure 2.11: Straightening of collagen fibrils in the skin provide high tear resistance. a) Orientation before loading; b) rotating of fibrils; c) straightening; d) stretching, sliding and delaminating; and e) fracture (adapted from Yang *et al.* 2015).

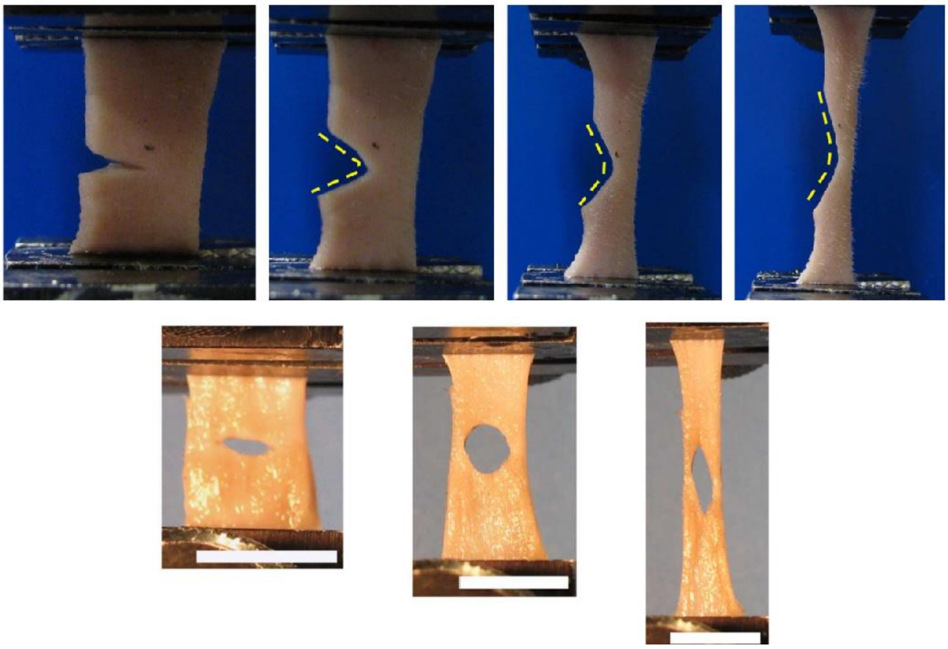


Figure 2.12: Wound propagation of side notch (top) and central notch (bottom) under tension (adapted from Yang *et al.* 2015).

Previous forensic studies have attempted to define the amount of energy, force or pressure needed to break skin. Bir *et al.* (2005) expressed the skin penetration threshold in energy per unit area (J / cm^2). A rubber bullet with impact surface area of 2.45 cm^2 was fired at eight different cadavers in 25 impact locations. It was found that the penetration threshold of human cadaver skin was between 24 - 53 J/cm^2. The high velocity at which these values were obtained (61 - 183 m/s) make them not directly transferable to sport-specific, lower velocity impacts because skin is loading rate dependent (Wood & Bladon 1985).

At lower inbound velocities, stabbing studies using both sharp and blunt instruments have developed mechanical models of skin penetration (Shergold & Fleck 2004; Shergold & Fleck 2005). This research showed that the penetration mechanism was dependent on the geometry of the punch tip. Generally, a larger cross-sectional area was associated with a larger penetration force, though specific impactor shape also played a role (Parmar *et al.* 2012; Shergold & Fleck 2005). It is likely that a similar effect can be found with the geometry of a stud.

2.5.3 Wound measurement

In the current studded footwear regulations and standards (World Rugby 2015; British Standards 2011) it is stated that the tested stud should 'not cause greater damage' onto a skin simulant than the defined comparator stud (see also Section 2.3.1). The precise measure of 'damage' to skin simulants is not defined in these regulations. When defining wound severity, clinicians tend to use one of two methods: a classification system (qualitative assessment of wound severity) or measuring wound size (quantitative assessment of wound severity).

Classification system

Skin tear classification systems have originally been developed to classify skin tears that occur in older, fragile people as a result of friction and / or shearing forces. The Payne-Martin skin tear qualification system was the first system developed to classify skin tears (Payne & Martin 1990). Three years later, a critique of this system with suggested revisions was published (Payne & Martin 1993). More recently, the Skin Tear Audit Research (STAR) classification system was developed which aimed to address problems of utilising the Payne-Martin classification system in a clinical setting (Carville *et al.* 2007). The STAR classification system is shown in Figure 2.13. This system currently remains the most up-to-date classification system for skin tears.

Figure 2.13: The five wounds categories identified by the Skin Tear Audit Research (adapted from Carville *et al.* 2007).

Further wound classification systems include the skin damage area and severity index (van den Eijnde *et al.* 2014a), a five-category system which can be used to score abrasive skin damage caused by skin-surface contacts. This system was able to quantify skin lesion injuries as well as providing good correlation with perceived discomfort of players.

Using classification systems for determining wound severity is a low-cost, simple solution especially when a test method requires a pass-or-fail criterion. Classification systems should be developed to provide a useful outcome measure, i.e. correlate with perceived wound severity. The classification systems discussed were developed for different skin injury types than is expected to result from stud-skin interactions. Despite the stated advantages of classification systems, their reliability is dependent on human interpretation of the wound properties and should be assessed before use.

Measuring wound size

Wound size can be quantified with one-dimensional (length, width), two-dimensional (surface area) and three-dimensional (3D, volume) measurements. In clinical practice, one-dimensional measurements taken with a ruler (Figure 2.14) are commonly used to quantify wound size and follow the healing progress (Goldman & Salcido 2002). Generally, wound length is taken as the longest wound distance, and its width is the greatest distance perpendicular to the wound length. If measured, depth is the deepest point of the wound bed. This measure is obtained by inserting a cotton-tip applicator into the deepest part of the wound bed, gripping the applicator by the wound margin and placing it against a ruler.

Figure 2.14: Ruler based measurements are commonly used to define wound size in clinical settings. Photo adapted from the Wound Care Education Institute.

Surface area measurements are either obtained by estimates from the one-dimensional measurements, by photography or by digital planimetry (Bilgin & Güneş 2014; Whittle *et al.* 2008). Estimating wound surface area by multiplying wound length and width is common practice among clinicians; though this method provides poor agreement for irregular shaped wounds with an overestimation of wound surface area (Goldman & Salcido 2002). An example of irregular shaped wounds can be found in Figure 2.15.

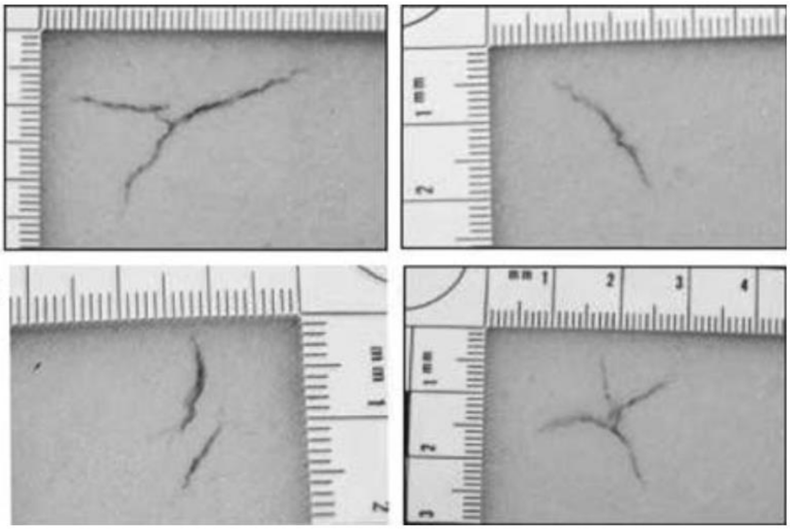


Figure 2.15: Example of four different shapes of blunt force impact wounds analysed using photography (adapted from Whittle *et al.* 2008).

Wound volume can be estimated from one-dimensional measurements (length x width x depth), reproduced with a moulding material or directly measured with a 3D scanner (Bills *et al.* 2016; Bilgin & Güneş 2014). Using one-dimensional measurements to calculate wound volume provides poor agreement for irregular shaped and shallow wounds (Goldman & Salcido 2002). 3D scanners can be used to re-create the wound bed without direct skin contact, therefore minimising the risk of infection for the patient. The accuracy of such devices for volume measures is higher than other non-invasive methods (Bills *et al.* 2016), though this has only been tested on ulcer-type wounds which are defined by a relatively wide, open wound surface compared to lacerations.

2.5.4 Summary

Human skin is an important organ for the body, providing protection and allowing for interaction with the outside world. Our skin generally consists of three layers: epidermis, dermis and hypodermis. A laceration occurs when at least the epidermal and dermal layers of the skin break, which can cause a blood vessel rupture. This usually occurs when tensile and / or shear strain is exerted on the skin. The mechanical properties of the skin are the result of both collagen and elastin fibres, and their mechanical response can vary widely between individuals and body locations. The skin can be damaged in a number of ways and whilst small damages can be easily repaired by the body, more severe damage can result in loss of the protective function and loss of sensory capabilities. Mammalian skin resists tearing by re-aligning collagen fibrils along the axis of the force, providing the skin with a high tear resistance. Quantifying skin damage in a clinical setting is commonly done by classification systems or by taking unidimensional measures such as wound depth, length and width. More complex measures such as wound surface area and volume can provide a useful metric of wound severity but are more time-consuming and their reliability is often dependent on the wound shape.

2.6 Skin simulants

Sports injuries happen at the point of human tissue failure, whether acute or due to overuse. Research into human tissue failure often uses surrogate materials for human tissue. Previous review papers on the use of human surrogates for injury biomechanics research showed that these surrogates range widely in biofidelity, repeatability, sensitivity, availability, application type, ethical restrictions and costs (Crandall *et al.* 2011; Payne *et al.* 2013). Although these studies have not focussed on skin injuries in particular, they do provide a detailed explanation of the advantages and limitations of a wide range of surrogates. Repeatability here refers to the ability of the surrogate to give the same response when the impact has not changed, whilst sensitivity refers to the ability to give different results when the injury-producing impact has changed.

Human volunteers, post-mortem human subjects (PMHS) and animals have traditionally been used as human surrogates in injury biomechanics research. Newer developments, such as anthropometric test devices (ATDs), computational models and bespoke, partial-body synthetic surrogates reduce the repeatability issues and ethical restrictions associated with biological tissue (Crandall *et al.* 2011; Payne *et al.* 2013). However, biological tissue is still used as a point of reference, or 'gold standard', throughout injury biomechanics. An overview of common human surrogates with their associated advantages and limitations is given in Table 2.3.

The complexity of human biological tissue assist its many functions, but it also means that finding a suitable simulant material for skin tissue is not straightforward. Various research fields, such as ballistics, injury biomechanics and sports performance assessments, have aimed to identify appropriate simulant materials for human skin. These physical skin models often replicate a single property of the human skin whilst being a strong simplification of real human skin (Dąbrowska *et al.* 2015). This section will give a brief overview of skin simulants used in previous research.

Table 2.3: Human surrogates in biomechanics research and their advantages and challenges for assessing laceration injury risk in this project. Summarised from Crandall *et al.* (2011) and Payne *et al.* (2013).

Simulant	**Advantages**	**Limitations**
Animal tissue	- Complex representation of human tissue - Easily available	- Ex-vivo or ethical restrictions - Differences in specific mechanical properties - Tissue not reusable, high inter and intra variability
Anthropometric test devices (ATDs)	- Off the shelf solution - Durable - Repeatable	- Not developed for skin tissue biofidelity - High initial cost - Expensive calibration procedures
Computational models	- Low physical costs - Easy to isolate and change impact parameters - High repeatability	- Modelling bio-materials is challenging; validity of current models is not high enough - Developed computer models still need to be validated with physical experiments
Human volunteers	- Not a surrogate; an exact representation of human tissue - Accurate representation	- Ethics limit testing to pain threshold - High variability between /within subjects - Repeatability is low because damaging impacts cannot be repeated on the same tissue
Part-body simulation by silicone materials	- Bespoke material properties can match human tissue response - Durable or frangible - Repeatable	- Validation necessary compared to biological tissue response - Specific to type of impact, generalisability still unknown - Currently only available for soft tissue, not skin
Post-mortem human subjects (PMHS)	- Exact representation of geometrical and anatomical structures - Breaking loads of tissue structures can be quantified	- Non-representative, elderly population - High costs of storage and disposal of human tissue - Tissue not reusable, high inter and intra variability - Loss of muscle tone post-mortem - Rigor mortis increases tissue stiffness

2.6.1 Biological surrogates

Human volunteers, PMHS, and animal studies all provide the researcher with complex and realistic biological responses during injury mechanics research. Biological surrogates show low repeatability between and within subjects. The test procedures further need to confirm to strict ethical guidelines for in-vivo surrogates, and the surrogate can suffer from tissue degradation when used ex-vivo (Bir *et al.* 2005; Payne *et al.* 2013).

Human volunteers, though technically not a simulant, are the most accurate representation of human tissue. In sports biomechanics research measuring human volunteers is preferred for injury mechanisms up until the pain threshold (Peppelman *et al.* 2013; Hrysomallis 2009). When investigating injury mechanisms above the pain threshold other simulants need to be considered.

PMHS, also known as cadavers, possess all the structures that also can be found in in-vivo human subjects. PMHS are therefore still an accurate representation of the living human body. Nevertheless, there are several issues associated with using PMHS; the availability of cadavers is low, average age of available cadavers is higher than expected from an athletic population, the loss of tonicity of muscle tissue post mortem and its associated changes in soft tissue response, rigor mortis, and logistical difficulties obtaining fresh cadavers. PMHS are therefore not a suitable simulant for testing protocols in standards or regulations for sports equipment.

Animal tissue is used in- and ex-vivo for injury biomechanics research (Crandall *et al.* 2011). Test protocols using anaesthetized animals for in-vivo injury mechanics research require similar, though less strict, ethical approval to human participants. The use of ex-vivo animal tissue has less ethical restrictions but the tissue has many of the disadvantages that can be found in PMHS; differences in stiffness of tissues post-mortem, variability of biological tissues and logistical issues of obtaining recently, i.e. ideally within 4 hours (Jussila *et al.* 2005), deceased animals. Pig skin has frequently been used as a human skin simulant because of its similar mechanical properties to human skin (Ankersen *et al.* 1999; Shergold *et al.* 2006; Parmar *et al.* 2012; Myouse *et al.* 2013; Dąbrowska *et al.* 2015). Generally, pig skin is thicker and tougher due to a higher collagen fibre content and it has been found to have a greater penetration force than human abdominal skin (Shergold & Fleck 2005; Lim *et al.* 2011).

2.6.2 Synthetic skin simulants

Synthetic materials are a more desirable option to replicate human skin for repeated mechanical testing (Carr & Wainwright 2011). Synthetic materials provide a low-cost, highly reproducible solution without the issues associated with biological materials such as ethical restrictions, tissue degradation, repeatability, temperature dependency.

Despite the obvious advantages of synthetic skin simulant materials, the properties of in-vivo human skin have recently been described as too complex and variable to simulate and standardise (Falland-Cheung *et al.* 2015; Dąbrowska *et al.* 2015). Dąbrowska *et al.* (2015) summarises current literature on physical skin models that have been developed across seven research fields. All models are specifically dependent on the skin properties that need to be simulated, and each model shown in Figure 2.16 is a significant simplification of human skin. The models aim to validly reproduce the properties for which they are developed, e.g. thermal properties or mechanical properties of human skin, without one model accurately reproducing them all.

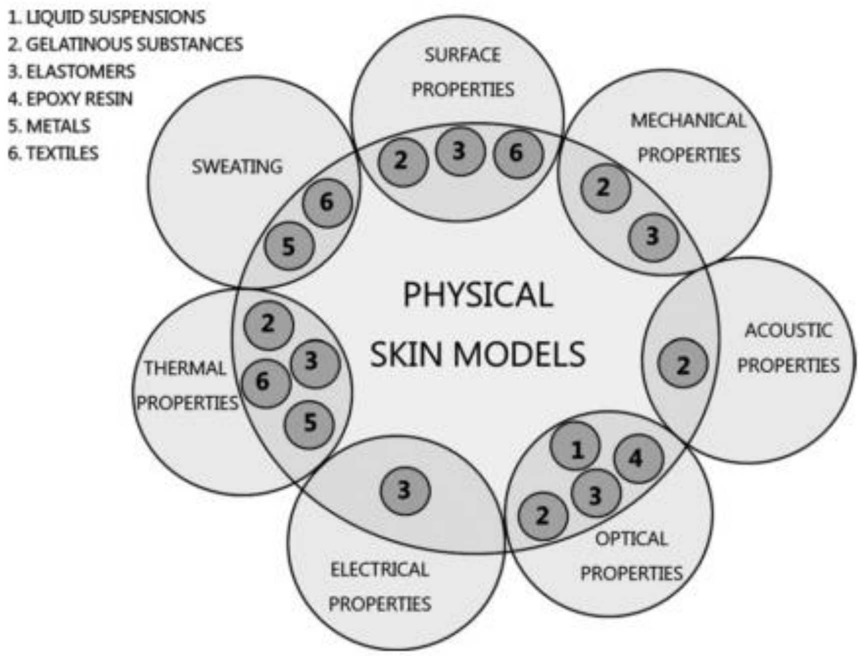


Figure 2.16: Main materials used to simulate different categories of skin properties according (adapted from Dąbrowska *et al.* 2015).

A wide variety of synthetic skin simulants have also been developed for the purpose of skin grafting, i.e. transplantation of skin (Shevchenko *et al.* 2010). These tissue-engineered skin substitutes aim to offer off-the-shelf solutions for patients with large skin loss, providing a barrier against infection and pain relief. Rejection of the skin graft is common among synthetic tissue-engineered skin substitutes, which leads to them being a temporary solution until sufficient material from a donor site can be grown. These materials have not been developed to mimic mechanical properties of the skin, such as tear resistance and breaking load.

Synthetic skin simulants have further been used in the area of skin tribology to increase the repeatability of friction testing. Selecting skin simulant materials with realistic friction behaviour can increase the biofidelity of stud laceration test methods. A variety of silicone materials are used as suggested skin simulants for tribology research, depending on the age, moisture content and type of skin that needs to be simulated (Chimata & Schwartz 2015; Tay *et al.* 2016; Tay *et al.* 2017). Further, artificial leather (Lorica ® Soft) has been shown to have a close correspondence of friction behaviour compared to human skin (Derler *et al.* 2007). This material has since been used as a skin simulant for skin tribology research (Cottenden & Cottenden 2013; Falloon & Cottenden 2016) though it is no longer available for purchase at the time of writing.

When selecting a synthetic skin simulant for a newly developed stud laceration test, emphasis should be given to the ability of the skin simulant to replicate the mechanical failure threshold of human skin. An unstructured literature search with combinations of the following search terms: *skin, simulants, models, synthetic, phantom, tissue, surrogate, forensic, ballistic* was conducted on a regular basis between January 2015 and May 2017. A variety of synthetic skin simulants were identified. In forensic studies, synthetic chamois (Ankersen *et al.* 1999; Bir *et al.* 2012), natural rubber (Shergold & Fleck 2004; Jussila *et al.* 2005) and silicone rubber (Shergold *et al.* 2006; Parmar *et al.* 2012; Whittle *et al.* 2008) were identified as realistic synthetic skin simulants, previously shown to reach similar mechanical failure threshold compared to mammalian skin. In sports specific injury studies, soft tissue simulants have been identified or developed and compared to human tissue response (Hrysomallis 2009; Payne *et al.* 2013), but no skin simulants replicating tissue failure under mechanical loading from sports impacts were found. Commercially developed skin simulant materials, such as the silicone based biofidelic skin simulant developed by the university of Alabama (Chanda *et al.* 2017) and SynTissue developed by Syndaver labs (Tampa, Florida, USA) can provide ready-made solutions to researchers. The extent to which these simulants replicate the mechanical failure threshold of human skin is still unknown.

2.6.3 Soft tissue simulants

Previous studies have suggested that the breaking load of skin is dependent on the hardness of its underlying tissue. Bir *et al.* (2012) found that the difference in penetration threshold velocity of skin on bony parts and skin on soft tissue was 24.0 m/s and 33.3 m/s, respectively. Skin damage of skin pinned to a solid background during blunt force impact testing is primarily based on crushing (Jussila *et al.* 2005). The indentation response of the soft tissue layer can thus influence laceration injury risk of the skin.

Gelatine is frequently used as a soft tissue simulant in ballistic studies, especially for analysing penetration threshold or penetration depth of munition (Shepherd *et al.* 2009; Appleby-Thomas *et al.* 2011; Bir *et al.* 2012). The weight percentage of gelatine powder to water is recommended to be between 10% (Jussila 2004) and 25% (Ragsdale & Josselson 1988; Shepherd *et al.* 2009). The greater the weight percentage, the firmer the gelatine solution will be. This hardness is also dependent on the temperature of the final solution, where warm gelatine is generally softer than cold gelatine.

Silicone rubbers have been previously used and validated as durable human soft tissue simulants specifically for sports impact testing (Hrysomallis 2009; Payne *et al.* 2014; Payne *et al.* 2014). In-vivo soft tissue properties have been investigated using a small drop hammer on relaxed thigh muscle of human volunteers until the pain threshold (Hrysomallis 2009). The deceleration response of the drop hammer was measured during this test. The drop test was repeated on a range of silicone materials and the deceleration response of the silicone was compared to the muscle tissue. Two silicone materials performed well; Silastic 3483, which differed 1.5% to the mean relaxed human tissue deceleration response, and Silastic 3481, which differed 12% from the mean relaxed human tissue deceleration response. Both materials performed within the 95% confidence interval (CI) boundary of the human tissue response.

Silastic 3483 is an off-the-shelf solution, where a single material is used to mimic the response of the complex, multi-material structure that is human tissue. Recently, a bespoke silicone blend was developed to mimic the mechanical properties of human relaxed thigh muscle (Payne *et al.* 2015a). The response properties of this developed silicone thigh were based on knee-thigh impacts as commonly seen in basketball lay-ups (Halkon *et al.* 2014). This type of multi-material silicone blends can improve the biofidelity of the material response (Payne *et al.* 2015b), though they are less cost- and time effective then off-the shelf solutions. As previously mentioned, off-the shelf solutions are at risk of future discontinuation of the material. These benefits and limitations of bespoke and readily available solutions need to be considered when choosing simulant materials.

2.6.4 Computational simulation

Computational modelling of human skin allows a large number of experiments to be performed, where the effect of individual parameters can easily be studied. These models have the added advantage of lower running costs compared to physical testing and they have limited ethical restrictions (Payne *et al.* 2013; Krosshaug *et al.* 2005). Developing accurate computational models of human tissue can reduce or replace animal models (Groves *et al.* 2013).

Skin is an anisotropic, viscoelastic substance with great inter and intra variability. Modelling in-vivo human skin is therefore a significant challenge, and computational models often simplify the skin to a single-layer, homogenous material (Delalleau *et al.* 2008). Computational models require input information from physical testing; if this information is obtained from in-vivo human participants, breaking loads cannot ethically be investigated (Bischoff *et al.* 2000; Delalleau *et al.* 2008; Flynn *et al.* 2011). If skin samples are tested ex-vivo these ethical restrictions are lifted, though it has long been known that the mechanical response of mammalian skin post-mortem changes rapidly (Marangoni *et al.* 1966).

The development of numerical models for injury biomechanics research has shown a clear progression over the past 50 years according to the review paper by Yang *et al.* (2006), with improved understanding of human impact response and injury mechanisms. However, developing well-validated human models still require more material property data, especially for rate dependent tissue - such as skin - at high strain rates (Payne *et al.* 2013; Yang *et al.* 2006). Modelling the breaking point of in-vivo skin thus remains difficult, since it is not possible to obtain experimental data to inform models from controlled human participant experiments.

2.6.5 Summary

Previous biomechanical research has used PMHS, animals, ATDs, computer simulations and synthetic materials such as silicone rubbers to simulate human tissue response at impact. The use of PMHS is ethically restricted and their usability is limited because of low repeatability between and within subjects, post-mortem tissue degradation, and unrepresentative populations. Computational models are currently not well enough developed to replicate the anisotropic and viscoelastic response of human skin, especially at tissue failure loads. The use of synthetic materials to simulate human skin and soft tissue is a constantly developing area. Skin is a highly complex tissue, and synthetic skin models often aim to replicate one or only a few of its properties. The failure threshold under mechanical loading is important for skin simulants in sports injury research. Forensic studies have previously used pig skin, synthetic chamois, natural rubber and silicone rubbers as skin simulant materials when investigating mechanical failure of the skin. The breaking load of skin is found to be dependent on the underlying soft tissue. Selecting an appropriate soft tissue simulant which has a similar indentation response to human soft tissue is therefore important. Bespoke, multi material silicone models have been developed for superior soft tissue response, though off-the-shelf solutions can already provide a deceleration response within human variance.

2.7 Literature review summary

Rugby union is a sport with one of the highest injury risks in the UK. It has an estimated 81 injuries per 1000 exposure hours in the professional game. Further information is required as to what proportion of these injuries are skin lacerations, whether or not they were caused by studded footwear, and what game scenario is related to this type of injury. Previous studies on studded footwear research have predominantly focussed on obtaining optimal traction parameters, since optimising traction can enhance performance. Outsole design can pose an increased injury risk to the wearer, either by localised foot pressures resulting in metatarsal fractures, or through excessive rotational traction, which has been related to knee- and ankle ligament injuries. Laceration injuries caused by studded footwear have not previously been investigated, though links between outsole design and laceration injury risk have been suggested.

Current studded footwear regulations in rugby union, published in Regulation 12, describe an optional test method that can be used to assess the laceration injury risk of stud designs. The test settings in Regulation 12 have not been informed by biomechanical parameters of stud laceration injuries in rugby union. The test method is also under defined; no drop height or mass are given for one of the suggested tests, and the skin and soft tissue simulant are not prescribed. Rugby league and association football do not have any test methods in place to assess the laceration injury risk of studded footwear. Football shin guards need to comply with BS EN: 13061:2009 which includes a stud-shin guard impact. In rugby union, rugby league and association football, it is up to the discretion of the referee to decide whether or not a player is allowed to wear certain stud or outsole design.

Measuring kinetic and kinematic parameters of injury events in sport is challenging due to the fast-paced, high impact environment these injuries often occur in. Kinematic data can be obtained from a variety of measurement systems, such as motion capture systems, accelerometers and high-speed cameras. Kinetic data can be measured using force plates, load cells or pressure sensors. Pressure sensors give a spatio-temporal profile of the impact, allowing for analysis of individual stud impacts forces. Piezo-resistive pressure sensors have been identified as a lightweight, flexible and safe solution to measure force. Sensor calibration methods need to be reviewed in order to obtain the most accurate results.

Human skin is a complex, anisotropic and viscoelastic tissue. Skin is often divided into three layers: epidermis, dermis and hypodermis. The mechanical properties of mammalian skin are determined by its collagen and elastin fibres, and the resultant skin structure has a high resistance against tearing. Clinical measures of wound size range from classification systems to uni-, bi- and three-dimensional measurements.

Simulating human skin is necessary when investigating injury thresholds. A wide range of human skin simulants have been used across research fields and the most appropriate simulant type depends on the properties that need to be simulated and the application. Biological surrogates can provide a similar complex-tissue response to in-vivo human skin; though they are often highly variable, ethically restricted and their mechanical properties quickly change post-mortem. Synthetic simulants are a simplification of the complex human tissue, and they should be selected based specifically on the properties that need to be simulated. The underlying soft tissue influences the failure threshold of skin and therefore should be included in a skin model. Computational modelling can provide a cost-effective, repeatable test method. However, current models have not been able to accurately replicate the failure mechanisms of human skin and need validating by real-life experiments.

3 Laceration injury prevalence

3.1 Introduction

Identifying the proportion and prevalence of skin injuries in rugby union is the first step of the injury prevention sequence by van Mechelen (1997) (Figure 1.2). This step establishes a baseline measure of the injury frequency, against which the effectiveness of an injury prevention strategy can be measured. Epidemiological studies in sport establish injury prevalence by observing the number and type of injuries in athletes, offset against the hours the sport is played (exposure hours). An earlier systematic review of published epidemiological studies in professional rugby union identified that overall match injury prevalence was 81 injuries per 1000 exposure hours (ITEH) and training injury prevalence was 3 ITEH (Williams *et al.* 2013). In the review by Williams *et al.*, lacerations and skin specific injuries accounted for 1 - 3 ITEH (95% CI) during match play. No prevalence statistics were reported for laceration and skin injuries during training sessions.

The introduction of professionalism in rugby union in 1995 has coincided with a higher injury prevalence among players at the highest level of the game (Garraway *et al.* 2000). Williams *et al.* (2013) also found that players at an international level were at higher risk of injury during matches than players at the national level, though this effect was not found for training injuries. Since playing level seems to influence injury risk in rugby union, it is still unclear what the prevalence of skin and laceration injuries is among amateur players. Potential influences of skin and laceration injury prevalence between playing levels include; different strategies used between professional and amateur rugby, the quality and potential wear of the footwear worn, and the higher forces associated with professional play. Identifying potential differences in skin injuries between amateur and professional rugby will help identify at which level of the game players are at an increased risk. Professional rugby players have traditionally been monitored on injury risk more frequently, so injury trends can clearly be established (Rugby Football Union 2017). Amateur injury trends are thus less likely to be identified at an early stage.

This chapter investigates the prevalence and proportion of laceration injuries in amateur and professional rugby, during both training sessions and matches by a systematic review and meta-analysis of published epidemiological studies. The results of this study can be used as a baseline assessment of skin and laceration injury prevalence after a potential intervention, such as a change in studded footwear regulations.

3.2 Aim and objectives

The aim of this chapter is to systematically review published epidemiological literature reporting on laceration or skin injury prevalence in rugby union during match play and training sessions and to identify potential differences in skin injury risk for amateur and professional players. This relates to stage B (*'Establish the prevalence of injury'*) of the project structure described in Figure 1.3.

The aim is achieved through the following objectives:

1. Identify the injury prevalence of skin and laceration injuries during training sessions and matches in published epidemiological studies.
2. Identify the proportion of skin and laceration injuries during training sessions and matches in published epidemiological studies.
3. Investigate if there is a difference in skin and laceration injury risk for amateur compared to professional rugby union players.

3.3 Injury definitions in epidemiological studies

The injury definition used in sports injury epidemiological studies influences their results and has remained a topic of debate (Ekstrand & Karlsson 2003; Hägglund *et al.* 2005; Fuller, Molloy, *et al.* 2007). Four commonly used injury definitions are:

- **Time-loss (one-day):** Any physical complaint sustained by a player during a match or training session that prevents the player from taking full part in a match or training session one day after the injury (Fuller, Molloy, *et al.* 2007).

- **Time-loss (immediate):** Any physical complaint sustained by a player during a match or training session that forces the player to retire and / or prevents the player from taking full part in a training session or match the following day.
- **Medical attention:** The player receives medical treatment for his / her injury.
- **Player reported:** The player reports his / her physical complaint as an injury.
- **Trainer reported:** The coach or (head) trainer reports a player injured.

A consensus statement on injury definitions and data collection procedures for epidemiological studies in rugby union was published in 2007 (Fuller, Molloy, *et al.* 2007). This document recommends using the time-loss, one-day injury definition. The consensus statement also recommends for laceration injuries to have their own injury type category. If a time-loss injury definition is used, minor injuries and injuries which fall under the blood injury rule (World Rugby 2016) could be treated on the side of the field (stitched or glued). This would subsequently be ignored in the injury count, leading to a likely underestimation of the skin injury prevalence (van den Eijnde *et al.* 2014b; Gibbs 1993). An example of such a situation is given in the consensus statement (Fuller, Molloy, *et al.* 2007), p330:

"A loose-head prop forward sustained a laceration to his head during a match; the player left the field of play to enable the team doctor to suture and protect the injury. The player returned to the field of play. The player continued to train and play with his head bandaged for the next three weeks. This episode should not be recorded as an injury"

In a systematic review on skin injuries in football, van den Eijnde *et al.* (2014b) warned that skin injuries were underreported in current literature and suggested this resulted from restrictive injury definitions. In this same study, the use of a medical attention injury definition rather than a time-loss injury definition led to a two to fourfold increase in the injury prevalence of abrasions and lacerations compared to the use of a time-loss injury definition.

3.4 Methods

This review was registered in the PROSPERO International Prospective Register of Systematic Reviews (registration number CRD42015024027). The 2009 PRISMA-P guidelines for preferred reporting items for systematic review and meta-analysis protocols (Moher *et al.* 2009) were implemented in preparing and conducting this review.

3.4.1 Literature search strategy

The literature search was conducted in July 2016 using the search engines Pubmed, Ovid, Scopus and Web of Science. The search strategy was designed with the purpose of finding epidemiological studies on rugby union injuries. Search terms for all search engines were:

[rugby] AND [epidemiology OR epidemiological OR epidemiologic OR injury OR injured OR injuries].

Search engine limits were set to find only articles published after January 1995, due to an innovative change in stud design in the previous year (1994) and the introduction of professional rugby union in 1995. The title, author, date of publication and abstract of each record were imported to a reference manager (RefWorks, ProQuest® LLC, 2016).

3.4.2 Selection criteria

After removing duplicates, the title and abstract of the remaining records were screened for eligibility. From resulting records the full-text article was obtained and again screened for eligibility. The following inclusion criteria were used throughout:

1. The study must report on prospective, epidemiological findings in a rugby union cohort of players.
2. The study must observe injuries in players over 18 years.
3. The study must not report only a sub-category of injuries, e.g. tackling injuries or head injuries.
4. The paper must report a skin or laceration injury category and provide enough information to calculate injury prevalence within this category.

5. The athlete exposure in the study should be over 400 match hours and / or 900 training hours.
6. The full-text article must be written in English and published in a peer-reviewed journal after 1995.

Only studies observing injuries in an adult cohort were included in this review since in junior (under 18 years) age categories, the rules of the game are slightly different. The athlete exposure must be of a size that at least one team (15 players) was followed for a full season (defined as 20 matches and / or 20 weeks of 2 training sessions a week), which results in a minimum of 400 match exposure hours or 900 training exposure hours. Only studies published after 1995 were included. No restriction was placed on level of play (amateur or professional), sex, or injury definition used. Athletes were defined as 'professional' if they were receiving monetary reimbursement for their participation in the team to the degree that maintaining full-time employment elsewhere was not necessary. Review papers, conference abstracts and studies using data which was already published in a different study were excluded. A study was also excluded if it did not report injury numbers and exposure hours for match and training sessions separately.

3.4.3 Quality and bias assessment

A four-item checklist was used to assess the risk of bias of each of the included studies. The checklist was adapted from Waldén *et al.* (2015) and based on the "Strengthening the reporting of observational studies in epidemiology" (STROBE) statement (von Elm *et al.* 2007). The checklist used four risk of bias items; item 1: unclear reporting on the amateur or elite level of players, unclear reporting of number of seasons; item 2: large dropout (>25%) during study, unclear or biased selection of teams; item 3: approximate or unclear registration and calculation of exposure hours; item 4: unclear (skin) injury definition, retrospective reporting of injuries (e.g. telephone interviews). Studies were assessed on each item for low risk (0) or high risk (1) of bias. A cumulative score for each study was calculated. Studies with a cumulative score of 0 were defined as at low risk of bias, studies with a score of 1 or 2 as at medium risk and studies with a score >3 as at high risk. Only studies with a low or medium risk of bias were included in the meta-analysis.

3.4.4 Data extraction

For each study that met the inclusion criteria, the injury definition, skin injury category and athlete level - professional or amateur - were recorded. Where available, the observed number of total injuries, the observed number of skin or laceration injuries, and their corresponding match and training exposure hours were extracted from each study. Injury prevalence was defined as the number of observed injuries per 1000 exposure hours. For each study, separate injury prevalence (± 95% CI) was calculated for injuries sustained during match play and during training sessions. The percentage of skin injuries in relation to total injuries observed in a study was also calculated for both matches and training sessions. If stated in a separate category, abrasion injuries were not included when calculating skin injury prevalence. A weighting factor based on study size (defined as exposure hours) was implemented when calculating the mean of each group of studies.

3.4.5 Statistical analysis

In this meta-analysis the influence of playing level (amateur or professional) on skin injury prevalence during match and training sessions was of interest. Influence of sex on injury risk could not be investigated due to a shortage of studies publishing female injury numbers. Skin injury prevalence data was modelled using a negative bionomial regression with log link. This generalised linear model is similar to a Poisson regression model (Lystad *et al.* 2009), but also accounts for overdispersion in the data set. In the used model, the response variable was the number of observed skin injuries, offset by the log of the exposure hours. A weight scaling variable was assigned to each study depending on its relative size. The odds ratio (and 95% CI) that an amateur player sustains a skin injury - as compared to a professional player - was calculated. Separate odds ratios were determined for training sessions and during match play. Alpha was set at 0.05.

3.5 Results

A flowchart of the search results and selection procedure is shown in Figure 3.1. A total of 1351 records were identified through the online search engines (Stage 1, Figure 3.1). After duplicates were removed, the titles and abstracts of 464 remaining records were screened (Stage 2). The screening excluded 424 articles; therefore 40 full-text articles were reviewed for eligibility (Stage 3). Twenty-eight articles were excluded based on the selection criteria, leaving twelve studies which were included in this review (Best *et al.* 2005; Bird *et al.* 1998; Brooks *et al.* 2005a; Brooks *et al.* 2005b; Brooks *et al.* 2005c; Fuller *et al.* 2008; Fuller *et al.* 2009; Fuller & Molloy 2011; Fuller *et al.* 2013; Jakoet & Noakes 1998; Kerr *et al.* 2008; Schwellnus *et al.* 2014) (Stage 4). Out of these twelve included studies, ten measured professional athletes (Best *et al.* 2005; Brooks *et al.* 2005a; Brooks *et al.* 2005b; Brooks *et al.* 2005c; Fuller *et al.* 2008; Fuller *et al.* 2009; Fuller & Molloy 2011; Fuller *et al.* 2013; Jakoet & Noakes 1998; Schwellnus *et al.* 2014) and two studies observed injuries in an amateur cohort (Bird *et al.* 1998; Kerr *et al.* 2008). Eleven out of twelve studies included a completely male cohort (Best *et al.* 2005; Brooks *et al.* 2005a; Brooks *et al.* 2005b; Brooks *et al.* 2005c; Fuller *et al.* 2008; Fuller *et al.* 2009; Fuller & Molloy 2011; Fuller *et al.* 2013; Jakoet & Noakes 1998; Kerr *et al.* 2008; Schwellnus *et al.* 2014); one study observed injuries of both males and female rugby players (Bird *et al.* 1998).

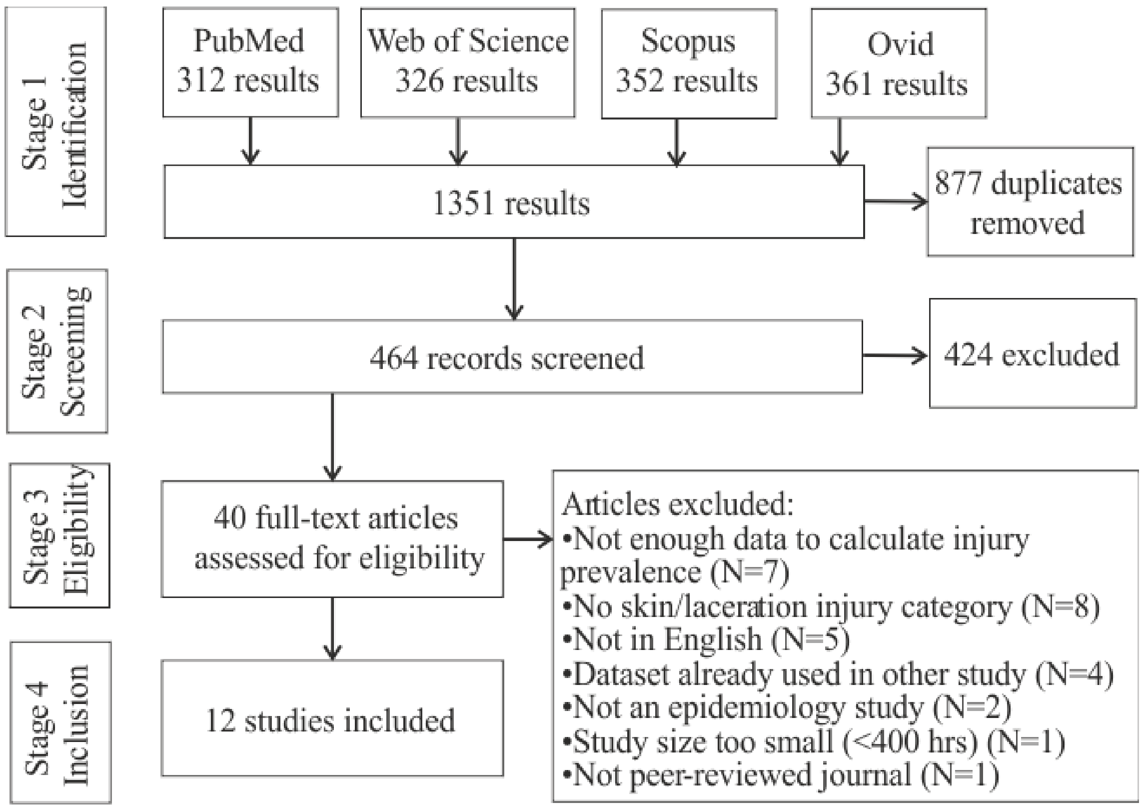


Figure 3.1: Flow chart of the literature search and selection process. Chart design adapted from Moher *et al.* (2009).

3.5.1 Quality and bias assessment

Results of the risk of bias assessment of each of the included studies can be found in Table 3.1.

Table 3.1: Risk of bias assessment of each of the 12 studies included in the meta-analysis. For each item, 0 indicates a low risk and 1 indicates a high risk. A cumulative score of 0 was defined as a low risk of bias, 1 or 2 as a medium risk and 3 or 4 as a high risk (von Elm et al. 2007).

Study	Item 1: Study setting and study period	Item 2: Eligibility criteria and player selection	Item 3: Exposure definition and measurement	Item 4: Injury definition, measurement and reporting	Cumulative number of items with bias
(Best *et al.* 2005)	0	0	0	0	0
(Bird *et al.* 1998)	0	0	1	1	2
(Brooks *et al.* 2005a)	0	0	0	0	0
(Brooks *et al.* 2005b)	0	0	0	0	0
(Brooks *et al.* 2005c)	0	0	0	0	0
(Fuller *et al.* 2008)	0	0	0	0	0
(Fuller *et al.* 2009)	0	0	0	1	1
(Fuller & Molloy 2011)	0	0	0	1	1
(Fuller *et al.* 2013)	0	0	0	0	0
(Jakoet & Noakes 1998)	0	0	0	1	1
(Kerr *et al.* 2008)	0	1	1	0	2
(Schwellnus *et al.* 2014)	0	0	0	0	0

3.5.2 Match injuries

Eleven out of twelve studies included in this review monitored and reported the match injuries of their athletes (Best *et al.* 2005; Bird *et al.* 1998; Brooks *et al.* 2005a; Brooks *et al.* 2005b; Fuller *et al.* 2008; Fuller *et al.* 2009; Fuller & Molloy 2011; Fuller *et al.* 2013; Jakoet & Noakes 1998; Kerr *et al.* 2008; Schwellnus *et al.* 2014). An overview of the match injury prevalence and proportion of included studies is given in Table 3.2. The majority of the studies (Brooks *et al.* 2005a; Brooks *et al.* 2005b; Brooks *et al.*

2005c; Fuller *et al.* 2008; Fuller *et al.* 2009; Fuller & Molloy 2011; Fuller *et al.* 2013; Kerr *et al.* 2008; Schwellnus *et al.* 2014) used a *time-loss one-day* injury definition. The other studies used *time-loss immediate*, sometimes in combination with *medical attention* (Best *et al.* 2005; Bird *et al.* 1998; Jakoet & Noakes 1998). No included study used a *player* or *coach reported* definition. Mean overall match injury prevalence in professional and amateur players combined was 53 ITEH (95% CI: 51 - 54).

Table 3.2: Total and skin injury prevalence of rugby union players during matches. Prevalence is reported as injuries per 1000 match hours. TL-1D = time-loss one-day; TL-IM = time-loss immediately; MA = medical attention.

Study	**Athlete level**	**Injury definition**	**Injury category**	**Match exposure hours**	**Match injury prevalence**	**Skin injury prevalence**	**No of skin injuries** [proportion]
(Best *et al.* 2005)	Professional	TL-IM	Open wound / Laceration	1930	97.9	21.2	41 [21.7%]
(Bird *et al.* 1998)	Amateur	TL-IM & MA	Laceration	5871	83.3	7.3	43 [8.8%]
(Brooks *et al.* 2005a)	Professional	TL-1D	Laceration & skin	445	218.0	4.5	2 [2.1%]
(Brooks *et al.* 2005b)	Professional	TL-1D	Laceration & skin	16782	91.4	1.1	18 [1.2%]
(Fuller *et al.* 2008)	Professional	TL-1D	Laceration	1920	83.9	0	0 [0%]
(Fuller *et al.* 2009)	Professional	TL-1D	Skin	5600	74.8	2.5	14 [3.3%]
(Fuller & Molloy 2011)	Professional	TL-1D	Skin	3320	52.4	1.5	5 [2.9%]
(Fuller *et al.* 2013)	Professional	TL-1D	Laceration	1920	89.1	1.6	3 [1.8%]
(Jakoet & Noakes 1998)	Professional	TL-IM	Laceration	2194	31.9	8.7	19 [27.0%]
(Kerr *et al.* 2008)	Amateur	TL-1D	Laceration	34324	17.0	0.8	29 [5.0%]
(Schwellnus et al. 2014)	Professional	TL-1D	Skin	1512	83.3	4.0	6 [4.8%]

A forest plot showing the mean and 95% CI of the skin and laceration injury prevalence of individual studies can be found in Figure 3.2. The skin injury prevalence during match play in amateur and professional players combined was 2.4 ITEH (95% CI: 2.0 -

2.7) (Figure 3.2). Mean skin injury prevalence during professional match play was 3.0 ITEH (95% CI 2.5 - 3.6) and during amateur match play 1.8 ITEH (95% CI 1.4 - 2.2). The skin injury risk for amateur players compared to professional players during matches was not significantly different (odds ratio: 0.60; 95% CI: 0.17 - 2.08; $p = 0.42$). The lowest skin injury prevalence observed was 0 laceration injuries after 1920 hours of match exposure (Fuller *et al.* 2008). The highest skin injury prevalence was 21.2 ITEH, where 41 injuries occurred during 1930 hours of match play (Best *et al.* 2005). The proportion of skin and laceration injuries in the included studies during match play ranged from 0 to 27%, with a weighted mean of 5.1%.

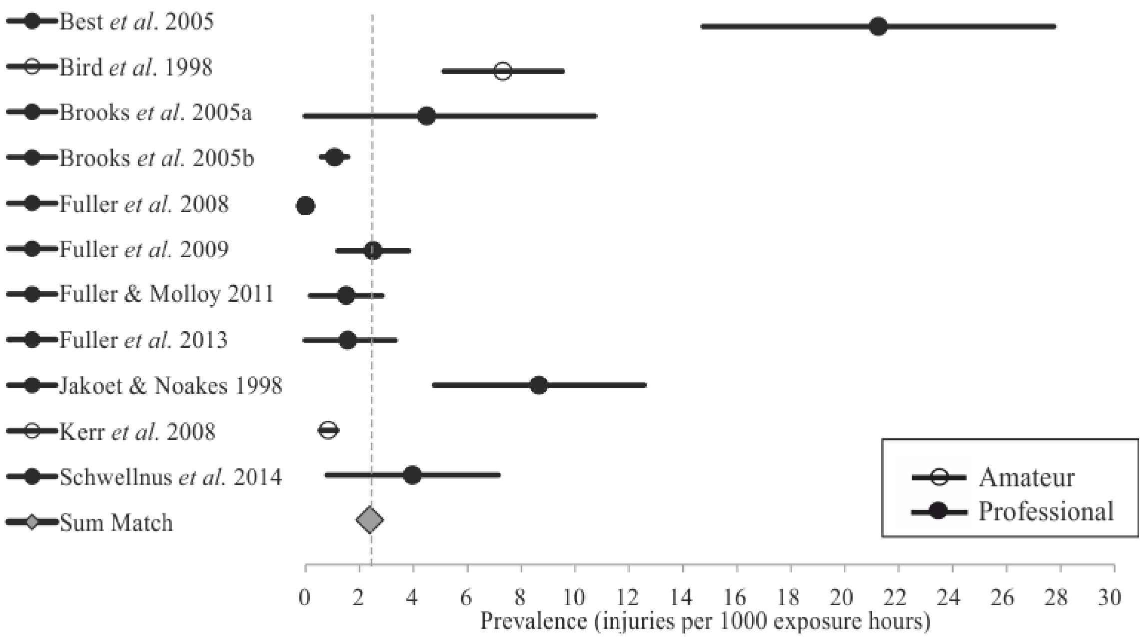


Figure 3.2: Forest plot of skin injury prevalence during match play in studies observing amateur and professional injuries.

3.5.3 Training injuries

Seven out of twelve studies included in this review (also) monitored and reported the injury data of their athletes during training sessions (Bird *et al.* 1998; Brooks *et al.* 2005a; Brooks *et al.* 2005c; Fuller *et al.* 2008; Fuller *et al.* 2013; Kerr *et al.* 2008; Schwellnus *et al.* 2014). An overview of the training injury prevalence and proportion is given in Table 3.3. Five studies monitored a professional population (Brooks *et al.* 2005a; Brooks *et al.* 2005c; Fuller *et al.* 2008; Fuller *et al.* 2013; Schwellnus *et al.* 2014), and two studies an amateur population (Bird *et al.* 1998; Kerr *et al.* 2008). One study used the *time-loss immediate* in combination with *medical attention* injury definition (Bird *et al.* 1998), whilst the other six studies used the *time-loss, one day* injury definition (Fuller, Brooks, *et al.* 2007; Brooks *et al.* 2005a; Fuller *et al.* 2008; Fuller *et al.* 2013; Kerr *et al.* 2008; Schwellnus *et al.* 2014). Mean overall training injury prevalence for amateur and professional players combined was 2.8 ITEH (95% CI: 2.6 - 2.9).

Table 3.3: Total and skin injury prevalence of rugby union players during training sessions. Prevalence is reported as injuries per 1000 training hours. TL-1D = time-loss, one day; TL-IM = time-loss immediately; MA = medical attention.

Study	**Athlete level**	**Injury definition**	**Injury category**	**Training exposure hours**	**Training injury prevalence**	**Skin injury prevalence**	**No of skin injuries** [proportion]
(Bird *et al.* 1998)	Amateur	TL-IM & MA	Laceration	12980	8.7	0.23	3 [2.7%]
(Brooks *et al.* 2005a)	Professional	TL-1D	Laceration & skin	7928	6.1	0	0 [0%]
(Brooks *et al.* 2005c)	Professional	TL-1D	Laceration & skin	196409	2.0	0.02	4 [1.0%]
(Fuller *et al.* 2008)	Professional	TL-1D	Laceration	17046	3.5	0	0 [0%]
(Fuller *et al.* 2013)	Professional	TL-1D	Laceration	15628	2.2	0	0 [0%]
(Kerr *et al.* 2008)	Amateur	TL-1D	Laceration	72039	3.7	0.14	10 [3.8%]
(Schwellnus *et al.* 2014)	Professional	TL-1D	Skin	15828	2.1	0	0 [0%]

A forest plot showing the mean and 95% CI of the skin and laceration injury prevalence of individual studies can be found in Figure 3.3. Mean skin injury prevalence during training sessions of amateur and professional players combined was 0.05 ITEH (95% CI: 0.02 - 0.07) (Figure 3.3). Skin injury prevalence during training sessions in professional players was 0.02 ITEH (95% CI: 0.00 - 0.03) and in amateur players 0.15 ITEH (95% CI: 0.07 - 0.24). There was a significantly higher risk of skin injuries for amateur players compared to professional players during training sessions (odds ratio: 7.92; 95% CI: 1.19 - 52.78; p = 0.03). Four out of seven studies (Brooks *et al.* 2005a; Fuller *et al.* 2008; Fuller *et al.* 2013; Schwellnus *et al.* 2014) reported that no laceration injuries were observed during training sessions (Figure 3.3). The proportion of skin and laceration injuries during training sessions ranged from 0 - 4.5%, with a weighted mean of 1.5%.

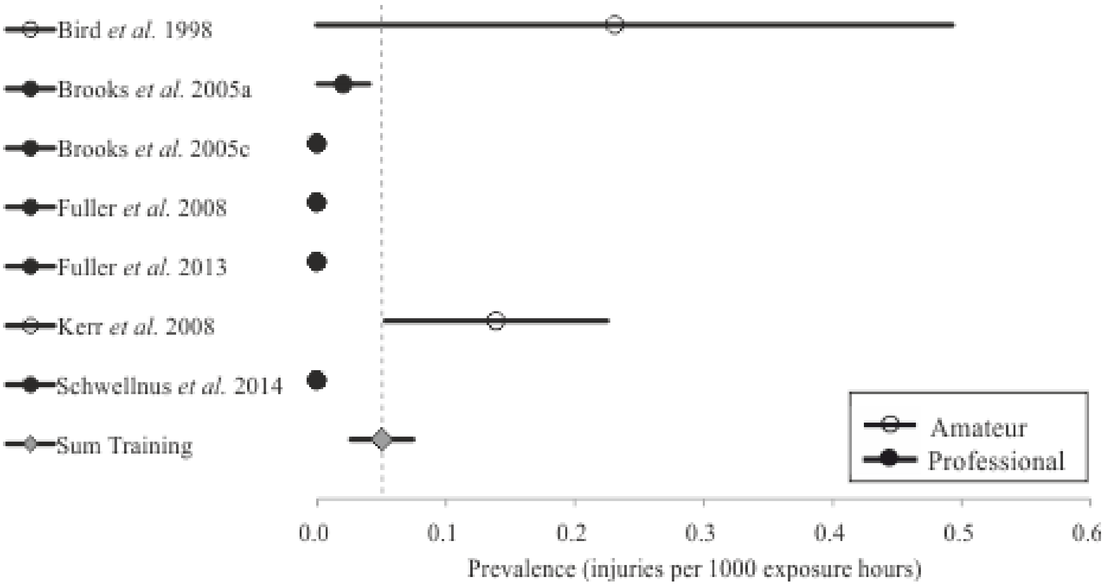


Figure 3.3: Forest plot of skin injury prevalence during training sessions in studies observing amateur and professional injuries.

3.6 Discussion

The objective of this review was to identify the prevalence of skin injuries in rugby union during match play and training sessions and to investigate if the risk of injury differs between amateur and professional players. The skin injury prevalence during match play from the combined data of eleven studies observing rugby union players was 2.4 ITEH, accounting for 5.1% of all match injuries. This number could be interpreted as one time-loss or medical attention laceration injury per team (15 players) per year (20 matches, 80 minutes per match). Seven of the included studies in this review provided sufficient information to calculate skin and laceration injury prevalence during training sessions. Mean skin injury prevalence during training sessions was 0.05 ITEH, accounting for 1.5% of all training injuries. Injury prevalence has previously been found to be lower during training sessions than in match play (Fuller *et al.* 2013; Brooks *et al.* 2005b; Brooks *et al.* 2005c), and in this review a similar trend is observed when measuring skin injury prevalence (training: 0.05 ITEH; matches: 2.4 ITEH). In comparison, the incidence of abrasion and laceration injuries in association football matches - systematically summarised by van den Eijnde *et al.* (2014b) - varied from 0.8 to 6.1 ITEH.

Ten out of twelve studies included in this review reported on professional rugby union injuries. No difference was found in skin injury risk during match play for amateur players compared to professional players (odds ratio: 0.60; 95% CI 0.17 - 2.08; p = 0.42). During training sessions, the risk of skin injuries was higher for amateur players than for professional players (odds ratio: 7.9; 95% CI: 1.2 - 52.8; p = 0.03). Only two studies included in this review reported on amateur injuries (Bird *et al.* 1998; Kerr *et al.* 2008), from which one used a time-loss immediate in combination with a medical attention injury definition (Bird *et al.* 1998). The injury definition could therefore have influenced this result. Williams *et al.* (2013) found no clear influence of playing level on overall injury risk during training sessions, though more elite levels of play were associated with a higher injury risk during matches. The conclusions of Williams *et al.* (2013) were based on various levels of professional rugby (club level 2 to international), which makes it unclear if this trend could be extrapolated to injuries sustained in amateur rugby as well. The impact of professionalism on injuries has previously been investigated by Garraway *et al.* (2000) who concluded that the introduction of professionalism in rugby union had increased the prevalence of injuries in both amateur and professional cohorts. Financial rewards were suggested to raise the overall injury rate and increase the pressure on athletes to return to play as soon as possible, making recurrent injuries more likely. The lower risk of skin injuries found in professional players could indicate that this type of injury is frequently ignored in professional rugby injury counts, therefore underestimating the extent of the injury problem (van den Eijnde *et al.* 2014b; Gibbs 1993). Approximately one-third of amateur rugby union players have found their own studs sharpened due to wear (Section 4.4.4, Oudshoorn *et al.* 2016a). Professional rugby union players are more likely to have thorough stud checks before each game and to renew their footwear or studs frequently, therefore lowering the occurrence of worn, damaged studs.

All studies included in this review used a 'time-loss' injury definition. Four studies used the 'time-loss immediate' definition and these studies had the highest percentage of laceration injuries of the twelve studies, suggesting that this type of injury often forces a player off the field, but not necessarily prevents them from playing subsequent matches. The consensus statement on the injury definitions used in epidemiological studies for rugby union recommends the use of time-loss, one day injury definitions (Fuller, Molloy, *et al.* 2007). Before the publication of this consensus statement, Ekstrand *et al.* (2006) warned that the use of time-loss injury definitions can lead to underestimation of the true prevalence of skin injuries. An example is given by Gibbs (1993) where 62 lacerations were observed which needed suturing during three rugby league seasons. One of these lacerations was officially marked as an injury because the wound became infected which resulted in missed training sessions and / or matches for the injured player. This situation reiterates the importance of appropriate injury definitions when estimating the skin injury risk of players - the medical attention injury definition is less likely to underestimate skin injury prevalence compared to one-day time-loss definition (van den Eijnde *et al.* 2014b).

3.6.1 Limitations

Skin and laceration injuries are rarely the focus of epidemiological studies of sports injuries and some studies fail to report their injury prevalence. Specifically, eight studies were excluded from this review because no separate skin or laceration injury category was available in the published article. Excluding these articles can have introduced a selection bias, if the authors of the excluded studies decided not to report this injury category because the injury frequency was too low. However, some studies included a laceration injury category even though no lacerations were observed in the study (Fuller *et al.* 2008). Four out of the seven studies reporting on training injuries also found no lacerations or skin injuries during training sessions in the observed time period, but still included the injury type as a separate category (Brooks *et al.* 2005a; Fuller *et al.* 2008; Fuller *et al.* 2013; Schwellnus *et al.* 2014). According to the consensus statement on injury definitions and data collection procedures for studies of injuries in rugby union (Fuller, Molloy, *et al.* 2007), it is recommended that studies should report separate laceration and abrasion injury categories. This is reiterated by van den Eijnde *et al.* (2014b) who recommends reporting of contusions, lacerations and abrasions as separate injury categories since the causes of the injuries and the consequences for the athlete differ between these categories. The consensus statement on the collecting and reporting of epidemiological data had not been adopted by all studies in this review because some studies were carried out before the statement was published. If future studies adopt these guidelines on reporting of injury categories, comparability between studies should improve. This increased standard in reporting can aid the validity of collated injury information across a number of studies and should encourage the reporting of 'null-results', where no injury in a specific category was observed.

A generalised linear model with negative binomial distribution with log link was used for the statistical analysis. The collected data showed overdispersion, which meant it was not appropriate to apply the Poisson distribution used by similar studies on injury risk (Lystad *et al.* 2009; Williams *et al.* 2013). Using a negative binomial distribution has lowered the statistical power, therefore making a conservative estimate of the effect of the predictor on the outcome variable. For that reason, it could be that there is an effect of playing level on skin injury risk during matches (Figure 3.2) but more studies would be needed to confirm or negate this.

The injury scenario or cause of the laceration injuries occurring in the studies that have been reviewed were not published. It is thus not possible to evaluate the proportion of skin and laceration injuries that were specifically caused by studded footwear in these studies. The skin and laceration injury prevalence reviewed in this chapter is therefore not an accurate measure of skin laceration prevalence caused by studded footwear.

3.6.2 Recommendations for future epidemiological studies

Far fewer studies reported on injuries sustained by amateur players than on professional players, leaving amateur players underrepresented in published epidemiological studies. Since injury trends are not necessarily transferable between playing levels, an increase in studies observing amateur injuries is needed. Within the studies included in this review, a higher risk of bias was found in the two studies observing an amateur cohort (Bird *et al.* 1998; Kerr *et al.* 2008) compared to the studies observing professional players (Best *et al.* 2005; Brooks *et al.* 2005b; Brooks *et al.* 2005c; Brooks *et al.* 2005a; Fuller *et al.* 2008; Fuller *et al.* 2009; Fuller & Molloy 2011; Fuller *et al.* 2013; Jakoet & Noakes 1998; Schwellnus *et al.* 2014) (Table 3.1). Therefore, future studies should focus on establishing relevant skin injury frequencies in amateur cohorts through equally robust methods as those applied to professional cohorts.

The female rugby population was also underrepresented in this review. A previous study on male and female rugby players found differences in injury patterns between sexes (Peck *et al.* 2013), though skin injuries were not investigated. The studies identified in this review did not include enough female participants to determine if sex had an influence on skin injury risk. The increase in women's participation in rugby union (World Rugby news 2015) should be followed by a rise in studies documenting injuries in female rugby players.

In future, incidence and severity of laceration injuries could be assessed with a modified version of the 'skin damage and severity index' which was originally developed for assessing abrasive injuries in football (van den Eijnde *et al.* 2014a). This would allow for a more accurate estimation of the laceration injury prevalence in rugby union, minimising the non-reporting of these injuries caused by current injury definitions.

3.7 Chapter findings

The aim of this chapter was to review the skin and laceration injury prevalence reported in epidemiological studies during match play and training sessions and to identify potential differences in skin injury risk for amateur and professional players. This study used a systematic literature search of four literature data-bases from 1995 to 2016. Studies using a prospective study design, reporting on adult rugby union injuries were included in this review. A negative bionomial regression with log link was used to identify the injury risk of amateur versus professional players in training and during match play. Following the inclusion criteria, twelve studies were selected for meta-analysis. Overall, skin and laceration injuries accounted for 5.1% of match injuries and 1.5% of training injuries in rugby union. A mean injury prevalence of 2.4 skin or laceration injuries per 1000 match exposure hours was found in this study, which could be interpreted as one time-loss injury per team, per season. During match play, no difference in skin and laceration injury risk was found between amateur and professional players. In training sessions, amateur players had a higher risk of sustaining skin and laceration injuries than professional players. None of the studies in the systematic literature review reported what proportion of skin and laceration injuries was specifically caused by studded footwear, and no conclusions could be drawn about the game-scenario causing stud laceration injuries.

4 Play scenarios causing stud laceration injuries

4.1 Introduction

In Chapter 3, the prevalence of laceration injuries during match play and training sessions for both amateur and professional players was summarised. It was found that the existing literature did not report on the cause of the observed skin and laceration injuries. Therefore, the proportion of these injuries resulting from studded footwear remained unknown, and it was not possible to identify play scenarios causing laceration injuries.

Mechanical tests for sports injury risk assessment should ideally replicate the loading conditions of relevant play-scenarios (Odenwald 2006; McIntosh 2012), which makes it important to accurately define such injury scenarios. The laceration injury risk tests of current stud regulations (World Rugby 2015) have not been based on biomechanical data of stud laceration injuries (Section 2.3). Identifying the playing scenarios causing stud laceration injuries in rugby union is therefore an important first step in this research, after which the biomechanics of the most prevalent play scenario can be investigated.

Investigating the mechanism of sports injuries can be done in several ways, these include athlete interviews, video analysis of injury events and mathematical modelling (Krosshaug *et al.* 2005). Surveys are a cost- and time-effective method for obtaining information from a large cohort (Kelley *et al.* 2003). A survey also provides the opportunity to collect information on self-reported injury incidence, player attitudes towards stud checks and can help to identify the body areas predominantly involved with stud laceration injuries.

4.2 Aim and objectives

The main aim of this chapter is to investigate the self-reported frequency and cause of stud laceration injuries in a predominantly amateur cohort of rugby union players. This relates to stage B ('*Establish the prevalence of injury*') and C ('*Identify mechanism causing the injury*') of the project structure described in Figure 1.3.

This aim will be achieved through the following objectives:

1. Assess the self-reported body location and frequency of stud laceration injuries.
2. Identify the play scenario most frequently reported to cause stud laceration injuries through quantitative and qualitative methods.
3. Identify the attitude of rugby players towards stud checks.

In order to meet these aims and objectives, a survey study was conducted.

4.3 Methods

4.3.1 Questionnaire development

The best practise guidelines from Kelley *et al.* (2003) and the hands-on guide by Boynton & Greenhalgh (2004) were used for the development, conducting and reporting of the questionnaire. The formulation of the questions and answer option range was based on interviews that were conducted with two experienced rugby union players. Both players had played rugby for over 10 years and had previously experienced at least one serious laceration injury caused by studded footwear. The players were interviewed about their experience with stud injuries, the play scenario causing the injury and their thoughts on stud checks. Their answers were used to inform the questions and the range of answer options in the questionnaire. A pilot version of the questionnaire was then developed and distributed to twelve participants with mixed experience of playing rugby union. The pilot version was used to assess the intelligibility and suitability of answer options. The pilot version included the option to give feedback to the researcher at each question.

The final version of the questionnaire was developed in Google Forms and was available through an online link. It consisted of 17 questions;

1. Informed consent.
2. What position do you usually play?
3. What kind of shoes do you wear when playing rugby?
4. On average, how often did you experience a minor stud injury when playing rugby? If zero, skip to 11.

5. In your whole rugby career, how often have you experiences a substantial stud injury?
6. When taking the most severe stud injury / injuries you have experienced, can you tell what type of event caused the injury?
7. Who's studs caused the injury described in the previous question?
8. Can you remember one specific stud injury, and describe the event that caused it? (Open question)
9. Where on the body have you had skin injuries caused by studs?
10. Does the referee usually check your studs before the match?
11. What do you think about the stud check at the start of the match?
12. Do you think you should have a stud check at the start of the match?
13. Do you ever check your own studs?
14. Have you ever found your studs sharpened due to wear?
15. What is your age (in years)?
16. How long have you been playing your sport (in years)?
17. On average over these years, how often do you train and play matches a week?

The full questionnaire with the answer range options can be found in Appendix A. In question 5, injuries were defined as 'minor' when referring to an injury which did not stop a respondent from playing their sport, e.g. a chafe. For question 6, a 'substantial' injury was defined as an injury which refrained a respondent from (fully) participating in training or matches. If respondents had never experienced stud injuries, they were automatically directed to a later section of the questionnaire containing stud check and demographic questions.

4.3.2 Questionnaire deployment

The study was ethically approved by the Health and Wellbeing ethics committee of Sheffield Hallam University. Respondents had to be over 18 years old and give informed consent for their answers to be used for research purposes. Respondents could be a current or past rugby player. The final version of the questionnaire was online-only and the link was open from the 9th March 2015 until 14th April 2015. In this time period, the questionnaire was publicised through various internet platforms such as Facebook and Twitter, blog posts and on rugby discussion forums. Furthermore, the university rugby union teams and five other local rugby union clubs were approached by email and asked to forward the link to the questionnaire to their members.

4.3.3 Analysis

Closed question responses were converted to percentages of total responses for the particular question. Injury prevalence was defined as number of injuries per 1000 player exposure hours (ITEH). The results of questions 6, 17 and 18 were used to calculate an estimate of substantial stud laceration injury prevalence. Question 18 gave an indication of the frequency of play for each respondent. Each respondent's exposure hours were calculated based on a season of 28 weeks per year and 1.5 hour training sessions or matches.

One open question was included in the questionnaire (question 9). The free-text response to this question was analysed using an iterative process based on a thematic analysis approach (Braun & Clarke 2006). In this approach the researcher first familiarises him or herself with the data, before loading it into a computer assisted qualitative data analysis software program (NVivo, v10, QSR International Pty Ltd., Australia). The data is then sorted and categorized since analysis of qualitative data in its raw form is unlikely to produce systematic and significant outcomes (Denscombe 2014).

The iterative thematic analysis process firstly identified higher order themes that emerged from the data. Figure 4.1 shows three higher order themes that were identified from the data: causation event, injury severity and person responsible. Then, each higher order theme (e.g. causation event) was divided into subthemes (e.g. ruck or tackle) which were subsequently split into smaller subthemes (e.g. stamping or raking). After all the responses were appropriately coded, the emerging key concepts were analysed following the method described by Denscombe (2014). The number of answer responses in each subtheme was used as a measure to quantify how prevalent that injury scenario was in the group of respondents that had answered the questionnaire. Furthermore, word frequency counts were used to identify key concepts relating to the injury event data. The results of the qualitative analysis of laceration injury events were compared to the closed question on the same subject.

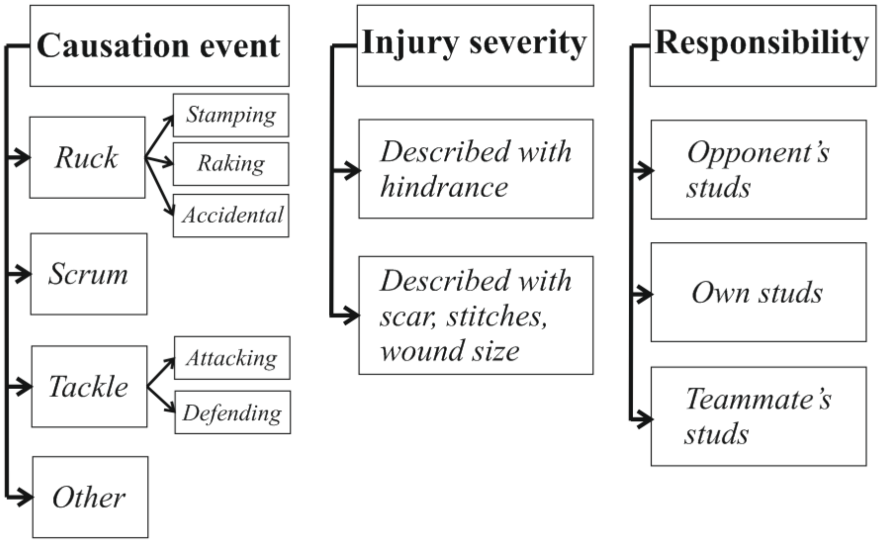


Figure 4.1: Resulting themes from the iterative thematic analysis approach on stud laceration injury events.

4.4 Results

A total of 191 rugby players filled in the questionnaire, of which 64% were forwards and 36% were backs. The respondents had on average 9.2 ± 7.0 years of rugby experience (mean ± SD), and their age was 24.7 ± 6.7 years. In this cohort, 72% of the respondents usually played in shoes with screw-in studs, whilst the other 28% preferred moulded studs. Bladed studs were the most commonly used moulded stud.

4.4.1 Injury prevalence

Out of the 191 players that filled in the questionnaire, two respondents stated they had never experienced minor stud injuries (question 5, 1% of total). Fifty-three respondents stated they never had experienced a substantial stud laceration injury (question 6, 28% of total). Mean stud laceration injury prevalence, estimated based on total injuries, playing frequency and playing years, was 3.2 ITEH. The distribution of the estimated injury prevalence is shown in Figure 4.2.

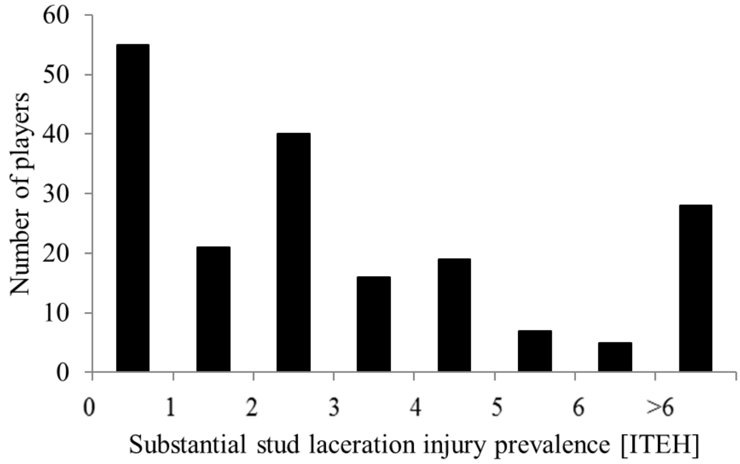


Figure 4.2: Spread of the estimated stud laceration injury prevalence of respondents.

4.4.2 Injury location

In question 10, respondents were asked to identify all body locations where they had experienced stud lacerations. Respondents were able to provide multiple answers to this question via tick-boxes. The results are shown in Table 4.1. The upper leg was the body location which was most frequently reported as injured - 81.7% of respondents reported injuries here.

Table 4.1: Body location(s) where respondents reported they experienced stud laceration injuries.

Body part	**Responses**
Foot / Ankle	60.7%
Lower leg	75.9%
Upper leg	81.7%
Chest and belly	25.7%
Back	29.8%
Arms	51.8%
Hands	48.7%
Head / neck / face	34.0%

4.4.3 Injury causation

Play scenarios causing stud laceration injuries were evaluated through a multiple choice question (question 7) and an optional open question (number 9). Table 4.2 shows the proportion of respondents that reported one of the pre-determined injury scenarios as the event causing their stud laceration injury. In this multiple choice question, over half of the respondents (54.5%) answered that their injury occurred during a rucking situation. The tackle was responsible for 27.1% of the injuries (Table 4.2).

Table 4.2: Results of closed question on the play scenario causing a stud laceration injury.

Injury event	**Responses**
Being tackled	11.4 %
Tackling someone	15.7 %
On the ground in a ruck	50.2 %
Stepping over a ruck	4.3 %
Collapsed maul	8.2 %
I don't remember	5.1 %
Other	5.1 %

In the open-ended question responses, similar results were seen with 56% of the injuries happening during the ruck and 34% resulting from a tackle. In 35% of all free-text responses, respondents described deliberate stamping in the ruck as the cause of their injury. This happened when players would drive over the ruck in order to gain or keep possession of the ball. Other causes of injury during the ruck were raking, where a player moves their foot in a backward motion to kick the ball [backwards] to a teammate (6% of free-text responses). Common free-text answer categories and example quotations are shown in Table 4.3.

During tackling, the player who was making the tackle would get injured through a shoe from the opponent, flicking up in their face or into their chest. A tackle made from behind accounted for 14% of the injuries described in all free-text responses. An example quote can be found in Table 4.3.

Other, less frequent injury scenarios were whilst a ball was kicked (6% of free-text responses), a collapsing maul (4% of free-text responses), and during the line-out (4% of free-text responses).

Table 4.3: Common open question answer categories and example quotes from the qualitative analysis.

Injury event	**Responses**	**Example quote**
Stamping in the ruck	34%	"*As I was tackled the opposition came over to ruck and purposely stamped and scraped their studs down my shin*"
Tackling from behind	14%	*"I tackled the opponent from behind when they were sprinting and I wrapped my arms around her waist and pushed forward. Her stud went into my sternum as her foot came up*"
Line-out	4%	*"I was lifting in the line out and my team mate accidentally lifted their legs"*
Collapsing maul	4%	*"Collapsed maul, I was at the bottom, a stray boot came down and the studs caught the left side of my face"*
Collapsing scrum	3%	*"Scrum pivoted and collapsed, leg drive of either team mates or opposite team onto shin"*

In question 8, respondents were further asked which player's shoes were responsible for the stud laceration injury they had received. In 66% of cases the injury was caused by the opponent's studs, in 14% by a teammate or through their own studs, and it was unclear which player was responsible in the remaining 20% of the cases.

4.4.4 Stud checks

In question 11 to 15, respondents were asked about their habits and opinions with regarding to checking studs on sharpness. The results of the stud check questions are shown in Table 4.4. Eighty-five percent of the respondents stated that stud checks were commonly performed by the referee prior to their matches. The majority of respondents (91%) were in favour of having stud checks by the referee prior to the match, although according to 34% of respondents this check does not happen thorough enough. Out of the respondents who check their own studs, 35% have at some point found them sharpened due to wear.

Table 4.4: Responses to stud check questions.

Question	**Answer options**	**Responses**
11. Does the referee usually check your studs before the match?	Yes, (almost) always	85%
	Sometimes	13%
	No (almost) never [jump to 13]	3%
12. What do you think about the stud check at the start of the match?	Good, definitely needed to keep the players safe	57%
	Good, but usually does not happen thoroughly enough	34%
	Not useful, wastes our time	5%
	Other […]	4%
13. Do you think you should have a stud check at the start of the match? [only when filled in 'no' at 11]	Yes	80%
	No	0%
	Not sure / no opinion	20%
	Other […]	0%
14. Do you ever check your own studs?	No [skip 15]	26%
	Yes	74%
15. Have you ever found your studs sharpened due to wear?	Yes	35%
	No	58%
	Can't remember	7%

4.5 Discussion

The questionnaire described in this chapter was completed by 191 rugby union players. Almost all (99%) of the respondents stated they had experienced minor stud injuries, and 72% reported one or more substantial injuries during their rugby career. The estimated skin injury prevalence for this cohort during training sessions and matches combined was 3.2 ITEH, which is higher than the 2.4 ITEH during matches found in the systematic review and meta-analysis (Chapter 3). The questionnaire results therefore suggest that laceration injuries caused by studded footwear are underestimated when measured by traditional epidemiological research. This finding is in line with previous literature (Ekstrand *et al.* 2006; Gibbs 1993). However, there are two potential biases that could have resulted in incorrect injury prevalence being inferred from the questionnaire data. Firstly, the injuries in this questionnaire were retrospectively reported which is less reliable than a prospective method. Secondly, players may be more likely to fill in the questionnaire if they perceived stud injuries in rugby a problem. In the questionnaire results it was not possible to determine the skin injury prevalence for match and training injuries separately. Training sessions are usually associated with lower injury prevalence (Brooks *et al.* 2005b; Brooks *et al.* 2005c). Separating match and training injuries could have resulted in higher stud laceration match injury prevalence in the outcome of the questionnaire.

Just over half (54%) of the laceration injuries sustained were caused in a rucking situation, which makes the ruck a clear risk factor for laceration injuries. The free-text responses described the cause of their injuries in varying detail. Overall the free-text responses confirmed that most of the laceration injuries had happened in the ruck. Respondents described a deliberate stamp in the ruck by an opponent player in 35% of the cases. The ruck has previously been identified as the dominant cause of injuries sustained by professional forward players during matches (Brooks *et al.* 2005b) and the results from the questionnaire support this. Most respondents reported stud laceration injuries in the lower body (foot, ankle, lower leg, upper leg; 95% of respondents). Fewer injuries were reported on the trunk (25.7 - 29.8%); this could be due to the thick jersey rugby players tend to wear, perhaps suggesting that clothing can offer some skin protection to players.

A referee is expected to check the studs of every player on the field before a match starts (Law 4.5, World Rugby 2016). It is thus recommended for players to check their own footwear regularly on sharpness and wear. Three-quarters (74%) of the respondents stated they check their own studs, and one-third (34%) of these respondents reported they have found their own studs sharpened due to wear. This finding supports the need for stud checks prior to matches. Future studies should investigate the effect of different stud materials and their wear on this sharpening effect, which is likely to be related to laceration injury risk.

The questionnaire used in this research was predominantly aimed at amateur rugby players, and the playing frequency of the respondents (76% of respondents played / trained 1 - 3 times a week) confirms this demographic. Potentially, different results could be found at the elite level of play but investigating injury scenarios specific to elite players was outside the scope of this project. The final version of the questionnaire was only available online, which might have excluded people with no access to the internet. Further, respondents were asked retrospectively to report the play scenario causing their injuries which could have biased the results. Recall bias is an inherent problem with retrospective research and the results should be treated with this limitation in mind. Krosshaug *et al.* (2005) recommends using retrospective interview data on injury mechanisms only for identifying the playing situation that caused the injury and athlete behaviour during the injury, and not for obtaining more detailed information such as injury kinetics and kinematics.

4.6 Chapter findings

The aim of this chapter was to investigate the self-reported frequency and cause of stud laceration injuries. A survey study was conducted which retrospectively asked players on their experience with stud laceration injuries. The cause of the injury was determined by combining the results of an open and closed question. A thematic analysis approach was used for the qualitative analysis of free-text responses. The questionnaire was filled out by 191 rugby union players. The estimated laceration injury prevalence specifically caused by studded footwear was 3.2 ITEH (training and match combined), which is higher than the match skin and laceration prevalence previously reported in literature (2.4 ITEH). Despite the different methods in obtaining these injury prevalence numbers, the results support previous statements that suggested that epidemiological research is prone to underestimating skin laceration injury prevalence. The lower limbs were most likely to incur stud laceration injuries, with 95% of respondents stating they have been subject to minor or substantial stud injuries to those regions of the body. Stamping in the ruck was identified as the dominant play scenario causing laceration injuries in rugby union. As such, it should be the focus of any test method designed to replicate stud laceration scenarios. The next stage of the research obtains the kinematic and kinetic parameters of stamping in the ruck; these parameters can then be used to inform the design of a representative test method for stud laceration injury risk.

5 Biomechanics of stamping in the ruck

5.1 Introduction

Chapter 4 identified stamping in the ruck as the dominant game scenario of stud laceration injuries. Although officially an illegal action in rugby union, stamping was reported as a frequent cause of these injuries. This stamping scenario should therefore be the focus of any test method designed to replicate stud laceration injury scenarios. An in-depth insight into the kinetics and kinematics of this injury scenario is needed to inform the design of a new test method for assessing laceration injury risk of rugby stud designs.

Test parameters of the current studded footwear regulations were discussed in Section 2.3. Communication with World Rugby (M Douglas, pers. comm., 21 January 2015) and the British Standards committee (M Marshall, pers. comm., 23 August 2016) showed that the test parameters of the current studded footwear test methods have been based on a running heel strike impact of an 80 kg male. However, this type of impact is not known to be representative of rugby-related stud-skin interactions. To the best of the author's knowledge, no studies investigating the biomechanical parameters of stamping in the ruck in rugby union have previously been published.

During slide tackling in football, a movement which could be comparable to stud-skin interactions, the effective mass of the foot has been estimated at 0.1 kg (Ankrah & Mills 2003) and 4.6 kg (Payne *et al.* 2013). According to Nigg (2010) the effective mass of the foot during a running impact is between 0.5 and 2 kg for forefoot landing and between 6 and 20 kg for a heel strike. The large difference between these effective masses reiterates the need for validating such impact kinetics by biomechanical studies.

Obtaining detailed biomechanical information on injury scenarios is important (Bahr & Krosshaug 2005), yet ethically difficult due to its injurious nature (Krosshaug *et al.* 2005). The biomechanics of sports injury scenarios have informed the development of a variety of mechanical test devices (Clarke *et al.* 2013; Grund *et al.* 2007; Peterson *et al.* 2008). These studies all applied a different approach for obtaining kinetic and kinematic data from their athletes whilst complying with ethical regulations. In order to measure the kinetics and kinematics of rugby stamping impacts it should be possible to replace the player receiving the injury with an artificial replica, minimising the injury risk for the participants involved.

Current test methods that asses the laceration injury risk of studded footwear use a human skin simulant to assess the damage caused by studs. Human skin is loading rate dependent (Wood & Bladon 1985; Dąbrowska *et al.* 2015). Implementing game-representative impact conditions during studded footwear tests is therefore important to appropriately analyse skin simulant damage. Quantifying the force required to lacerate human skin is considered complex due to the large number of influencing variables; e.g. sharpness and material of the impacting object, and its inbound velocity (Parmar *et al.* 2012). Test methods for assessing laceration injury risk should therefore focus initially on replicating inbound velocity, attack angles and impact mass of injury events.

5.2 Aim and objectives

The aim of this chapter is to identify the kinetics and kinematics of simulated rugby stamping impacts during a rucking scenario. This chapter relates to stage D (*'Identify injury parameters'*) of the project structure described in Figure 1.3.

The aim is achieved through the following objectives:

1. To measure the inbound velocity magnitude, inbound velocity direction and foot orientation angle at initial impact.
2. To measure the total peak force and individual peak stud force during impact.
3. To determine stud effective mass and stud impact energy during impact.

5.3 Methods

5.3.1 Study protocol

All procedures in this study were approved by the Health and Wellbeing ethics committee of Sheffield Hallam University. Twelve male participants (mean ± SD: age: 27.7 ± 4.2 years, height: 176.5 ± 5.8 cm and weight: 76.3 ± 7.6 kg) were recruited. All participants were recreationally active and gave informed consent to participate in this study. Ten out of twelve participants had previous experience of playing rugby.

Participants were asked to set up a one-on-one rucking scenario and perform a stamping motion onto the chest of an anthropomorphic test device (ATD: Hybrid III 50th percentile man, Humanetics Innovative Solutions, Plymouth, USA), used as a surrogate player (Figure 5.1). Each pair of participants was shown an instructional video to help standardise the test protocol. In this instructional video was shown how to safely bind and push during the experiment. No run-up was allowed for safety reasons; participants were starting from a stationary position. Participants were accustomed to the required movements by performing test trials until they felt comfortable with their rucking partner and the movements. Participants were asked to perform 10 stamping trials each. A trial was successful if the participant pushed their partner away from the ATD ('rucked over') and stamped on the ATD's chest without losing balance.

Figure 5.1: Two participants in a one-on-one rucking formation with the participant on the right performing a trial.

Participants wore a pair of rugby shoes (Kipsta Density 300, Decathlon, size 8.5 - 10.5 UK), which had an 8-stud configuration of aluminium rounded studs (10 mm diameter and minimum stud spacing of 32 mm). Three high contrast circular markers were placed on each shoe; at the heel cup level with the lateral malleolus, at the lateral front stud and at the lateral rear stud (Figure 5.2).

Figure 5.2: Illustration of shoe and three markers used to calculate shoe velocity and orientation.

5.3.2 Measurement equipment

Figure 5.3 shows a schematic of the test set-up of the measurement equipment. Two gen-lock synchronised high-speed cameras (Phantom Miro Lab 320, Vision Research, Wayne, USA) were positioned 3 m away from the ATD at an angle of 60° to each other. Each impact was filmed at 1000 frames per second for 1.5 s at a resolution of 1280 x 800 pixels. A three-dimensional motion capture volume, measuring 1 x 1 x 1 m was calibrated using in-house developed software (Check3D, http://www.check3d.co.uk/). The global coordinate system was defined with the z-axis corresponding to the true vertical.

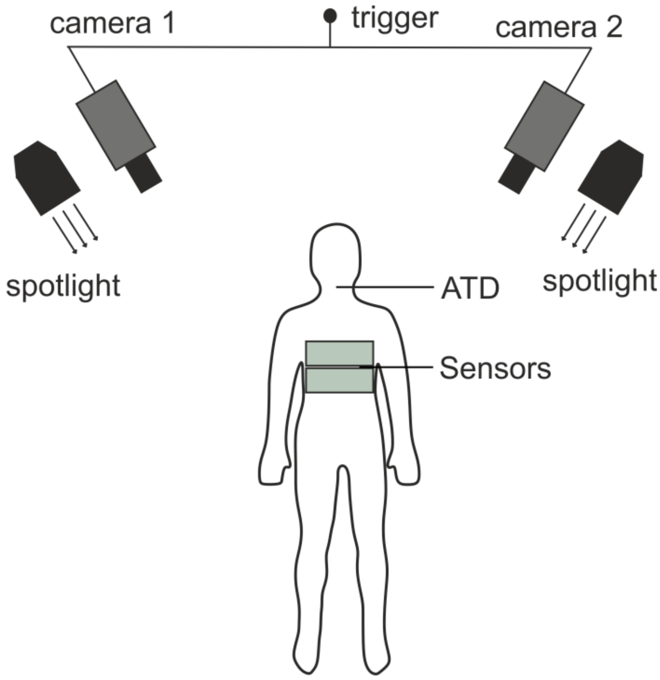


Figure 5.3: Schematic representation of test set-up.

Two pressure sensors (Tekscan, F-scan, 3000E 'Sport') sampling at 750 Hz were used to measure exerted pressure of each impact. Each sensor had 956 pressure sensing elements with a spatial resolution of 25 mm^2. The sensors were calibrated with a custom developed calibration method, based on peak forces expected during rugby stamping impacts (Oudshoorn *et al.* 2016b). To find the expected peak forces, a pilot study with seven rugby players performing three stamps each was performed. Subsequently, a custom calibration procedure was developed, using a drop hammer and damping materials to replicate the experimentally observed forces and loading rates. The impact force was simultaneously measured with the pressure sensors and a force platform (9281CA, Kistler Instrument Corp, Winterthur, Switzerland; sample frequency 1000 Hz). Expected error of measurement with the custom calibration method was 7.5%, versus 132% with the manufacturer's standard calibration method. The standard calibration method was based on the force exerted by the body mass of the researcher in a static position which was approximately 600 N. The expected peak forces during stamping were in the range of 1800 to 2800 N (Oudshoorn *et al.* 2016b) and the custom calibration method included impacts resulting in these forces. Full details on the calibration procedure can be found in Appendix B. The two pressure sensors were placed on the chest of the ATD in such a way that maximised their surface area and minimised the overlap between sensors (Figure 5.3). The chest was used because of its relatively large surface area, making full use of the sensors' size. The pressure sensors were replaced when visual damage showed, or after a maximum of four participants were measured.

5.3.3 Analysis

A trial was included for kinematic analysis if the calibration reprojection error was below 2.0 mm and the cameras captured at least 10 ms prior to first contact. Marker position was manually identified for the three shoe markers shown in Figure 5.2 for 10 ms prior to first impact. Mean velocity of the three shoe markers was determined over these 10 ms prior to first impact and defined as the inbound velocity. The orientation of the shoe was calculated over this same time interval following a modified approach to that described by Driscoll *et al.* (2015). For this method, the relative position of each marker and stud is required. A three-dimensional (3D) static reference position of shoe markers and studs were obtained by a 3D laser scanner (CIMCORE Arm 5024, EuroPac

3D, Crewe, United Kingdom). A local shoe coordinate system was then defined using scanned markers as reference points. Subsequently, the direction cosine matrix (order ZXY) was used to transform local coordinate system coordinates into the global coordinate system, deriving three Euler angles (pitch, yaw and roll). A positive pitch angle (rotation about the x-axis) refers to plantar-flexed shoe orientation prior to impact. Yaw and roll angle were not used in this study.

Within the calibration procedure, the pressure data from the pressure sensors was multiplied by their corresponding surface area to calculate force. Calibrated stamping impact data was loaded into MATLAB® (R2015a, the Mathworks Inc, Massachusetts, United States). Peak total force was calculated by summing the force of all sensing elements of both sensors per time frame, and consecutively finding the frame with the highest summed force. A trial was included for total force analysis if no studs contacted the ATD outside the pressure sensors. For individual stud force analysis, partial contact of the studs with the pressure sensors sufficed.

A custom written script was used to calculate peak stud force. Preliminary analysis of pressure sensor data showed that the number of sensing elements activated by one stud during an impact was related to on the impact force of the stud, with higher forces activating a larger number of elements. Data from the pair of sensors was combined into one grid and the three elements with the highest-pressure values were identified as the starting points (black square, Figure 5.4). A grid of 5 x 5 elements around a peak defined a stud impact (grey squares, Figure 5.4). A 'pass or fail' criterion was put in place for consecutive elements because of the growing nature of the impact (white squares, Figure 5.4). An element 'passed' and thus was included in the stud force when it had a lower or similar force (≠ 0 N) than one of its neighbouring elements which are closer to the starting element. If an element passed, the algorithm then evaluated three neighbouring elements (or five when on a corner) to asses if the neighbouring elements should pass as well (Figure 5.4). Every element which passed would become a 'searching' element itself, and would search in three outward directions for lower or similar forces. An element 'failed' when it had a higher force than all of its neighbouring elements; it was assumed that the element was part of a different stud impact. Figure 5.5 shows an example of the pressure sensor data from an impact with three studs. A maximum of five elements in each direction from the starting point was included in a single stud impact, corresponding to 25 mm distance. The closest distance between two studs in the shoes used for this research was 32 mm. The process of defining stud impacts was repeated for the three highest pressure values in each time frame. Peak stud force was defined as the highest single stud force at a single time frame.

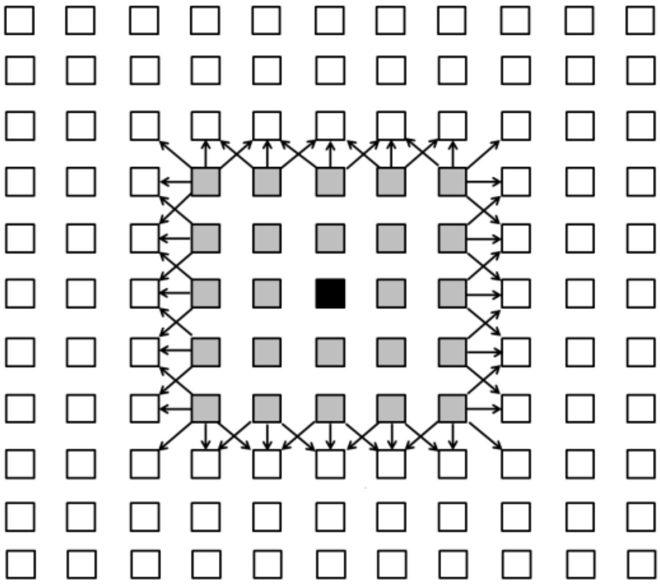


Figure 5.4: Process of defining a stud impact with a central starting point (black), elements always included in stud impact (grey) and search grid (white). Each element is 5 x 5 mm in size.

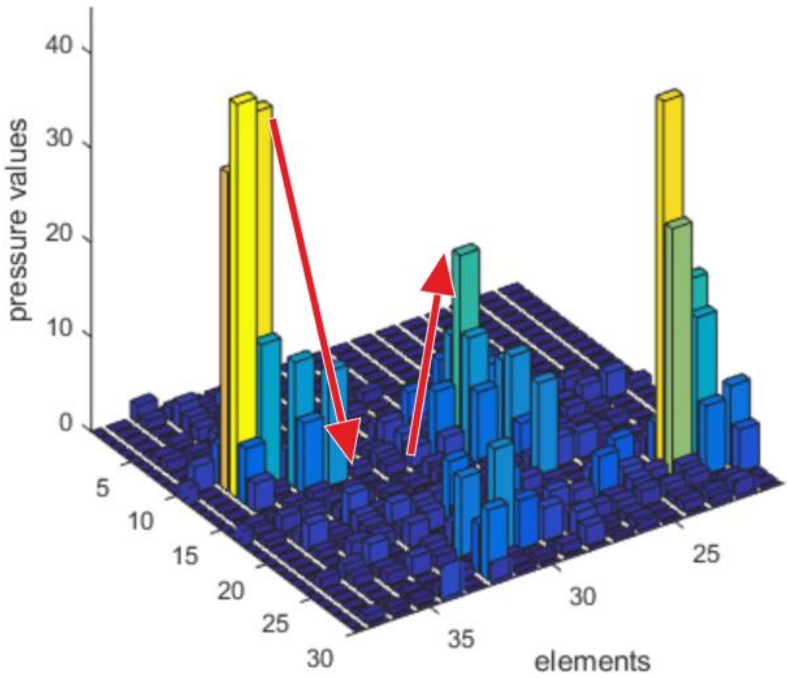


Figure 5.5: Example of the search algorithm on a grid with multiple stud impacts.

The effective mass (m_e) of the total stamping impact and per peak stud impact was calculated using Equation (2.1) which was previously presented in Section 2.4.4. The values of t_1 (time of first impact) and t_2 (velocity reaches approximately zero) were obtained from visual analysis of the high-speed videos. Peak stamping force coincided with a velocity value of approximately zero (t_2). The impact energy (E_{kin}) of the total stamp and individual studs was calculated using Equation (5.1):

$$E_{kin} = \frac{1}{2} m_e \Delta v^2 \tag{5.1}$$

Where change in velocity (Δv) is equal to the inbound velocity; assuming that at maximum displacement the foot becomes momentarily stationary and all energy is dissipated.

5.4 Results

A total of 110 trials from 12 participants were measured in this study. Table 5.1 shows the mean and range of the shoe velocity and pitch angle in the 10 ms prior to first contact. Following the inclusion criteria for kinematic analysis, 75 trials from eight participants were used to identify these pre-impact kinematics. The reprojection error of the camera calibration for the included trials was 0.26 - 0.70 mm. Accuracy of the manual identification of shoe markers was ± 0.13 mm (SD over 105 marker identifications).

Table 5.1: Pre-impact kinematics of stamping in the ruck (N = 75).

	Horizontal shoe velocity (m/s)	**Vertical shoe velocity** (m/s)	**Resultant shoe velocity** (m/s)	**Shoe pitch angle** (°)
Mean	3.0	2.8	4.3	10.3
Range	0.1 - 5.7	1.1 - 4.5	2.1 - 6.3	-13.2 - 35.4

The range of resultant inbound velocities of the shoe during stamping impacts was 2.1 to 6.3 m/s. The mean resultant inbound velocity was 4.3 m/s. The inbound velocity of the shoe during rugby stamping impacts consisted of a horizontal and vertical component, and they were of a similar magnitude (mean 3.0 and 2.8 m/s, respectively).

The pitch angle of the shoe prior to impact ranged from 35° (plantarflexion) to -13° (dorsiflexion), as shown in Figure 5.6.

Figure 5.6: The range of shoe pitch angles observed.

Fifteen trials were excluded from total force analysis because one or more studs missed the sensors. Therefore, peak total force was calculated over 95 trials and peak stud force over 110 trials. Table 5.2 shows the mean and range of the impact kinetics measured. Peak total force measured during the stamping impacts ranged from 482 to 2670 N; peak stud force ranged from 93 N to 370 N. Effective mass and impact energy were calculated using both force and velocity data. Following the inclusion criteria for force and velocity data, the stud effective mass and stud impact energy was determined from 75 trials and total effective mass and total impact energy from 67 trials. The total effective mass ranged from 1.6 to 13.5 kg and total impact energy ranged from 15 to 122 J (Table 5.2).

Table 5.2: Impact kinetics of stamping in the ruck

	Peak total force (N)	**Total effective mass** (kg)	**Total impact energy** (J)	**Peak stud force** (N)	**Stud effective mass** (kg)	**Stud impact energy** (J)
Mean	1245	6.5	56.9	214	1.2	9.5
Range	482 - 2670	1.6 - 13.5	15.1 - 122.4	93 - 370	0.5 - 2.9	1.5 - 18.7
No. of trials	95	67	67	110	75	75

5.5 Discussion

Chapter 2 showed that limited kinetic and kinematic information on stud-skin interactions was available in literature. Stamping in the ruck in rugby union has been reported as the dominant cause of laceration injuries (Chapter 4, Oudshoorn *et al.* 2016a), and kinetic and kinematic information of such impacts should be used to inform mechanical test methods assessing laceration injury risk of stud designs. Stamping in the ruck, however, is not allowed under the current laws of game of rugby union (World Rugby 2016). This could explain the low body of literature on the biomechanics of this particular movement.

Current studded footwear standards (BS 6366:2011 and Regulation 12, World Rugby, 2015) stipulate an inbound velocity of ~1 m/s (50 mm free fall) and an 8.5 kg drop mass. These test parameters have been based on a heel-striking running impact of 80 kg male and the representativeness of this movement to stud laceration injury scenarios has not previously been verified. Mean resultant inbound velocity measured in this study was 4.3 m/s, which is markedly higher than prescribed in the current studded footwear standards. Mean effective mass per stud in this study was lower than the current standards (1.2 vs. 8.5 kg, respectively). However, mean stud impact energy was higher than the current standards (9.5 vs. 4.2 J, respectively). This was due to higher inbound velocities measured in this study, compared to those prescribed by the studded footwear standards. It can be concluded that current studded footwear standards overestimate the effective mass per stud but underestimate the inbound velocity and impact energy for stud-skin interactions during stamping. It is therefore necessary to develop of a new task representative mechanical test method.

The results of this study showed a wide range of impact parameters (Table 5.1 and Table 5.2). Developing a test method based on all possible combinations of the range of impact parameters will lead to a high number of tests that need to be performed per stud. Combining these impact parameters into clusters, representing different movement solutions utilised by participants, could reduce the number of tests needed per stud design and maintains a combination of test parameters that is representative of the measured impacts.

Both the camera measurements and pressure sensor measurement led to measurement error. For the camera measurements, calibration reprojection error needed to be below 2.0 mm for a trial to be included in the analysis. The mean foot velocity in this study was 4.3 m/s and the measurement frequency was 1000 Hz. This meant that on average, the foot movement between frames would have been 4.3 mm. The mean foot velocity was calculated over ten frames prior to first impact to further minimise this potential measurement error. For improving the measurement accuracy of the pressure sensors used in this study, a custom calibration protocol was developed based on expected peak forces and loading rates. This led to a reduction in the expected error of the sensors as compared to manufacturer's recommended calibration, though there was still a mean expected error of 7.5% (Oudshoorn *et al.* 2016b). The results of this study have to be interpreted with this limitation in mind. The pressure sensors used in this study measured only the force perpendicular to the sensor's surface. Unlike many force plates, the pressure sensors are unable to measure shear force. The raking phase of the stamping impacts is likely to have created shear forces and these were not recorded in the current test set-up.

Further limitations to this study include that there was no minimum playing experience in rugby union for the participants. This study aimed to quantify impact parameters associated with amateur rugby players. Stamping in the ruck is an illegal action which is unlikely to require a specific skill level from the participants. The body weight of the participants (76 ± 7.6 kg) was low in comparison to professional male rugby players (100 ± 12.1 kg; Brooks *et al.* 2005b). However, amateur players have previously been found to have a similar body mass (77.6 ± 10.6 kg; Nicholas, 1997) to the participants in this study. Selecting amateur, and thus lighter, participants for this study could therefore have led to a lower effective mass in comparison to professional, usually heavier players. Regardless, the mean stud impact energy generated by participants was still twice as high as recommended in the current standard (9.5 J versus 4.2 J, respectively).

The Hybrid III ATD was used as a surrogate player in this study. The pressure sensors were placed on the ATD's chest due to its relatively large and flat surface area. The pressure sensor placement can have influenced the stud angles measured in this study; i.e. if the pressure sensors would have been placed on the thigh, the relative angle between the curved surface of this body part and the stud would have been different if the foot was kept in the same absolute position. According to the pressure sensor's manufacturer, greater accuracy can be achieved if the sensors are placed on a relatively flat surface, which influenced the decision to place the sensors on the chest. The chest stiffness of this ATD has previously been found 10% stiffer than the chest stiffness of human volunteers (Backaitis & St-Laurent 1986), which could have led to an increase in peak stamping force in this study. Nevertheless, the use of a Hybrid III ATD as a surrogate player was necessary for both safety and ethical reasons.

5.6 Chapter findings

The aim of this chapter was to identify the kinetics and kinematics of simulated rugby stamping impacts. Impact parameters of 110 stamps, performed by 12 participants, were recorded with high-speed cameras and pressure sensors. Inbound velocity, shoe pitch angle, stud force, stud energy and stud effective mass were calculated for the stamping impacts. Mean inbound velocity was 4.3 m/s and both heel striking and toe-down impacts were observed. Stud effective mass was 1.2 kg (mean; range 0.5 - 2.9 kg). The findings of this study show that the current test method described in Regulation 12, for assessing laceration injury risk of stud designs, is not representative of the rugby stamping impacts that were measured. Specifically, a new test method should adopt a higher inbound velocity and lower impact mass. Further, the shoe pitch angle and inbound velocity angle should be incorporated. In this study the mean and range of the observed stamping impact parameters were calculated. In the next chapter, the results of the current study are clustered and used to inform the design of a new test method for stud designs.

6 Development of a mechanical test method

6.1 Introduction

Mechanisms of the target injury must be well understood when developing mechanical tests for sports equipment (McIntosh 2012; Odenwald 2006). In Chapter 4, stamping in the ruck was identified as one of the most common events causing laceration injuries. Consequently, the purpose of Chapter 5 was to better understand this identified injury mechanism by measuring its kinetics and kinematics. This chapter will use these impact parameters to inform the design of a new, game-representative test method for assessing the laceration injury risk of stud designs.

The design process by Pugh (1991) provides a model for product development in engineering. In this total design activity model, the design core consists of the user needs, a product designs specification (PDS), the conceptual designs, the detailed designs and manufacturing and sales. Figure 6.1 shows the design process that was followed for the development of this new test method, informed by the Pugh model. The design structure includes an iterative process of developing conceptual designs, pilot testing, developing new conceptual designs and a final design.

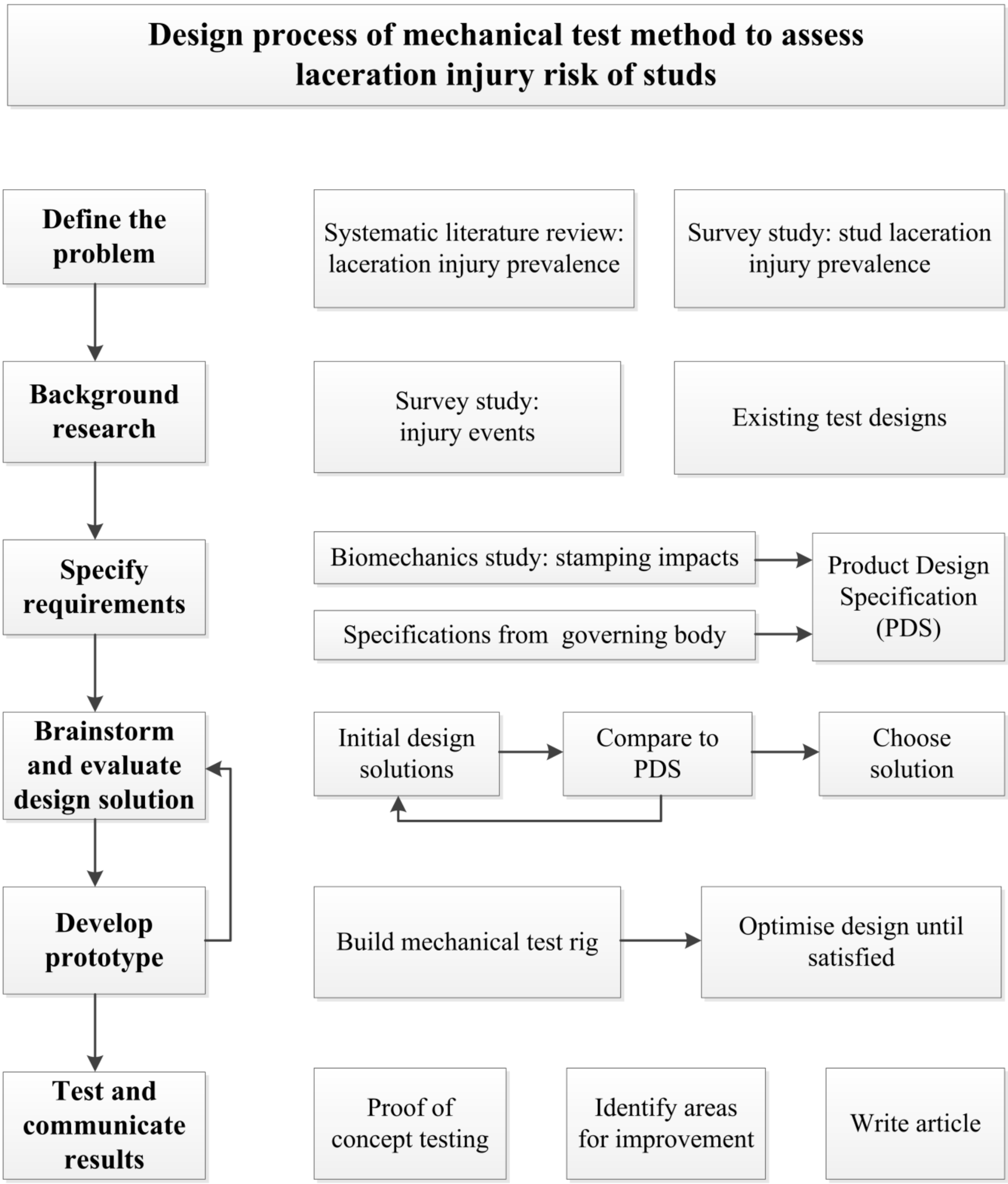


Figure 6.1: Design process based on the model by Pugh (1991), adjusted to suit the development of a mechanical test method to assess the laceration injury risk of stud designs. The right hand side outlines the steps taken for this development.

6.2 Aim and objectives

The aim of this chapter is to design and build a mechanical test method that is able to replicate the stamping impacts measured in Chapter 5 as closely as possible. This chapter relates to stage E ('*Develop test method*') of the project structure described in Figure 1.3.

The aim is achieved through the following objectives:

1. Transform the impact parameters from Chapter 5 into test parameters for a mechanical test rig.
2. Identify the design criteria of the test method.
3. Design a mechanical test rig using the identified design criteria.
4. Manufacture the test rig.

6.3 User needs

6.3.1 Governing body

The potential adoption of a new mechanical test method to assess the laceration injury risk of studded footwear in rugby union is determined by World Rugby, the international governing body of rugby union. It is therefore important for the mechanical test design to align with the strategy and opinion of World Rugby. An interview was conducted with the Research, Artificial Turf and Equipment manager of World Rugby (M Douglas, pers. comm., 21 March 2017). In this interview, World Rugby expressed their desire to collaborate with governing bodies of other field sports - specifically FIFA - to develop a single, overarching standard for studded footwear in field sports. World Rugby stated:

"We have been talking to FIFA, field hockey, American football - because as one sport, we are too small. A lot of players within rugby wear football boots."

Compared to association football, rugby union is a relatively small sport. Aligning the studded footwear regulations between governing bodies would especially benefit amateur players and players in countries where rugby union is not a popular sport; allowing them to use studded footwear which is not sport specific. The test method design should be flexible enough to handle conceivable impact parameters relevant to football in order to meet World Rugby's wish for an overarching standard.

World Rugby was also asked if there were any limits on the complexity of a future studded footwear test method, its maximum number of tests performed, and its labour intensiveness. They stated that the complexity and labour intensiveness of a new mechanical test was unlikely to inhibit the uptake of this test method. A multitude of tests could be performed in the case that this would result in a better understanding of how the stud performs. In the current regulations, a stud should pass two tests that both require one set of test parameters. Increasing this number of tests or its test parameters was acceptable. As an outcome variable, a quantitative number would be preferable over the 'no greater risk of injury' criteria that is described in the current equipment regulations (World Rugby 2015). This would allow the performance assessment of different stud designs. Consistency of a test method between users (research centres and test houses) is important to World Rugby and adequate calibration procedures should be in place. Future proofing of the new stud laceration test method should be considered; any change in World Rugby's equipment regulations means re-validating the test method and re-testing of existing equipment. World Rugby stated that potential future changes in their equipment regulations should ideally only reflect innovative advances in test methods, sensors and materials. The priorities identified by the interview with World Rugby are summarised in Table 6.1.

6.3.2 End users

Evaluating sports equipment to standards or regulations is often performed by test houses. Test houses provide an objective evaluation of a product's performance to the manufacturer and the relevant governing body. To inform design requirements from a likely end user's perspective, the managing director of the UK branch of an international test house specialising in sports equipment testing was interviewed (Labosport, C Young, pers. comm., 23 March 2017). The test house employs an engineer for building test equipment when necessary. For each project, the cost-effectiveness of buying off-the-shelf equipment (where possible) and building testing equipment in-house was balanced. The interview showed it is important to use unambiguous, concise language when describing a test method. Further, every test parameter needs to be defined including their tolerance (margin of error). In the experience of this test house, using skin simulants in standardisation can be problematic due to the inconsistency in biological simulants such as porcine skin and the low biofidelity of other skin simulant materials. The priorities identified by Labosport are summarised in Table 6.1.

Table 6.1: Summary of the priorities identified by rugby union's governing body and an UK based test house for sports equipment.

User type	Priorities identified by user
Governing body (*World Rugby)*	- Developing an overarching standard for field sports - Best understanding possible of how a stud performs - Prefers to be able to rank the performance of range of studs - Consistency between test centres
Test house *(Labosport UK)*	- Both bespoke and off-the-shelf test equipment possible - Unambiguous, concise language in published standards - Include margins of acceptable error in published standards - Avoid biological skin simulant materials where possible

6.4 Product design specification

Chapter 5 presented an in-depth investigation into the kinetics and kinematics of stamping in rugby union. High speed video footage from this study showed that in 53% of the analysed trials, a raking motion was observed after initial foot impact. Figure 6.2 shows the observed foot movement during trials where raking was present, split up in an 'initial impact phase' and a subsequent 'raking phase' Although not all trials showed a clear raking phase after initial impact, the raking phase is thought to be an important factor in the development of lacerations (Section 2.3.1 and Section 4.4.3, Oudshoorn *et al*. 2018).

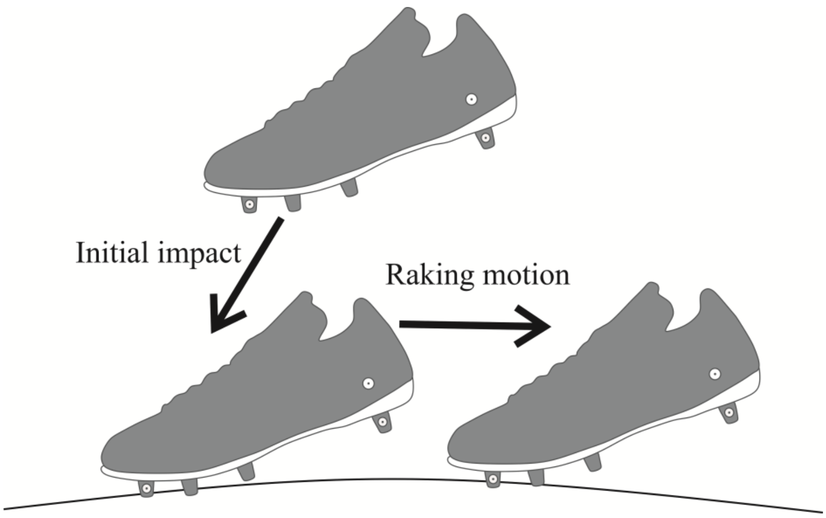


Figure 6.2: Foot movement during stamping in the ruck, split into an initial impact and raking motion.

In Chapter 5, impact parameters relating to the initial impact phase were presented. Variance within these impact parameters was large, e.g. 2.1 to 6.3 m/s for inbound velocity (Table 5.1) and 0.5 kg to 2.9 kg for stud effective mass (Table 5.2). For a realistic test design, the range of these impact parameters needed to be reduced and the intervals at which they could be controlled identified. To achieve this, two approaches were used: a clustering method for identifying common movement solutions during the initial impact phase, and a representative trial for the following raking phase.

6.4.1 Initial impact phase

An unsupervised k-means clustering algorithm was used to group the results presented in Chapter 5. The k-means clustering algorithm is based on a centroid approach, where group centroid points are selected at random and data points are assigned to a group based on proximity. Its centroid is subsequently moved to evaluate if this provides better grouping. K-means is an unsupervised clustering method, without prescribed labels or a priori group values of the data. The silhouette value in a k-means cluster algorithm summarises the quality of the resulting clusters. It assesses how well a data point fits with its assigned cluster and how dissimilar it is from other clusters, on a scale from 0 (no fit) to 1 (near to perfect fit). The silhouette value can be used to determine the number of groups that would fit the data best. Stud energy, stud effective mass and resultant inbound velocity were given to k-means as its input parameters. The mean silhouette value reached a peak (0.95) when four clusters were used for this data.

The four identified clusters (A - D) and their associated impact parameters are shown in Table 6.2. The mean of each cluster informs test conditions that a mechanical test design should be able to replicate. It is thought that each cluster characterises a different movement solution.

Table 6.2: Results of the stamping impacts (mean ± standard deviation) grouped into four clusters.

Participant *Cluster*	Stud Energy (J)	Stud Mass (kg)	Inbound velocity (m/s)	Inbound velocity angle (°)	Stud pitch angle (°)
1	6.0 ± 1.9	1.4 ± 0.4	2.9 ± 0.6	34 ± 12	-4 ± 5
2	6.1 ± 3.0	1.5 ± 0.7	2.8 ± 0.4	25 ± 14	18 ± 5
cluster A	*6.0*	*1.5 ± 0.6*	*2.9 ± 0.5*	*29 ± 13*	*7 ± 12*
3	8.2 ± 1.5	0.7 ± 0.2	4.9 ± 0.6	60 ± 6	27 ± 5
4	9.1 ± 1.9	0.8 ± 0.2	4.8 ± 0.5	36 ± 4	8 ± 5
cluster B	*8.7*	*0.8 ± 0.2*	*4.8± 0.5*	*48 ± 13*	*18 ± 11*
5	11.0 ± 4.2	1.8 ± 0.5	3.5 ± 0.6	51 ± 12	-4 ± 4
6	11.0 ± 4.3	1.6 ± 0.6	3.9 ± 1.0	38 ± 9	26 ± 6
cluster C	*11.0*	*1.7 ± 0.5*	*3.7 ± 0.8*	*45 ± 13*	*11 ± 16*
7	12.0 ± 3.3	0.9 ± 0.4	5.3 ± 0.9	47 ± 8	2 ± 8
8	12.0 ± 2.4	0.9 ± 0.2	5.4 ± 0.6	62 ± 5	3 ± 9
cluster D	*12.0*	*0.9 ± 0.3*	*5.4 ± 0.7*	*54.2 ± 10*	*2 ± 8*

6.4.2 Raking phase

To identify test parameters from the raking motion for the second phase of the test method, a representative trial was used. Using a representative trial omitted the need for manually identifying the shoe marker positions of all trials for the full duration of stud-ATD contact. Identification of the representative trial was based on stud force data (all trials), filtered with a 4th order bi-directional low-pass Butterworth filter (cut-off frequency: 50Hz) and time-synchronised based on a force threshold (30 N). Figure 6.3 shows the stud force data, split up into an initial impact phase and subsequent raking phase. The initial impact phase was defined as the first 50 ms post impact, which is similar to the impact phase of runners described by Nigg (2010). The raking phase was defined as the following 80 ms.

The mean stud force of all time-synchronised trials during the raking phase was calculated. A trial with mean stud force during raking phase closest to mean stud force during raking of all trials was chosen as the representative trial. Mean (± SD) stud force of all trials during the raking phase was 137.6 ± 39.0 N; mean stud force of participant 4, trial 9 was 136.8 ± 13.5 N. This trial was selected as the representative trial (Figure 6.3).

The lateral front stud shoe marker of the representative trial was manually identified from high speed video footage for 200 frames after first impact, giving a velocity trace for 200 ms. At the end of the raking phase (t = 130 ms), a horizontal foot velocity of

0.93 m/s was reached. The vertical velocity stayed approximately zero after initial impact. Foot displacement during the raking phase was 52 mm. The developed test method should therefore aim to accelerate the stud to a horizontal end velocity of 0.9 m/s whilst travelling at least 52 mm.

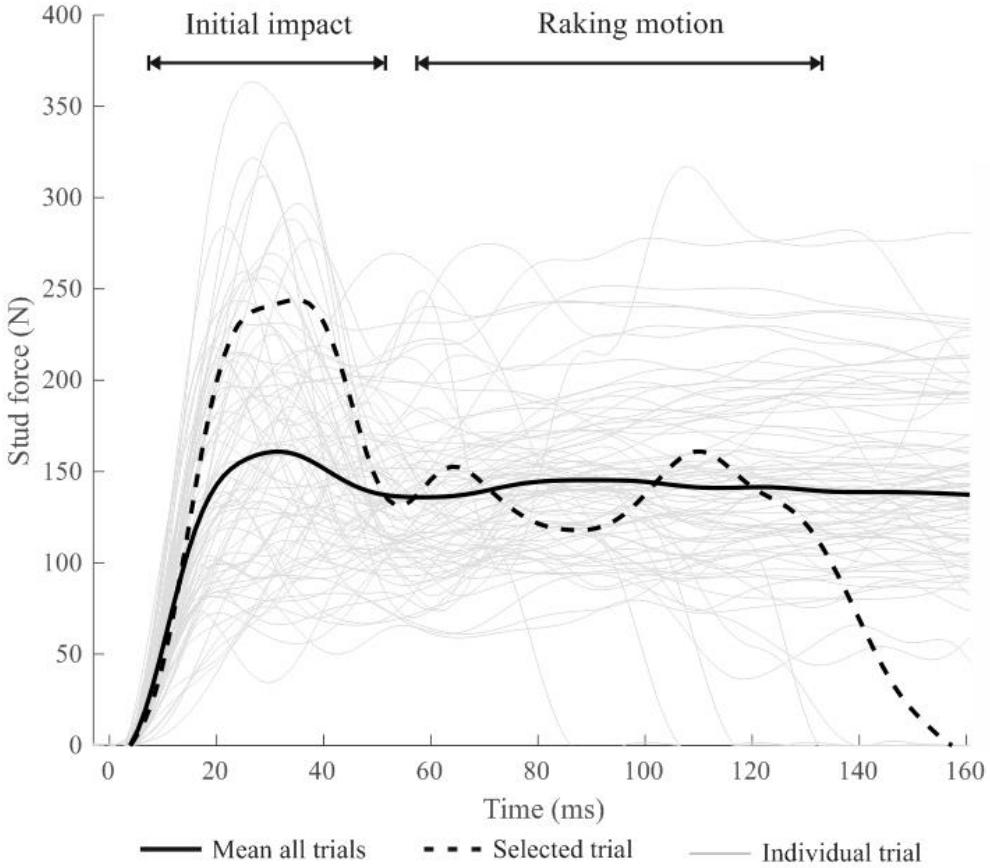


Figure 6.3: Selection of a representative trial. Average raking phase of selected trial (136.8 ± 13.5 N) is similar to average of all raking phases (137.6 ± 39.0 N).

6.4.3 Product design criteria

The 32-element model of Pugh (1991) was used to identify the product design criteria for a new test method. A full overview of the 32 elements that were considered can be found in Appendix C. Table 6.3 shows the seventeen primary design criteria and six secondary design criteria that were identified for the test rig design. These design criteria were used to evaluate a range of design solutions. Primary design criteria 1 - 7 were based on the findings of the study described in Chapter 5 and the following ten primary design criteria followed from Pugh's 32-element model. A design decision was made by scoring each a design on a 1 - 5 scale for each criterion, where a score of 1 means it would not satisfy the criterion, and a score of 5 fully satisfies the criterion.

Design decisions were based on the summed score of the primary design criteria, and the secondary design criteria were used only when an equal score between two designs was obtained at the primary criteria.

Table 6.3: Primary and secondary design criteria of the product design specification.

Number	Description	Parameters
	Primary Design Criteria	
1	Should replicate the inbound speeds from identified clusters	2.9 - 5.4 m/s
2	Should replicate the inbound velocity angles from identified clusters	30° - 60°
3	Should replicate the impact mass calculated from identified clusters	0.8 - 1.7 kg
4	Should replicate the pitch angle from identified clusters	-10° - +30°
5	Should incorporate a raking phase, with end velocity of stud:	~ 0.9 m/s
6	Sliding distance during the raking phase should be:	≥ 52 mm
7	Should replicate ± 2 SD of stud force during the raking phase	60 - 215 N
8	All impact parameter settings should be repeatable and accurate	
9	Needs to easily mount a variety of studs (e.g. a screw thread)	
10	Able to mount skin and soft tissue simulant in a suitable way	
11	Easy to change the skin and soft tissue simulant after each trial	
12	The test rig should not pose a risk to the investigator	
13	The test rig needs to be designed and built within 6 months	
14	Material costs cannot exceed £1500	
15	The height of the test rig cannot exceed normal ceiling height (2.4 m)	
16	The test rig should have a minimum life in service of 5 years	
17	The design should reproducible by other test centres	
	Secondary Design Criteria	
1	Portability: possible to move between buildings / laboratories	
2	Test outcome is unambiguous and easy to interpret	
3	Utilising readily available machines; such as standard pendulums, Instron© machines or a drop hammer	
4	Requires low maintenance and little to no recalibrating	
5	The test rig has an ergonomic, user friendly design	

6.5 Conceptual designs

6.5.1 One-phase designs

Idea generating sessions produced three viable design solutions: a drop hammer with moving impact surface, a powered impactor with moving impact surface, and a double-hinged pendulum with fixed impact surface (Figure 6.4). It was hypothesised that the moving impact surface would move backwards (i.e. left in Figure 6.4) to allow for a sliding movement of the stud after initial impact. In the double hinged pendulum design solution, it was hypothesised that the inertia after initial impact would result in a raking phase moving up along the stationary impact surface.

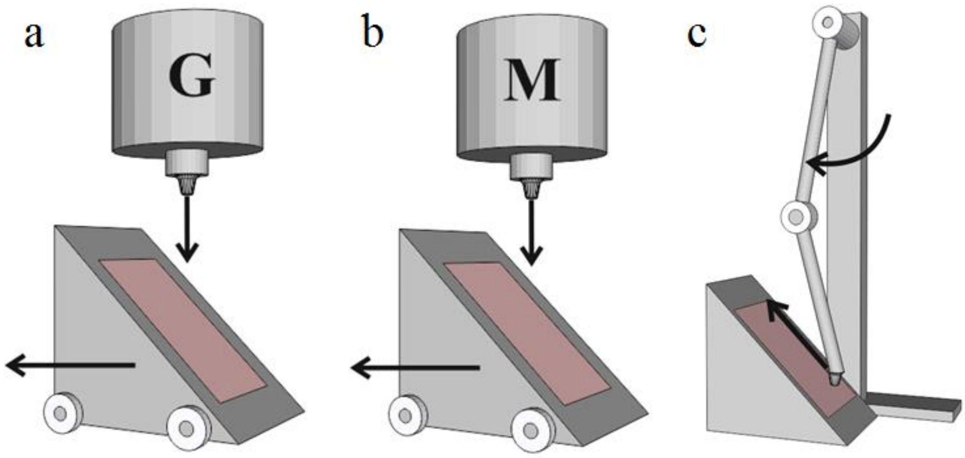


Figure 6.4: Three design solutions were considered, a) drop hammer (solution 1); b) powered impactor (solution 2); and c) double-hinged pendulum (solution 3). G indicates gravity driven, M indicates driven by a motor.

The decision matrix presented in Table 6.4 was used to identify which design solution complied best with the primary design criteria as were previously outlined in Table 6.3. In Table 6.4, each design is rated on a 1 - 5 score rating for its feasibility to meet the design criteria, where a score of 1 means it would not satisfy the criterion, and 5 fully satisfies the criterion.

Table 6.4: Design criteria summarised in demands and wishes with a 1-5 score rating.

	Solution 1: Drop hammer	**Solution 2:** Powered impactor	**Solution 3:** Double-hinged pendulum
Primary design criteria			
1 Inbound speed	4	3	3
2 Inbound velocity angle	5	5	3
3 Impact mass	5	5	5
4 Stud pitch angle	4	4	3
5 Raking velocity	3	5	2
6 Sliding distance	4	4	2
7 Raking force	3	4	1
8 Settings repeatable & accurate	5	5	4
9 Simple stud mount	5	5	5
10 Skin simulant mounting	5	5	5
11 Changing skin simulants	4	4	4
12 Safe to use for the operator	4	4	3
13 Build within 6 months	5	3	3
14 Material costs under £1500	5	1	3
15 Height under 2.4 m	3	5	5
16 Service life of 5+ years	4	4	4
17 Repeatable design	5	2	4
Total	*73*	*68*	*59*
Secondary design criteria			
1 Portability	3	3	5
2 Simple outcome variable	3	3	3
3 Readily available machines	5	1	3
4 Low maintenance	4	2	3
5 Ergonomic design	3	5	3
Total	*18*	*14*	*17*

Design solution 1, the drop hammer, scored highest on both the primary and secondary product design criteria. This design had the potential to meet most of the design criteria outlined in Table 6.3 - nevertheless, the impact mass and raking force could potentially not be controlled separately. The double-hinged pendulum design would also not be able to control impact mass and raking force separately. Additionally, the double-hinged pendulum design was likely to result in a low raking velocity. The powered impactor design solution exceeded the budget for this project.

Pilot test

A pilot version of design solution 1 was realised and an experiment to review its feasibility and performance was conducted. An existing drop hammer with options of 1.7 and 5.0 kg drop mass, combined with a 13 cm drop height (~1.6 m/s inbound velocity) was used. The inbound velocity was kept lower than required for a finalised test design in order to protect the guide rails of the existing drop hammer. An extended stud with a total length of 50 mm was used to impact a wedge-shaped block (solid aluminium). The extended stud length was necessary to impact the inclined surface (Figure 6.5a). The impact surface was angled at 47° from horizontal (mean inbound velocity angle, Table 5.1) and the block had low-friction wheels that allowed it to easily roll backwards (Figure 6.5). A variety of skin simulant and soft tissue simulant materials were attached to the wedge. All impacts were filmed with a high-speed camera. It was anticipated that upon impact, the wedge and simulant would roll backwards and allow for a sliding (raking) movement of the stud over the simulant material.

The first tests showed that the wedge and simulant rolled backwards as anticipated. However, there was no sliding of the stud over the simulant material because the stud and impact mass bounced up after initial impact (Figure 6.5).

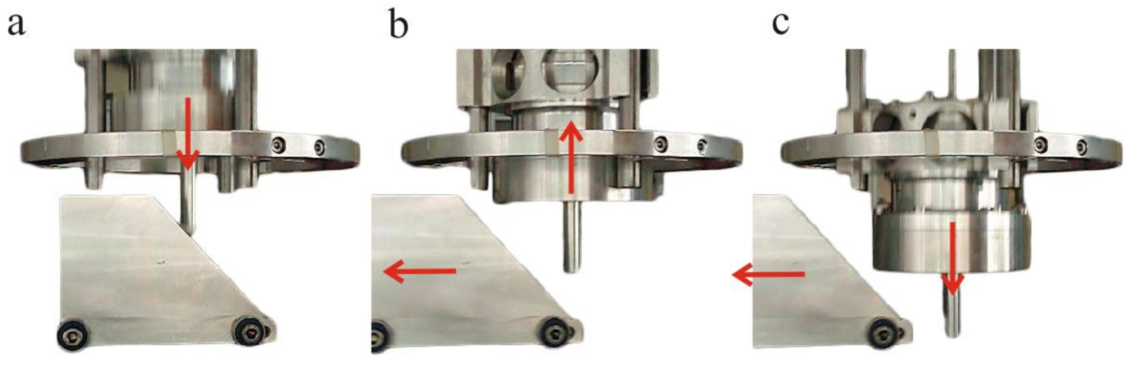


Figure 6.5: Bouncing occurred during the pilot test where a stud was attached to a drop hammer impacting an inclined surface. Red arrows indicate movement direction. a) Initial impact; b) bounce after impact, wedge moving back; and c) drop hammer coming down on table after the bounce.

Subsequently, nine different tests were conducted, with three strategies to inhibit the stud from bouncing; 1) changing the simulant material, 2) reducing friction between the stud and simulant material and 3) increase the drop mass from 1.7 to 5.0 kg. An interaction of these test parameters were used to evaluate their effect on inhibiting bouncing of the stud. The drop height was not increased above the stated 13 cm because the high-speed footage already showed horizontal movement of the guide rails at the current height due to the large transverse forces being generated. The combinations of tests parameters and their results are stated in Table 6.5.

Table 6.5: Results of the pilot test with varying input conditions.

Test	**Weight** (kg)	**Height** (m)	**Skin simulant**	**Soft tissue simulant**	**Friction reduction**	**Bouncing behaviour**
1	1.7	0.13	None	Silastic 3483	None	Single bounce
2	5.0	0.13	None	Silastic 3483	None	Single bounce
3	1.7	0.13	Silskin 10	None	None	Multiple bounces
4	5.0	0.13	Silskin 10	None	None	Single bounce
5	1.7	0.13	Silskin 10	Silastic 3483	Silicone spray	Single bounce
6	1.7	0.13	Silskin 10	Silastic 3483	Soap	Single bounce
7	1.7	0.13	Silskin 10	None	Soap	Single bounce
8	1.7	0.13	Chamois dry	Silastic 3483	None	Single bounce
9	1.7	0.13	Chamois wet	Silastic 3483	None	Single bounce

The pilot test showed that the drop hammer solution did not appropriately transition from the initial impact to the raking phase within this single test design. The directional change during the stamping motion (Figure 6.2) could not accurately be replicated in a single movement. Further, the drop hammer design solution was unable to control the raking force and raking velocity separately from the impact mass and inbound velocity. Finally, the drop hammer that was used in this experiment was unable to sustain the horizontal forces resulting from the off-axis loading caused by an inclined impact surface. A two-phase test design, splitting the initial impact and raking phase was therefore considered in the next design stage.

6.5.2 Two-phase designs: initial impact

The pilot study in Section 6.5.1 showed that replicating an initial impact phase and a subsequent raking phase of the stud was ineffective within a single test design. The two-phase test design splits the test into the 'initial impact phase' and a subsequent 'raking phase', as previously discussed (Figure 6.2). The test parameters of the initial impact phase were defined in Chapter 5 and Section 6.4.1. The design criteria previously identified in Table 6.3 were used for the initial impact test design, with the exception of primary criteria 5 - 7 which only apply to the raking phase.

No further drop hammer solutions were considered in this phase since the pilot tests showed that the off-axis loading was likely to damage the bearings in the guide rails. Figure 6.6 shows the three design solutions that were considered for replicating the initial impact phase. All three solutions were able to control the inbound speed by changing the release height of the pendulum arm. The stud angle was adjusted using a circular stud attachment, and impact mass could be controlled by adding weights to the stud attachment. Inbound velocity angle was adjusted in three different ways: 1) by changing the pivot point of the pendulum arm, 2) by adjusting the height of the simulant mount, and 3) by changing the angle of the simulant mount (Figure 6.6).

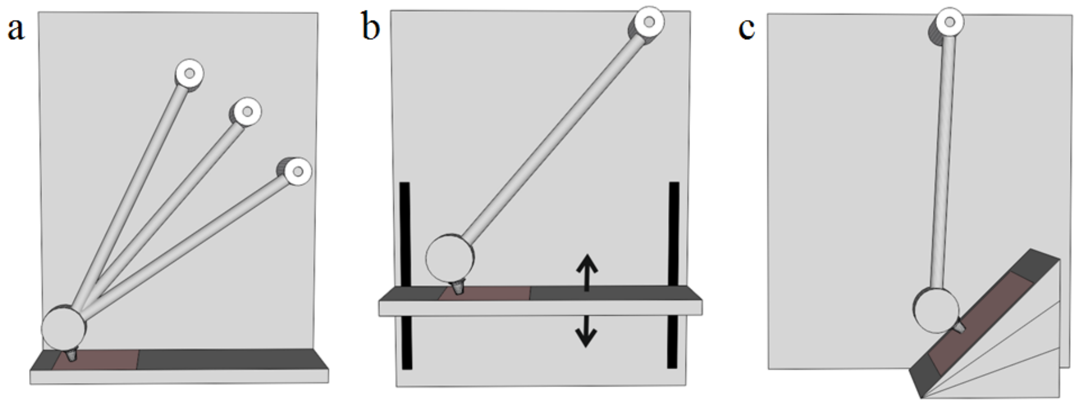


Figure 6.6: Three design solutions were considered, a) changing pivot point (solution 1); b) raised platform (solution 2); and c) inclined surface (solution 3).

Table 6.6 shows the design matrix that was used to identify which design solution complied best with the primary design criteria. Each design was again scored on a 1 - 5 score rating for its feasibility to meet the design criteria, where a score of 1 means it would not satisfy the criterion, and 5 fully satisfies the criterion.

Table 6.6: Primary and secondary design criteria of the initial impact phase summarised in a decision matrix with 1 - 5 score rating.

	Solution 1: Pivot point	**Solution 2:** Raising platform	**Solution 3:** Inclined surface
Primary design criteria			
1 Inbound speed	5	5	5
2 Inbound velocity angle	3	5	3
3 Impact mass	4	4	4
4 Stud pitch angle	4	4	4
8 Settings repeatable & accurate	5	3	4
9 Simple stud mount	5	5	5
10 Skin simulant mounting	5	5	3
11 Changing skin simulants	5	5	4
12 Safe to use for the operator	5	4	5
13 Build within 6 months	4	3	4
14 Material costs under £1500	4	3	5
15 Height under 2.4 m	4	4	5
16 Service life of 5+ years	5	5	5
17 Repeatable design	5	4	5
Total	*63*	*59*	*61*
Secondary design criteria			
1 Portability	5	5	5
2 Simple outcome variable	3	3	3
3 Readily available machines	3	3	3
4 Low maintenance	5	3	5
5 Ergonomic design	5	3	5
Total	*21*	*17*	*21*

The pendulum with changing pivot points (design solution 1) scored highest on the primary product design criteria. The design provides a simple solution to adjusting the inbound velocity angle by moving the pivot point of the pendulum between three pre-set points. This design was further developed. The other two designs also provided viable solutions for changing the velocity angle in the initial impact test. Design solution 2 (the raised platform) was thought to be more complex and less cost effective than design 1. In design solution 3 (the inclined surface), mounting the skin simulant would have been more challenging than in design 1.

6.5.3 Two-phase designs: raking phase

The second test in the two-phase test design replicates the 'raking phase' of stamping impacts (Figure 6.2). The test parameters of the raking phase were defined in Section 6.4.2. The design criteria previously identified in Table 6.3 were used for the raking phase test design, with the exception of primary criteria 1 - 3 which only apply to the initial impact phase.

Figure 6.7 shows three design solutions that were identified for simulating the raking phase of stamping impacts. In these raking tests, a stud is placed at the impact location of the initial impact test. A constant load is applied to the stud through an adjustable mass to replicate the stud raking force, within 2 SD from the mean raking force identified in Section 6.4.2. The stud angle is set at the same value as the initial impact angle. Instead of moving the stud over the simulant, all three design solutions move the skin simulant tray whilst the stud remains stationary. Accelerating the stud with its additional mass that is necessary to provide the raking force would require more force than accelerating the skin simulant tray. The three design solutions differ in the method of accelerating the skin simulant tray to the raking end velocity; either by 1) pulling the simulant tray using gravity, 2) pulling the simulant tray with a tuneable spring-damper system, or 3) by pulling the simulant tray with a motorised solution (Figure 6.7).

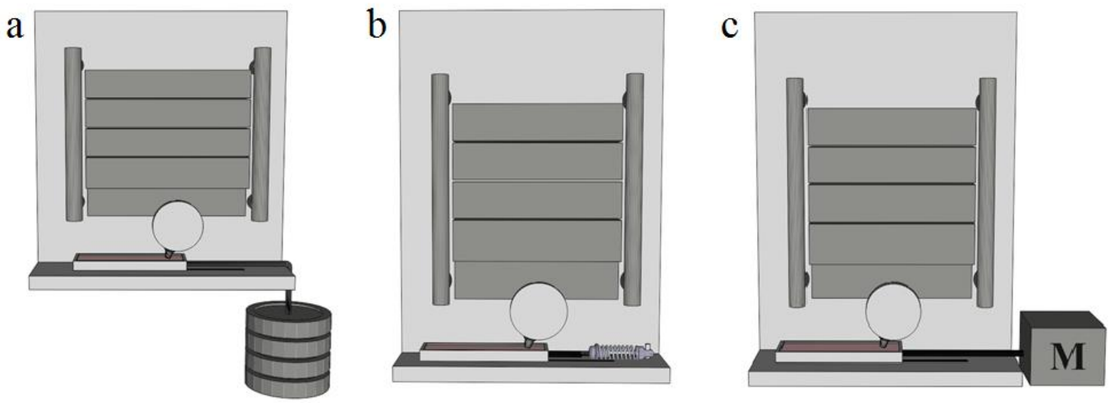


Figure 6.7: Three design solutions were considered for the raking phase test, a) pulling weights (solution 1); b) tuneable spring-damper system (solution 2); and c) motorised movement (solution 3).

Again, the three potential design solutions were tested against the primary and secondary design criteria set out in Table 6.3. The results of this decision matrix are shown in Table 6.7. The scores in this matrix again range from 1 - 5, where 1 means it would not satisfy the criterion, and 5 fully satisfies the criterion.

Table 6.7: Primary and secondary design criteria of the initial impact phase summarised in a decision matrix with 1-5 score rating.

	Solution 1: Pulling weight	**Solution 2:** Spring - damper	**Solution 3:** Motor driven
Primary design criteria			
4 Stud pitch angle	3	3	3
5 Raking velocity	2	4	5
6 Sliding distance	3	3	5
7 Raking force	5	5	5
8 Settings repeatable & accurate	4	4	3
9 Simple stud mount	5	5	5
10 Skin simulant mounting	5	5	5
11 Changing skin simulants	5	5	5
12 Safe to use for the operator	3	3	4
13 Build within 6 months	4	4	4
14 Material costs under £1500	5	5	1
15 Height under 2.4 m	5	5	5
16 Service life of 5+ years	5	5	3
17 Repeatable design	5	5	1
Total	*59*	*61*	*54*
Secondary design criteria			
1 Portability	5	5	5
2 Simple outcome variable	4	4	4
3 Readily available machines	1	1	1
4 Low maintenance	5	5	2
5 Ergonomic design	3	4	4
Total	*18*	*19*	*16*

The tuneable spring-damper system (design solution 2) scored highest on both the primary and secondary product design criteria. This design is likely to provide control over the raking end velocity of the skin simulant tray, whilst keeping manufacturing costs to a minimum. The spring-damper design solution was further developed.

6.6 Finalised design

Based on the conceptual design choices presented in Table 6.6 and Table 6.7, a mechanical test rig consisting of two sequential test phases was developed. In the developed test method, the stud is first placed on the pendulum test and impacts the skin simulant (phase 1, initial impact). Then, both the stud and the skin simulant tray are moved to the second phase (raking phase). At the start of this test, the stud should be aligned with any potential damage from the initial impact test. The stud is then dragged over the skin simulant after which skin damage was be evaluated.

6.6.1 Initial impact test

The pendulum design outlined in Table 6.6 was realised. The final design can adjust the four required impact parameters separately:

1. Inbound velocity, adjusted by release height, ranging from 0 - 5.4 m/s.
2. Inbound velocity angle, adjusted by changing the pivot point, at three options: 30°, 45° and 60°.
3. Impact mass, adjusted by weights at 0.2 kg intervals, range: 1.0 - 2.2 kg.
4. Stud angle, adjusted by rotating stud attachment at 5° intervals, range: -10° - +30°.

The final design of the initial impact phase was developed as a computer-aided design (CAD) model before manufacturing (Figure 6.8, SketchUp Pro 2017, Trimble Inc, Sunyvale, CA, USA).

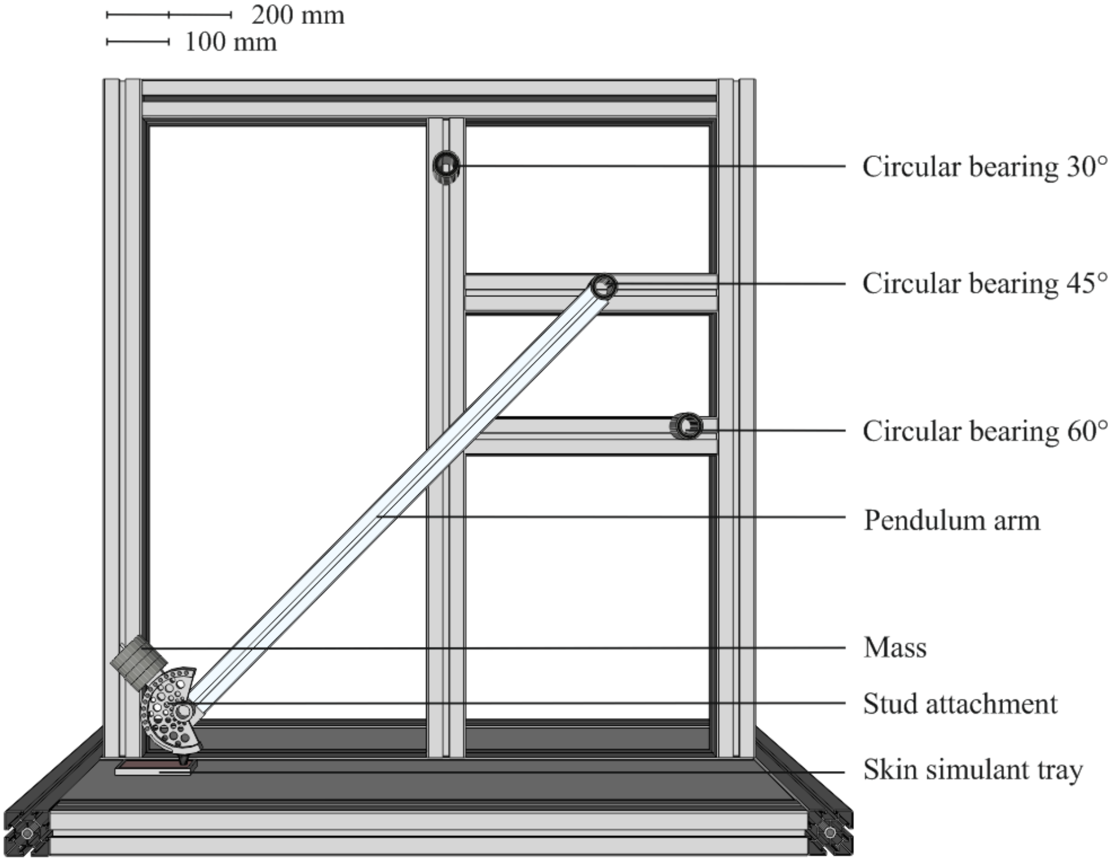


Figure 6.8: CAD model of initial impact test in phase 1.

The pendulum arm length required to obtain up to 5.4 m/s inbound velocities was calculated using Equation (6.1) for the conservation of kinetic and potential energy (not taking friction into account):

$$\frac{1}{2}mv_s^2 + mgh_s = \frac{1}{2}mv_i^2 + mgh_i \tag{6.1}$$

Where m is the impact mass in kg, v_s is the starting velocity in m/s, g is the gravitational constant in m/s^2, h_s is the starting height in m, v_i is the impact velocity in m/s and h_i is the height at impact in m. Equation (6.1) showed that the release height of the pendulum has to be at least 1.49 m to achieve a 5.4 m/s impact velocity. Therefore, the pendulum arm needed to be at least 0.99 m, to achieve 1.49 drop height with a 60° inbound velocity angle (Figure 6.8).

Impact mass was estimated using half of the mass of the pendulum arm added to the mass of the stud attachment. The mass of the pendulum arm was calculated using its cross-sectional area, length and material density (aluminium). The mass of the stud attachment was estimated using its predicted radius, thickness and material density (steel).

The most challenging part of the pendulum design was realising a lightweight and durable rotating stud attachment for 5° stud angle intervals. The lower bound of the target impact mass was 0.8 kg. The minimum radius of a circular stud attachment design which allows 5° stud angle intervals was 126 mm. At 20 mm thickness and formed from steel, this design would have weighed approximately 7.8 kg. Instead, a design was constructed where 13 bolt holes were drilled at 10° intervals (Figure 6.9), halving the required radius. The stud could be screwed in at two points which were 15° apart. A combination of stud attachment rotation and stud mount gives 5° intervals. Further mass reduction was achieved by removing material from the inner parts of the disc as shown in Figure 6.9.

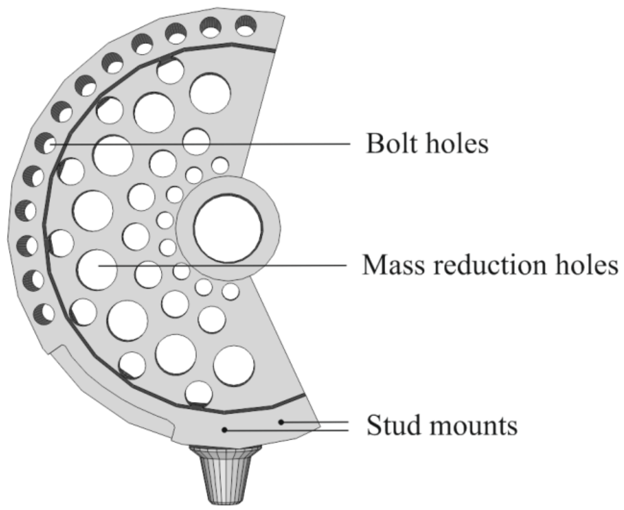


Figure 6.9: Close-up view of the stud attachment with 5° interval stud angle options.

The finalised stud attachment weighed 0.44 kg. Besides the stud attachment, part of the pendulum arm also has to be added to the impact mass. The force exerted at the stud when the arm is held in a horizontal position was identified as the effective impact mass and measured using portable scales (EK-600G, A&D Instruments, Abingdon, UK). The minimum effective impact mass was 1.0 kg (9.8 N).

During the building phase of the pendulum test, a combination of aluminium, brass and steel materials were used depending on expected forces and torques during the pendulum arm swing, and the material density required. The final test set-up for phase 1 is shown in Figure 6.10. The outer frame was build using aluminium structural framing (Rexroth, Bosch Group, Charlotte, NC, USA), allowing flexibility for potential future adjustments to the design.

Figure 6.10: End result of the initial impact phase of the developed test method.

A brief description on how to operate the initial impact test including more detailed photos of the final design is given in Appendix D.

6.6.2 Raking test

The spring-damper solution outlined in Table 6.7 was realised. The final design can adjust the three required impact parameters separately:

1. Raking force, adjusted by weights at 4 kg intervals, range: 6.0 – 22.0 kg.

2. Stud angle, adjusted by rotating stud attachment at 5° intervals, range: -10° - +30°.
3. Raking end velocity, adjusted by removing or adding springs and opening or closing pneumatic damper. Depending on friction, ranges approx. 0.2 – 1.5 m/s.

A CAD model of the final design was created to determine its dimensions (Figure 6.11, SketchUp Pro 2017). A theoretical model was developed to estimate the force required to accelerate the simulant tray to 0.9 m/s. This pulling force is dependent on the normal load exerted on the stud (stud mass, range: 6.0 - 22.0 kg), the distance travelled, the friction between the stud and the skin simulant, the mass of the accelerating simulant tray and the friction on the bearings.

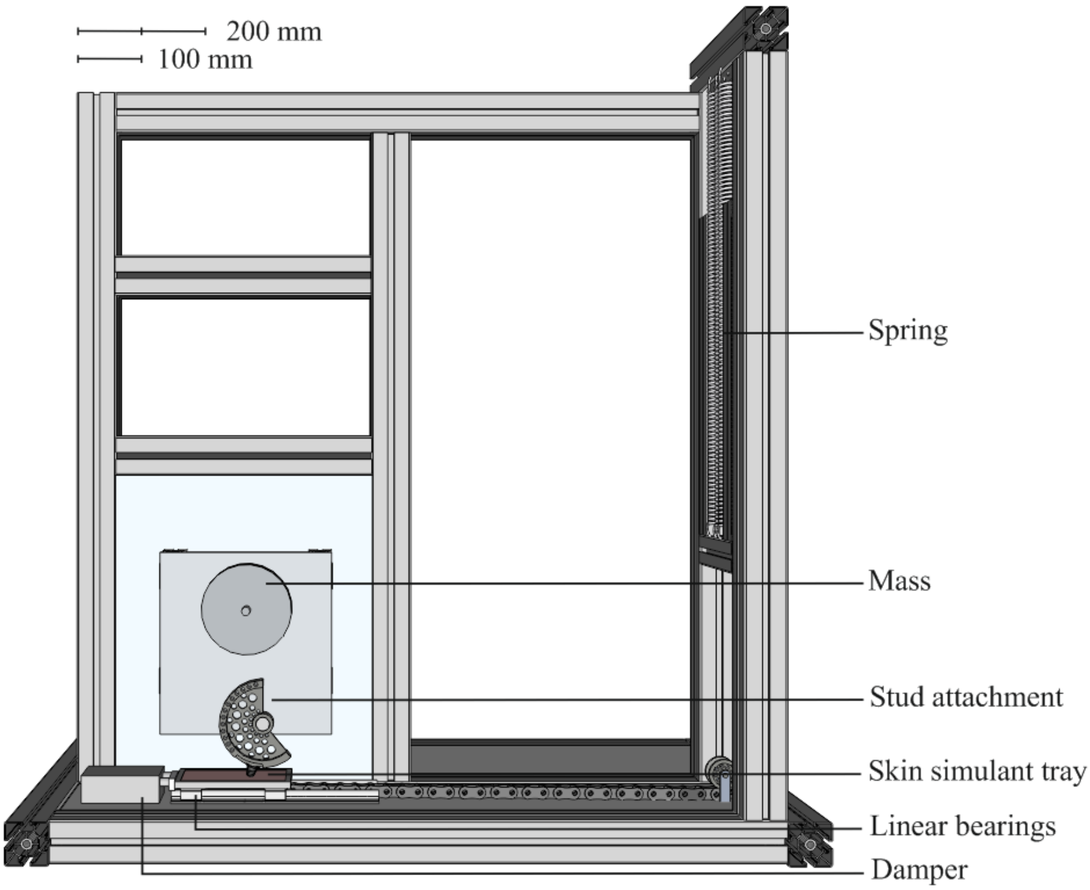


Figure 6.11: CAD model of raking test in phase 2.

During the building phase of the raking test, a combination of aluminium, cast iron and steel materials were used depending on the expected force and the material density that was required. The final test set-up for the raking phase is shown in Figure 6.12. An electrical unit was added to the final design that pre-tenses the springs and lifts the stud mass, in order to comply with health & safety requirements and to improve the ergonomics for the test operator. The final test design can mount 10 springs and one damper. Each spring has a spring constant (*k*) of 280 N/m.

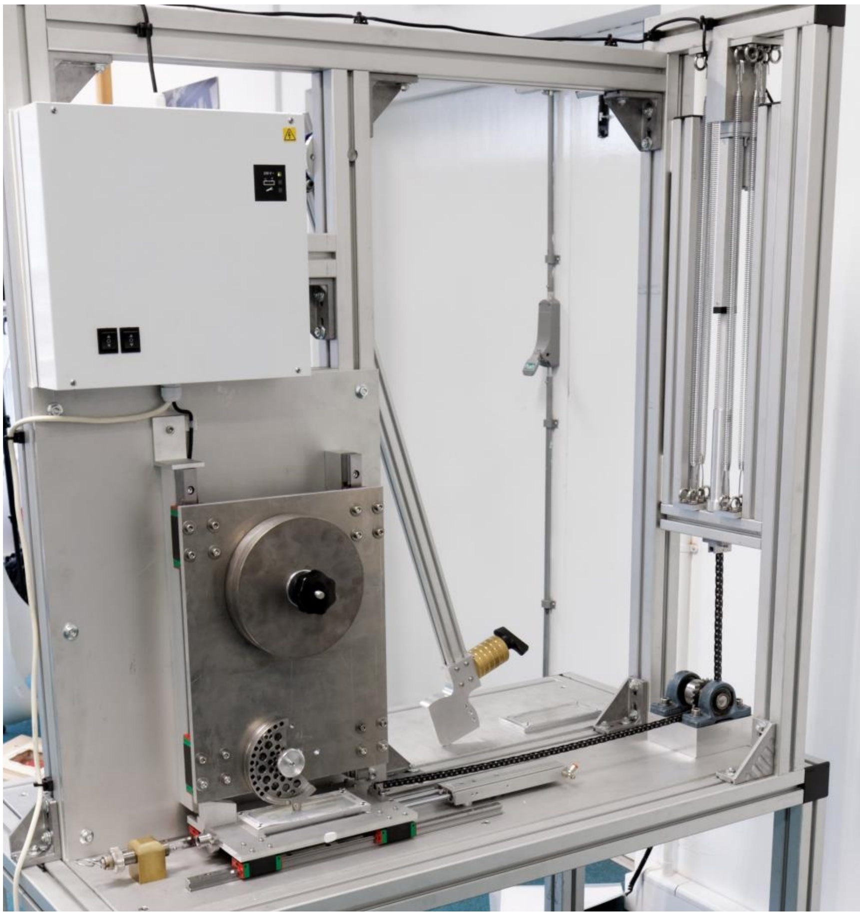


Figure 6.12: Finalised test set-up for the raking phase of the developed test method.

The spring-damper system can be tuned by removing and adding springs, and by opening and closing the valves on the damper. High speed video footage was used to measure the end velocity of the simulant tray at various settings. When using synthetic chamois as a skin simulant material at 6 kg stud mass, the system reached a velocity of 1.5 m/s. Lower velocities were achieved by increasing the damping, decreasing the number of springs and / or increasing the raking force and thus the friction force between stud and simulant.

A brief description on how to operate the raking test including more detailed photos of the final design is given in Appendix E.

6.7 Discussion

A two-phase mechanical test rig was designed based on kinetic and kinematic parameters measured during the study described in Chapter 5. In the user needs it was identified that the reproducibility of a newly developed test method was important to World Rugby, as was developing a test method which can be used as an overarching standard for football and rugby.

The design decisions in this chapter have been guided by a decision matrix with 17 primary criteria and five secondary criteria. Decisions were based on the final score of the primary criteria; the secondary criteria were only used when the primary design criteria did not give a conclusive decision. All primary design criteria were thought to be vital to the successful implementation of the test method, therefore no weighing factor was assigned to the different criteria. Giving different weighing factors to the primary design criteria could have influenced the final decision matrix outcomes.

The developed mechanical test is a simplification of the observed stamping movement; the developed test needs to be replicable in such a way that other test centres could manufacture it as well. In the first conceptual designs, a test method where the initial impact and raking phase was tested in one continuous movement was explored. Compared to a two-phase approach, the single phase test method would provide a more realistic impact to the skin simulant. The removing and repositioning of the simulant and stud between phases could excite a different response from the skin simulant material compared to a continuous test. However, developing a two-phase test method allowed greater control over the test parameters in each of the test phases and improved the flexibility and accuracy of the design.

The first phase of the developed method replicated the inbound velocity (magnitude and direction), stud pitch angle and effective mass that were observed in simulated stamping impacts. The roll angle of the shoe was not accounted for in the developed test method, nor was the time-to-peak force replicated in the initial impact phase. Implementing these impact parameters in a durable and reproducible manner was not deemed feasible within the restrictions of this project. Future versions of the developed test method could incorporate these impact parameters to improve its biofidelity.

The effective mass of the initial impact phase was calculated using the force-time integral (Neto *et al.* 2012). The term effective mass accounts for the stiffness in the joints of the entire kinetic chain involved (Lenetsky *et al.* 2015), though the stiffness in the developed test method was not subsequently controlled. Not controlling for the stiffness of the developed test method was likely to lead to a shorter contact time and subsequent higher peak forces in the developed test method compared to the observed stamping impacts, if inbound velocity and impact mass are kept constant. Further, the forces measured by the pressure sensors in Chapter 5 produced on average 7.5% error in the calculation of peak impact force (Oudshoorn *et al.* 2016b) and this error can have subsequently influenced the effective mass results. The shortage of literature on stud-skin impacts means that the impact parameters that were selected for the test method design could not be usefully compared to other studies.

The initial impact phase can replicate an effective mass range of 1.0 - 2.2 kg. The 1.0 kg lower limit of the designed test method is 0.2 kg higher than the mean effective mass of cluster B in Table 6.2. The impact mass in phase 1 consists of the stud attachment, part of the pendulum arm, stud attachment mount and a mount to add mass. The pendulum arm was made from lightweight aluminium structural framing (Rexroth, Bosch Group) measuring 20 x 40 mm. Lowering the mass in the pendulum arm through a smaller framing size was thought to make the design less durable. The minimum radius of the stud attachment was determined by the required 5° stud angle intervals. A solution where the stud attachment could be changed at 10° intervals combined with two screw-in mounts for the stud at 15° intervals reduced the required diameter from 252 mm to 126 mm, saving 73% of mass (based on solid disc mass). The mass of the stud attachment was further reduced by removing material from its middle section (Figure 6.9). The mass of the final design of the stud attachment was 0.44 kg; 1.52 kg lighter than a solid disc of that diameter and thickness, from the same material. Future improvements of the test design could reduce the mass of the stud attachment further by lowering the stud angle spacing or by using a fixed stud angle.

The second phase of the developed method replicated the sliding velocity and minimum sliding distance as observed in a selected representative trial of the stamping study. The test method was designed to replicate the force exerted on the skin surface within the mean ± 2 SD of all the observed raking trials. The pitch angle during the raking phase was kept consistent with the pitch angle during the initial impact phase. Future biomechanics studies of stud laceration impacts should consider measuring the stud angle directly instead of inferring it from shoe-markers, as this would provide information on the stud angle during the raking phase. The end raking velocity for this test method was based on a single trial, acting as a representative trial for this test phase. This approach meant that no variation in end velocity was identified from the biomechanics study, though the current test design can replicate velocities between 0.2 – 1.5 m/s (depending on stud-simulant friction and raking force).

The developed test method is suitable to test studs in isolation. This way, the effect of each individual stud design can be investigated. For a test method assessing the laceration risk of full outsoles, a higher impact mass and raking force than is possible in the current set-up should be used. The study described in Chapter 5 provided sufficient information to identify test parameters for the development of a test method for full outsoles, if required in the future.

6.8 Chapter findings

The aim of this chapter was to design and build a test method replicating rugby stamping impacts that can be used in studded footwear regulations or standards. The new test method has to be repeatable, reproducible by other research institutes or test houses and ideally the test method is flexible enough to incorporate test parameters for other studded footwear sports as well. To replicate the rugby stamping movement as described in Chapter 5, a two-phase test method approach was developed. Three conceptual designs for each phase were considered. A decision matrix was used for choosing the designs which met most of the design criteria. The final design was based on four initial impact 'types' observed in the biomechanics study, and a representative trial during the raking phase. The developed test rig is designed to test studs in isolation and can at this stage not be used to evaluate the effect of full outsoles on their laceration injury risk. It is based on the assumption that the measured biomechanical parameters were representative of impacts causing lacerations in rugby game-scenarios. In the next chapter, the developed test rig is used to assess the laceration injury risk of six stud designs.

7 Laceration risk of standardised and commercial studs

7.1 Introduction

Chapter 6 described the design and development of a new test method for assessing laceration injury risk of individual stud designs. The outcome measure of this test method was not yet determined in the previous chapter. In World Rugby's current regulations, the damage a stud inflicts on a skin and soft tissue simulant should be compared to the damage resulting from the comparator stud (World Rugby 2015, Section 2.3.1). However, the type of skin and soft tissue simulant that should be used and how skin damage is assessed, is not specified within the regulations.

Skin and soft tissue simulants used in forensic and sports injury research have been discussed in Section 2.6. Forensic studies have previously used synthetic chamois, silicone rubbers and porcine skin as skin simulant materials when investigating mechanical failure of the skin (Bir *et al.* 2012; Parmar *et al.* 2012; Shergold *et al.* 2006). The feasibility of using these materials in combination with the developed test method needs to be investigated. In human skin, the dermal and epidermal layer of the skin combined is 1.93 - 2.35 mm thick across the arms, trunk and legs (Akkus 2012). The subcutaneous adipose layer in 63 relatively lean (mean BMI: 21.8), healthy young men across twelve standardised body locations ranged from 2.2 - 7.9 mm, with a mean of 4.9 mm (Jones et al. 1986). It has previously been outlined in Section 2.6.3 that the breaking load of skin is dependent on the hardness of its underlying adipose tissue; skin on bony parts breaks more easily than skin with a thicker subcutaneous layer (Bir et al. 2012; Jussila et al. 2005). The thickness of the soft tissue layer that is used in combination with the skin simulant layer ideally reflects the lower end of adipose thickness in the target group.

When defining wound severity, clinicians tend to use one of two methods: 1) measuring wound size (quantitative assessment of wound severity) or 2) by using a classification system (qualitative assessment of wound severity). Measuring wound size is often done by taking unidimensional measures such as wound depth, length and width (Goldman & Salcido 2002). These measures are known to have poor correlation with actual wound size for complex shapes (Bilgin & Güneş 2014; Goldman & Salcido 2002).

Classification systems are frequently used to define clinically important differences in wound severity (Payne & Martin 1990; Carville *et al.* 2007; van den Eijnde *et al.* 2014a). A good wound classification system provides clear injury definitions and example pictures (Payne & Martin 1993).

The new developed test method allows for a wide range of combinations of test settings. A variety of test setting combinations have been identified in Chapters 5 and 6. The final set of test parameters used in this test method should not only reflect rugby stamping impacts, but also aid in distinguishing between laceration injury risks of stud designs. The current chapter tests a small range of standardised and commercially available studs, showing how the test could be used in a standard or regulation.

7.2 Aim and objectives

The aim of this chapter is to show how the developed test method can be used to compare stud designs on their predicted risk of laceration. This chapter relates to stage F (*'Proof of concept'*) of the project structure described in Figure 1.3.

The aim is achieved through the following objectives:

1. Identify a suitable skin and soft tissue simulant for the test method.
2. Provide a relevant outcome measure of skin damage for the test method.
3. Show that the test method can effectively differentiate between the outcome measures of different studs.
4. Provide an example of how the method would work for commercially available studs.

7.3 Methods

7.3.1 Studs tested

The six studs that were used in this study are shown in Figure 7.1. Studs 1 - 4 are bespoke, standardised studs whilst studs 5 - 6 are commercially available, aluminium screw-in studs. Stud 1 is made from aluminium and has a 10 mm top diameter, 1 mm rounded top edge radius, and 15 mm length. It is based on current Regulation 12's 'comparator stud' (Figure 2.4), which has been widely agreed as presenting an acceptable risk of laceration injury. Stud 2 has the same materials and dimensions as stud 1, however it does not possess the 1 mm rounded top edge radius. Both studs 3 and 4 have a 5 mm top diameter and are 15 mm length. Stud 3 has a 1 mm rounded top edge radius, whilst stud 4 does not have the 1 mm rounded top edge radius. It was hypothesized that decreasing the top edge radius and decreasing the stud's top diameter would both result in an increased laceration injury risk. These standardised stud shapes are used to evaluate if the developed test method can effectively differentiate between controlled stud design changes.

Stud 5 is a commercially available stud which has previously passed BS 6366:2011 (British Standards 2011) and is commonly used in rugby shoes, especially those worn by forwards. In the questionnaire study presented in Chapter 4, this stud design was most prevalently used by rugby players. Stud 6 is a SmartPower© Profiler (ADN Sports, Lyon, France), a design which has entered the market early 2017 and has passed World Rugby's current studded footwear regulations. The introduction of the SmartPower© Profiler has caused controversy in the rugby community, where concerns were raised about its potentially increased laceration injury risk (McGeady 2017). It was hypothesized that stud 5 and stud 6 would result in a similar or lower risk of laceration injury than the comparator stud (stud 1).

Figure 7.1: Six studs selected for initial testing. From left to right: Stud 1 (comparator stud), stud 2 (standardised), stud 3 (standardised), stud 4 (standardised), stud 5 (commercial, kite-marked stud), stud 6 (new stud design).

7.3.2 Skin and soft tissue simulant

A pilot study informed the choice of skin simulant for the stud comparison tests. Four skin simulant materials that have previously been identified in Section 2.6 were used in the pilot; pig skin, two types of silicone rubbers and a synthetic chamois.

- The pig skin was sourced fresh from local butchers and was tested on the same day as slaughter. The skin was cut to 5.0 ± 1.0 mm thickness which left a small subcutaneous layer on the skin to act as a soft tissue layer.
- The first type of silicone rubber tested was Silskin 10 (MB Fibreglass, Newtownabbey, UK), prepared following manufacturer's instructions.
- The second silicone material was developed by the University of Alabama as a biofidelic skin simulant (Chanda *et al.* 2017).
- The synthetic chamois was cross-woven polyvinyl acetate (PVA) (KCIC200, Kent Car Care, Manchester, UK). The moisture content of each chamois sample was controlled to 0.53 mg of water per mm^2 (compared to dry weight). To achieve this moisture content, samples of known dry weight were soaked in water at room temperature (20º - 22º C) until saturation. Water was subsequently squeezed out by hand until the desired weight was reached.

A silicone soft tissue layer (Silastic 3483, Dow Corning Corporation, Auburn, MI, USA) of 4.6 ± 0.3 mm was added to the synthetic chamois and silicone rubbers to mimic underlying tissue. Silastic 3483 has a similar indentation response to human relaxed muscle tissue (Hrysomallis 2009). The thickness of the soft tissue simulant layer was similar to the subcutaneous adipose layer measured of healthy young men (Jones et al. 1986).

The pilot study testing pig skin confirmed the results from literature that this material has a poor repeatability (Lim et al. 2011; Parmar et al. 2012). In this pilot, belly skin, hind leg skin and back skin were tested. The skin damage was dependent on the body location of the tissue selected for each trial. Belly skin was softer and easier to break than skin sourced from other areas. Further, the pilot results showed that the skin damage also varied when keeping the skin region constant, concealing the effect of stud shape. Sourcing the pig skin from different pigs further decreased the test outcome consistency, and this masked the effect of stud shape or impact energy (Figure 7.2).

These results are in line with previous forensic studies, showing that synthetic simulants generate more reproducible results than biological materials (Carr & Wainwright 2011).

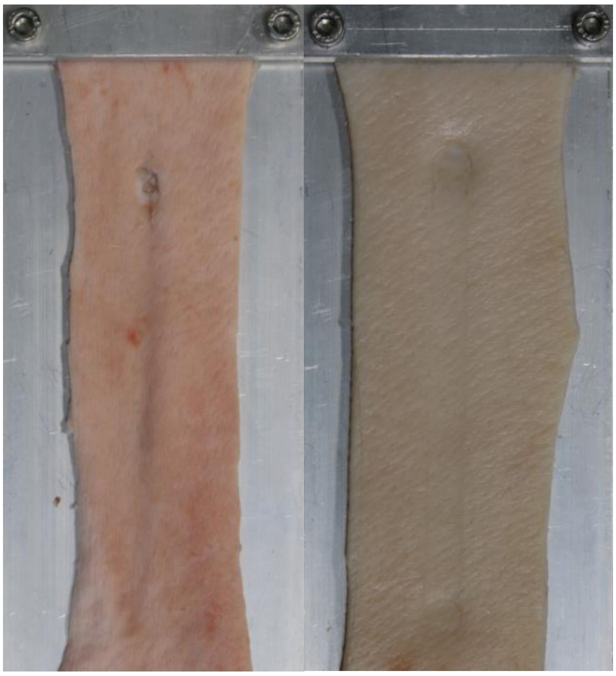


Figure 7.2: Two pig skin samples producing different results under constant impact conditions.

Figure 7.3 shows one of the silicone rubbers that were tested during the pilot study. During the raking test, both silicone skin materials stretched further than can be expected from human skin. This elasticity of the silicone skin simulants made them unsuitable for the raking phase of the test method. Further, both materials showed a low resistance again tear propagation, which is not representative of human skin tissue behaviour (Yang *et al.* 2015, Section 2.5.1).

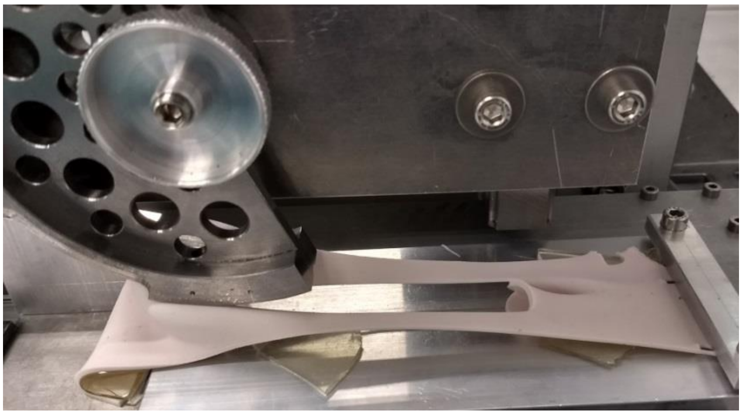


Figure 7.3: Silicone skin layers were highly elastic and showed low resistance against propagation of tearing.

The synthetic chamois material showed different skin damage responses when tested with the four standardised studs (studs 1 - 4), with a high consistency and lower elasticity than the silicone layers. The material resisted tear propagation, as can also be expected in human skin. Figure 7.4 shows an example of skin breakage whilst using synthetic chamois.

Figure 7.4: Moistened synthetic chamois provides a resistance against tearing even after initial penetration.

A summary of the materials tested and the findings regarding to their suitability are presented in Table 7.1. Synthetic chamois was found most suitable for assessing skin damage within the newly developed test method.

Table 7.1: The suitability of four potential skin simulant materials that were considered for the newly developed test method.

Material	Soft tissue material	Elasticity	Friction	Repeatability	Suitable
Silicone 1 (Silskin 10)	Silastic 3483	High	High	High	No
Silicone 2 (University of Alabama)	Silastic 3483	High	High	High	No
Pig skin (belly region)	Subcutaneous tissue	Low	Low	Low	No
Synthetic chamois	Silastic 3483	Low	Low	High	Yes

7.3.3 Test settings

Table 7.2 provides an overview of test settings used for the initial impact phase. The test parameters were set to replicate the mean of all stamping impacts (Table 5.1 and Table

5.2). On repeat tests, the impact energy was lowered by adjusting the inbound velocity until the comparator stud (stud 1, Figure 7.1) did not lacerate the skin and soft tissue simulant combination (damage category 1 or 2, specified in Section 7.3.4). All studs were tested at 1.5 J above and below this identified threshold, at 0.5 J intervals.

The raking phase was conducted after the initial impact phase with the same skin simulant. The stud was aligned with any potential damage from the initial impact phase (see Appendix E). Stud angle during the raking test was kept the same as the stud angle of the initial impact phase. The spring-damper system was set to pull the skin simulant tray at a speed similar to that of the representative trial (0.9 m/s, Section 6.4.2) when using the comparator stud. The raking force was lowered by adjusting the mass at 40 N intervals until the comparator stud did not cause a laceration. Subsequently, using this mass to provide the raking force, all six studs were tested. An overview of test settings used for the raking motion phase is shown in Table 7.2.

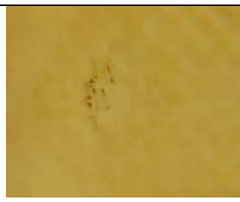
Table 7.2: Test settings of the initial impact and raking motion phases.

Phase 1: Initial impact		**Phase 2: Raking motion**	
Input parameters	**Settings**	**Input parameters**	**Settings**
Inbound speed	0.9 - 2.4 m/s	Stud mass	10 & 14 kg
Velocity angle	45°	Stud angle	+10°
Stud angle	+10°	End raking velocity	0.9 m/s
Impact mass	1.2 kg		

7.3.4 Analysis of skin damage

Based on wound results from pilot tests, a four-level categorical classification system was developed to assess wound damage resulting from the initial impact test. Table 7.3 states the definitions and shows examples of each damage category. In categories 1 and 2 no full tear in the skin simulant is found; such damage would not class as a laceration. Categories 3 and 4 show a full break of the skin simulant; this damage would be classified as a laceration. Each stud / impact energy combination is tested five times and the mode of the damage categories is used as the outcome.

Table 7.3: Categories for assessing skin damage resulting from test 1 (initial impact).

Category	Description	Example
1	No laceration, no visual damage	
2	No laceration, with visual damage	
3	Laceration, smaller than diameter of stud	
4	Laceration, equivalent or larger than diameter of stud	

Furthermore, a three-level categorical skin damage classification was developed for assessing potential additional damage after the raking test. Table 7.4 states the definitions and shows examples of each damage category. Categories 1 and 2 did not (further) lacerate the skin and would subsequently be classed as 'no laceration'. Category 3 caused (further) tearing of the skin simulant, and would be identified as a laceration. Again, each stud is tested five times and the mode damage category is the outcome measure.

The reliability of a new skin damage assessment system is an important feature of the test outcome, especially between test operators at potentially different facilities (Section 6.3, Table 6.1). Therefore, the proportion of agreement between four independent raters (inter-rater reliability) using the two developed classification systems was evaluated using Fleiss' kappa (κ) (Fleiss 1966).

Table 7.4: Categories for assessing skin damage resulting from test 2 (raking phase).

Category	Description	Example
1	No new laceration, no visual damage	
2	No new laceration, with visual damage	
3	Laceration, exacerbated after Test 1 or new.	

7.4 Results

7.4.1 Test 1: Initial impact

The mode of the damage categories resulting from the six studs at 0.5 - 3.5 J are presented in Figure 7.5. The comparator stud lacerated (category 3 or 4) the skin simulant in all trials at 3.0 J and above. At 2.5 J, three impacts were classed as category 2 and two impacts resulted in category 3 (mode: category 2). At 2.0 J all impacts of the comparator stud resulted in damage categories 1 or 2, thus this energy was defined as the 'no laceration limit' for the comparator stud (stud 1). In this test set-up, any stud causing damage category 3 or 4 (mode of five impacts) at 2.0 J or lower would therefore have a higher predicted risk of laceration than the comparator stud. This was the case for studs 2, 3, 4 and 6 (Figure 7.5). Stud 5 on the other hand, had a higher threshold for laceration at 3.5 J and thus showed a lower predicted risk of laceration than the comparator stud in the initial impact phase (Figure 7.5).

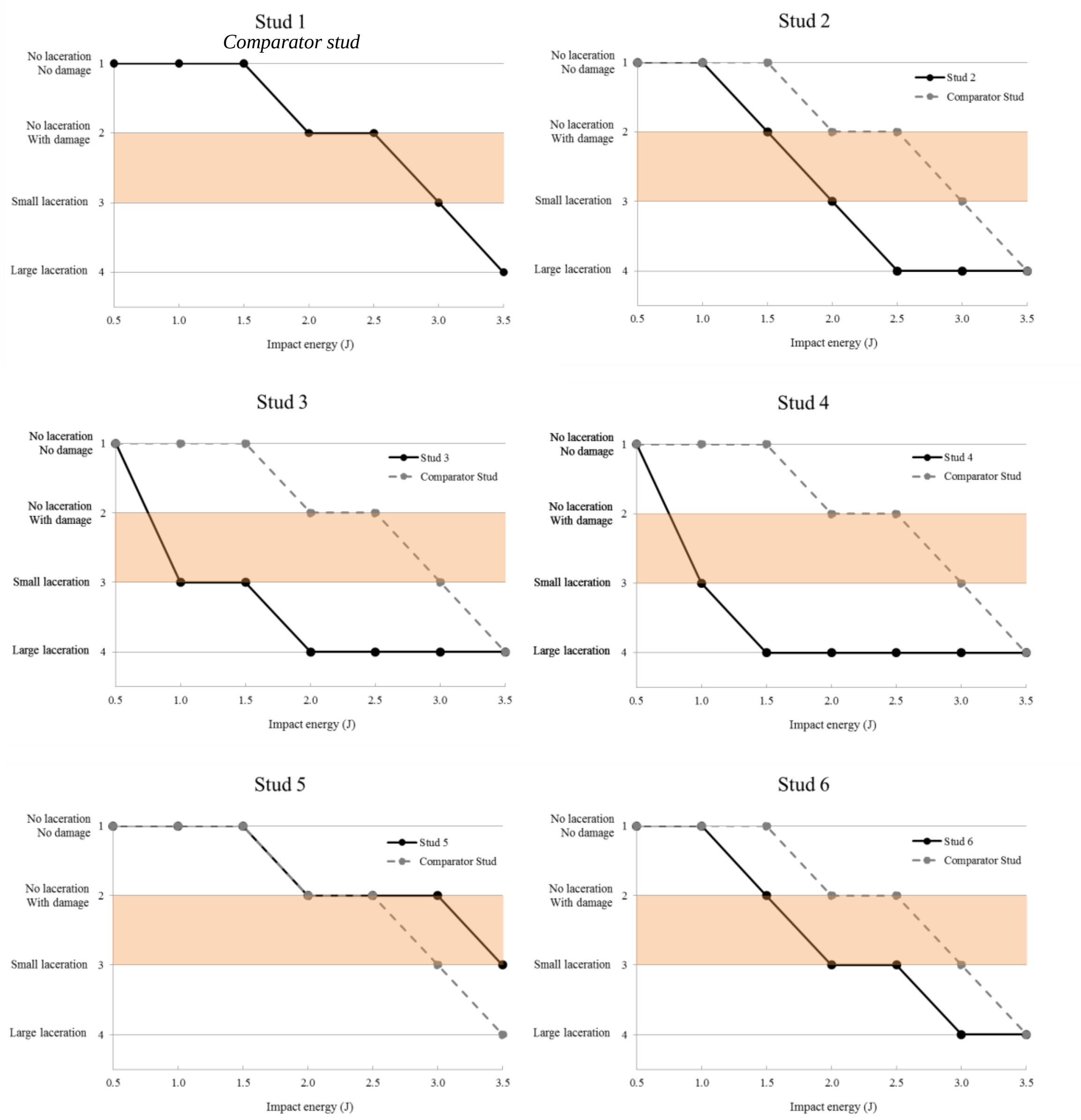


Figure 7.5: Resulting laceration injury categories of studs 1 - 6 during the initial impact test at 0.5 - 3.5 J. Studs 2, 3, 4 and 6 lacerated at a lower impact energy than the comparator stud

7.4.2 Test 2: Raking phase

Each stud was first impact tested at 2.0 J (the 'no laceration limit' for the comparator stud) before the stud and simulant are moved to the raking phase. The comparator stud was tested at 14 kg and 10 kg stud mass during the raking phase, representing approximately 137 and 98 N stud raking force. When the raking mass was lowered to 10 kg, the comparator stud did not cause the simulant to lacerate. This mass was defined as the 'no laceration limit' for the comparator stud during the raking phase, and was used to test studs 2 - 6. The results of all six studs tested at 10 kg raking mass are shown in

Table 7.5. Studs 2, 3 and 4 further lacerated the skin simulant (damage category 3) at the raking test. Stud 5 did not cause further damage when raked with a 10 kg mass (damage category 1), whilst the SmartPower© Profiler (stud 6) caused visual damage (damage category 2).

Table 7.5: Mode of five repeats from both the initial impact (test 1) and raking phase (test 2) performed at the 'no laceration limit' for the comparator stud.

Stud	**Properties**	**Damage category Test 1 (2.0 J)**	**Damage category Test 2 (10 kg)**
1	Comparator stud Ø 10 mm, 1 mm edge radius	2 no laceration, with damage	1 no damage
2	Standardised Ø 10 mm, no edge radius	3 small laceration	3 lacerated
3	Standardised Ø 5 mm, 1 mm edge radius	4 large laceration	3 lacerated
4	Standardised Ø 5 mm, no edge radius	4 large laceration	3 lacerated
5	Commercial approx. Ø 10 mm, 2 mm edge radius	1 no laceration, no damage	1 no damage
6	Commercial T-shaped stud	3 small laceration	2 visual damage

7.4.3 Inter-rater reliability

Fleiss' κ was run on both classification systems to determine what the agreement was between four independent raters. There was excellent agreement on the Landis & Koch (1977) scale between the four raters' judgements, both on classification system 1, κ = 0.848 (95% CI: 0.682 - 0.984, $p < 0.001$) and classification system 2, κ = 0.944 (95% CI: 0.836 - 1.000, $p < 0.001$).

7.5 Discussion

The study described in this chapter was performed to show how the test method that was developed in Chapter 6 can be used to assess the laceration injury risk of stud designs. This chapter identified a skin and soft tissue simulant material which can be used in conjunction with the proposed test method. The selected skin and soft tissue simulant combination was hygienic, easy to use, and gave consistent results.

Nevertheless, alternative materials could be considered in the future in order to improve the validity of the test method. The puncture resistance of synthetic chamois during a quasi-static knife blade penetration experiment had previously been compared to the puncture resistance of pig skin, showing similar results for both materials (12.4 vs 12.9 N, respectively) (Ankersen *et al.* 1999). Though pig skin is often used in these situations as a validation model for human skin (Parmar *et al*. 2012; Myouse *et al.* 2013), it has also been described as tougher due to its higher collagen content (Shergold & Fleck 2005; Lim *et al.* 2011). For both forensic and sports injury research, the development of a low-cost, off the shelf skin simulant solution replicating mechanical properties such as tear resistance, friction, puncture resistance and elasticity of human skin is needed.

The standardised studs used in this study (studs 1 - 4, Figure 7.1) lacerated the skin simulant material at lower impact energies and at a lower raking force when the top edge radius or the top diameter of the stud was reduced. The lowest impact energy causing laceration was 3.0 J for stud 1, 2.0 J for stud 2, and 1.0 J for studs 3 and 4. These findings confirm the hypothesis that standardised studs 2, 3 and 4 have an increased risk of laceration compared to the comparator stud. In comparison, the energy required to break the skin simulant in this test was higher than previously investigated for 2 mm edge radii simulated head impacts (0.25 - 1.56 J, Myouse et al. 2013). The head form used in this study was made from solid steel, potentially decreasing contact time and thus increasing peak force during impact.

When looking at the energy per surface area that is required to break the simulant skin, stud 1 lacerated at 3.8 J per cm^2 top stud surface area, stud 2 at 2.5 J/cm^2 and stud 3 and 4 at 5.1 J/cm^2. This suggests that the change in stud top surface area does not fully explain the skin breaking point. The inclined impact (stud angle and inbound velocity angle) of this test in combination with the variable edge radius of the studs is likely to have influenced the skin breaking point as well. The earlier study by Myouse et al. (2013) showed that the edge radius of the impact had a large effect on the energy required to lacerate a skin simulant, which is in line with the findings of the current study.

The shape difference between the standardised studs was pronounced, with an edge radius of 1.0 mm (stud 1 and stud 3) compared to the machine cut edge (stud 2 and stud 4). The top diameter of the comparator stud was double the top diameter of studs 3 and

4 (10 mm vs 5 mm). The sensitivity of the test method to identify smaller differences in stud shape has not been investigated yet.

Two commercially available stud designs were also tested. The aluminium rounded stud (stud 5, Figure 7.1) did not cause more damage than the comparator stud in the newly developed test method. This stud has previously passed BS 6366:2011 (British Standards 2011), which is similar to World Rugby's current Regulation 12 tests. The SmartPower© Profiler lacerated the skin simulant at a lower impact energy than the comparator stud in the first phase of the test method, with a damage category score of 3 at 2.0 J. In the second phase, the SmartPower© Profiler stud caused visual damage (category 2) whilst the comparator stud showed no additional damage during the raking phase (category 1). The introduction of the SmartPower© Profiler stud has caused media controversy (McGeady 2017) even though it had recently passed World Rugby's current test method for assessing the laceration injury risk of stud designs (World Rugby 2015). The tests described in this chapter identified an increased risk of laceration when using the SmartPower© Profiler compared to the comparator stud.

The test parameters used in this study were selected based on the mean of all stamping impacts (stud angle, velocity inbound angle, impact mass, raking speed). The inbound velocity in test 1 and the raking force in test 2 were adjusted until they reached the threshold of laceration for the comparator stud. All other test parameters were kept constant. Future testing should investigate if adjusting the other test parameters, according to the clusters identified in Table 6.2, influences the outcome of the test. Especially with non-symmetrical studs, the stud angle can influence which side of the stud first comes into contact with the simulant, potentially leading to a different test outcome.

The impact energy required to lacerate the skin simulant using the comparator stud in the initial impact test was between 2.5 - 3.0 J, which is lower than the mean of all stamping impacts (9.5 J, Table 5.2). There are three potential explanations for this: 1) each of the stamping impacts in the biomechanics study would have caused an injury even when using a comparator stud, 2) the chamois - silicone combination that was used in the final test set-up had a lower breaking threshold than human skin, and 3) the stiffness of the initial impact test was higher than during the measured shoe-ATD impacts. As previously discussed in Chapter 6, the initial impact test does not replicate the time-to-peak force as observed in the stamping impacts. The increased stiffness of this test is likely to have led to a shorter contact time and therefore higher peak forces when the effective mass and the inbound velocity are kept constant. Human structures provide a complex response to impact, with internal structures providing both damping and muscle activation. During the development of the test method its increased stiffness compared to human interactions was considered, but lowering this stiffness would reduce the durability and reproducibility of the test method, and could have also increased its cost. The developed test method should be used to compare the injury risk of new stud designs against the injury risk of the defined comparator stud, to which no absolute threshold values on the risk of injury should be assumed from the outcome.

The outcome measure developed for the new test method was two damage category systems, where the skin damage was rated after each trial. The mode of five repeat trials was taken for each stud. The inter-rater reliability of the developed damage category systems was high (phase 1, $\kappa = 0.813$; phase 2, $\kappa = 0.944$), showing that the marking of the damage category system was robust. The intra-rater reliability was not assessed. The mode of five repeat trials was chosen as an outcome measure because of the ordinal nature of the damage category scores.

Varying the impact energy by 0.5 J increments at the initial impact phase ensured that each stud followed a 'transition phase', where both laceration and non-laceration wounds occurred. For the comparator stud, this transition phase was at 2.5 J. In the raking phase, stud mass could be adjusted at 4 kg increments. The studs did not show a similar transition phase during the raking test, i.e. all five repeats for each stud always fell into the same category. To determine a more accurate threshold for stud laceration resulting from the raking phase, smaller mass increments should be developed.

7.6 Chapter findings

The aim of this chapter was to show how the test method developed in Chapter 6 could be used to compare stud designs on their predicted risk of laceration. A skin and soft tissue simulant material was identified which was used in conjunction with the developed mechanical test. Subsequently, the laceration injury risk of six different stud designs, four bespoke made standardised studs and two commercially available studs, was investigated. The threshold of laceration injury for the comparator stud was found by adjusting the impact energy for test 1 and the raking force for test 2. All studs were subsequently tested at this laceration threshold and the test outcome was determined using a skin damage category system. The results showed that decreasing the edge radius of a stud increases its laceration injury risk, and the same result was found for decreasing the top diameter. A commercially available aluminium rounded stud design had a lower predicted laceration injury risk than the comparator stud, whilst the SmartPower© Profiler stud had an increased predicted risk of laceration. This chapter showed that the developed test method can distinguish between standardised stud designs and can be used to relate the laceration injury risk of commercially available studs to a comparator stud. More research is needed to investigate the ability of this test method to differentiate between smaller differences in stud designs. The representativeness of the test method could be improved by lowering the stiffness of the pendulum arm and improving the biofidelity of the skin simulant used.

8 Conclusions

8.1 Summary of findings

Objective 1: Investigate the current prevalence of skin laceration injuries in rugby union.

A systematic review of published epidemiological studies on rugby union players found that skin and laceration injuries accounted for 5.1% of match injuries and 1.5% of training injuries in rugby union. Furthermore, 2.4 skin or laceration injuries occur per 1000 match exposure hours, which could be interpreted as one time-loss injury per team, per season. There was no difference in skin and laceration injury risk between amateur and professional players during match play; though in training sessions, amateur players had a higher risk of sustaining skin and laceration injuries than professional players.

Objective 2: Identify the most prevalent game scenario in rugby union that leads to laceration injuries caused by studded footwear.

A survey study was conducted to investigate the dominant laceration injury scenario resulting from stud-skin interactions. Of the questionnaire respondents (N=191), 72% had sustained one or more laceration injuries caused by studded footwear serious enough to hinder play and / or leave the pitch. Over half of the reported injury scenarios in the survey results were classed as 'stamping in the ruck', which was subsequently identified as the dominant stud laceration injury scenario.

Objective 3: Determine the kinetics and kinematics of stud-skin impacts during the game scenario identified in Objective 2.

A laboratory-based study was conducted in which participants replicated a one-on-one rucking scenario and stamped on the chest of an anthropomorphic test device. High-speed cameras and pressure sensors were used to capture the kinetics and kinematics of a total of 110 stamps, performed by 12 participants. The inbound velocity, stud angle, stud peak force, stud energy and stud effective mass were calculated for each of the stamping impacts. A range of stud angles were observed, between -13° (heel strike) to +35° (toe down), suggesting that these angles should be implemented in future test methods. The inbound velocity (mean: 4.3 m/s) was higher than the current test method recommends (1.0 m/s) whilst the stud effective mass (mean: 1.2 kg) was lower than suggested in current test method (8.5 kg). The findings of this study showed that in the design of a new test method a higher inbound velocity but a lower impact mass should be adopted. Further, two distinct phases of stamping impacts were identified: an initial impact phase and a subsequent raking phase.

Objective 4: Design and build a mechanical test for assessing stud laceration risk based on the kinetics and kinematics measured.

A mechanical test method replicating these two impact phases was designed and manufactured. Interviews with World Rugby and a sport product test house identified that such a method should be reproducible across test centres. An overarching standard for both rugby and football studs was found desirable and therefore the new test method should ideally incorporate flexibility to test at different impact parameters. The developed test method focussed its design parameters on the kinetics and kinematics of the measured rugby stamping impacts. Various design solutions were considered during the development phase. The final design uses a pendulum with adjustable pivot points for replicating the initial impact phase, and a spring-damper system to recreate a raking motion in the second test phase.

Objective 5: Compare the laceration injury risk of stud designs using the developed method, thus providing a proof-of-concept.

A study comparing six stud designs on their predicted risk of laceration was conducted using the newly developed test method. Three types of skin simulant materials were identified in literature and their appropriateness for use with the developed test method was assessed. The combination of a synthetic chamois with a 5 mm silicone layer acting as a soft tissue simulant was found suitable. One of the studs included in the study was defined as a comparator stud following World Rugby's current regulations, providing a benchmark of 'acceptable injury risk'. The developed test method was able to distinguish between standardised studs with different edge radii and top diameters. Two commercially available stud designs were also tested, a commonly used aluminium rounded stud and the recently introduced SmartPower© Profiler, which has passed World Rugby's current studded footwear regulations. The aluminium rounded stud was classed as a lower predicted risk of injury than the comparator stud. The SmartPower© Profiler resulted in an increased predicted risk of injury when benchmarked against the comparator stud. The final study showed how the developed test method can be used to identify a potential increase in laceration injury risk of stud designs.

8.2 Limitations

The prevalence, aetiology and biomechanics of laceration injuries caused by studded footwear in rugby union had not previously been investigated. It was therefore often not possible to compare the results of this project to other research. The limitations of each stage of this research project have been discussed at the individual chapters. The main limitations impacting on the conclusions of this work are:

- When investigating the prevalence of skin and laceration injuries in published epidemiological studies, it was found likely that the injury definitions used were prone to under report skin injuries. Adopting guidelines on reporting of injury categories, as previously published (Fuller, Molloy, *et al.* 2007), should help eliminate this bias.
- The survey study retrospectively investigated the type of play scenarios causing laceration injuries resulting from studded footwear. Retrospective self-reporting of injuries can suffer from recall bias, where participants might misremember, exaggerate or understate the severity of their injury. The pressure sensors that were used to measure the kinetics of stamping impacts have previously shown poor accuracy. This error was minimised with custom calibration procedures, but remained an estimated 7.5%. The peak force, impact energy and effective mass results should be interpreted with this limitation in mind.
- Participants in the stamping study did not require a minimum amount of rugby experience. Stamping was not regarded a 'skilled' movement in rugby because it is not officially allowed in the game. However, the limited rugby experience of the participants could have influenced the results of the stamping study.
- In the developed test method, the initial impact phase was not designed to replicate the time-to-peak force of stamping impacts. The arguably high stiffness of the pendulum arm in the developed test method ensures a repeatable and reproducible impact that can be used as a standard.

- The developed test method should be used to compare stud designs on their predicted laceration injury risk. No absolute impact energy or raking force values causing lacerations can currently be identified from the test method.

8.3 Recommendations for future research

Future work could focus on improving the developed test method in several ways. Firstly, matching the stiffness of the pendulum arm in the initial impact phase to the measured kinetics of stamping impacts will increase the representativeness of the developed test method. The pressure exerted by the stud using the developed test method can be measured with the previously used pressure sensors, and subsequently compared to the pressure measurements in the biomechanics study. Secondly, this project focussed on replicating one injury scenario (stamping in the ruck) in the developed test method, though other injury scenarios accounted for the remaining 45% of stud laceration injuries. The kinetics and kinematics of secondary injury scenarios identified in Chapter 4 can be measured to determine further test setting combinations and / or adjustments to the test design. Finally, the development of a low-cost, off the shelf skin simulant solution which more closely replicates the mechanical properties of skin such as tear resistance, friction, puncture resistance and elasticity can help improve the validity of the developed test method.

Besides using the developed test method as a standard or regulation, it could also be used as a design tool. The influence of different materials, stud shapes and wear on laceration injury risk can be investigated. Such data can help inform stud design solutions which lower their laceration injury risk whilst keeping stud performance parameters constant.

8.4 Recommendations for future implementation

The developed test method would need more work before it could be implemented as a test method in rugby union regulations. Firstly, the accuracy required for each of the test method's settings such as its inbound velocity, impact mass, and stud angle need to be identified. The required accuracy of the test method should inform the margins of error that any future test method design has to adhere to.

Secondly, the influence of using more combinations of test settings (as identified in Table 6.2) on test outcome needs to be investigated. The current proof-of-concept study (Chapter 7) only measures studs at test settings replicating the mean of measured stamping impacts. Other test settings, replicating the mean of the four identified groups, could provide different results.

Finally, impact parameters from stud laceration injury scenarios in other field sports can be used in combination with the research presented in this thesis to develop a test method that is suitable to use as an overarching standard for sports such as rugby union, rugby league and association football. For this, injury scenarios from other field sports using similar studded footwear should be identified and the kinetics and kinematics of such stud-skin interactions measured.

8.5 Contributions to knowledge

This project provides an original contribution to the existing literature on sports injuries by developing a test method that can be used to investigate the effect of stud design on laceration injury risk in rugby union. Its main contributions to knowledge are:

- The current skin and laceration injury prevalence of amateur and professional rugby players was identified. This provides a benchmark of injury prevalence which can later be used to evaluate whether the newly developed test method helps reduce this injury prevalence.

- Common game scenarios causing stud laceration injuries in a predominantly amateur population were identified. This information was utilised to inform the type of injury scenario that a test method for stud laceration injury risk should replicate.

- The kinetics and kinematics of stamping impacts in a laboratory environment were measured. This data was essential for informing the design criteria of the developed test method.

- A mechanical test method was developed that can distinguish between stud parameter changes (edge radius and top diameter) and can be used to compare the laceration injury risk of commercial studs to a standardised stud. The test parameters of this method were based on rugby game-relevant loading conditions.

- A commercially available, cost-effective and repeatable skin and soft tissue simulant combination was identified which can be used in conjunction with this test method.

9 References

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10 Appendices

10.1 Appendix A: Questionnaire

Thank you in advance for taking the time to fill out this questionnaire. For my PhD, I'm researching skin laceration injuries caused by the studs on rugby boots. I would like to know your experience with, and opinion about these injuries. The survey will take about 5 minutes to complete.

All information is automatically anonymised. By completing this survey you give consent that your anonymised results may be used in publications and in support of my PhD research.

If you have any questions, please don't hesitate to contact me.

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Consent & Playing position

Question 1

I hereby state that...

- ☐ I am over 18 years old
- ☐ I agree for my anonymised answers to be used in this study

Question 2

What position do you usually play?

If you change positions frequently, click the two most frequently played positions.

- ☐ Prop (loose- or tight head
- ☐ Hooker
- ☐ Second row
- ☐ Flanker
- ☐ No. 8
- ☐ Scrum half
- ☐ Fly half
- ☐ In- or outside centre
- ☐ Wing
- ☐ Full back

Question 3

What type of shoes do you wear when playing rugby?

I am interested in the type of studs (cleats) are underneath your sole. If you have multiple pairs of boots depending on the surface you are playing on, you can check multiple boxes. The image below can help you identify your stud type

- ☐ 1 Shoes with screw-in studs, completely aluminium
- ☐ 2 Shoes with screw-in studs, plastic with aluminium tip
- ☐ 3 Shoes with screw-in studs, completely plastic
- ☐ 4 Shoes with a combination of screw-in studs and moulded studs
- ☐ 5 Shoes with moulded soles, round studs
- ☐ 6 Shoes with moulded soles, astroturf
- ☐ 7 Shoes with moulded soles, 'blades'
- ☐ 8 Shoes with moulded soles, triangular studs
- ☐ Other, please specify ____________________

Stud injuries

Question 4

For my research, I am interested in skin injuries caused by studs / cleats

On average, how often do / did you experience a minor stud injury when playing rugby?

'Minor' here means that it did not stop you from playing. Includes chafes, cuts and bruises.

- ☐ Every week
- ☐ About once a month
- ☐ Couple of times per year
- ☐ Once a year or less
- ☐ I never had such an injury (skip to question 10)

Question 5

During your whole rugby career, how often have you experienced a substantial stud injury?

'Substantial' means an injury which hindered you from (fully) participating in training or a match.

_____________ times

Question 6

When taking the most severe stud injury / injuries you have experiences, can you tell what type of event caused the injury?

If you have only had minor stud injuries, take one of the more substantial or memorable ones. In the 'Other' box you can describe a different event if necessary.

- ☐ Being tackles
- ☐ Tackling someone
- ☐ On the ground in a ruck
- ☐ Stepping over a ruck
- ☐ Collapsed maul
- ☐ I don't remember
- ☐ Other: ___________________________

Question 7

Who's studs caused the injury described in the previous question?

- ☐ Your own studs

- ☐ Your own team mate's studs
- ☐ Opponent's studs
- ☐ I don't remember

Question 8

Can you remember one specific stud injury, and describe the event that caused it? Please be as detailed as you can.

For example 'I tackled the opponent from behind, and whilst he tried to push off my arms were wrapped around his ankles, and his stud caught my eyelid'. If you can't remember, you can leave this box blank.

[free text response]

Question 9

Where on the body have you had skin injuries caused by studs?

Minor or substantial. Multiple answers possible.

- ☐ Foot / ankle
- ☐ Lower leg
- ☐ Upper leg
- ☐ Chest & belly
- ☐ Back
- ☐ Arms
- ☐ Hands
- ☐ Head / neck / face
- ☐ I don't remember
- ☐ Other: ____________________

Stud checks

Question 10

Does the referee usually check your studs before the match?

- ☐ Yes, (almost) always
- ☐ Sometimes
- ☐ No, (almost) never *(go to question 12)*

Question 11

What do you think about the stud check at the start of the match?

- ☐ Good, definitely needed to keep the players safe
- ☐ Good, but usually does not happen thoroughly enough
- ☐ Not useful, wastes our time
- ☐ Other: ____________________

(go to question 13)

Question 12

Do you think you should have a stud check at the start of the match?

- ☐ Yes
- ☐ No
- ☐ Not sure / no opinion
- ☐ Other: ____________________

Question 13

Do you ever check your own studs?

- ☐ No, I never think about checking them *(skip to question 14)*
- ☐ No, I only have moulded studs and don't think they need checking *(skip to question 14)*
- ☐ Yes, I only have moulded studs and I do check them
- ☐ Yes, but only my screw-in studs, to see if they are still tightened
- ☐ Yes, but only my screw-in studs, on tightness and sharpness
- ☐ Yes, both my screw-in and moulded studs on tightness of the screws and sharpness.

Question 14

Have you ever found your studs sharpened due to wear?

- ☐ Yes
- ☐ No
- ☐ I can't remember

General information

Just before rounding off, we would like to know a little bit about you in order to interpret your responses appropriately.

Question 15

What is your age (in years)?

____

Question 16

How long have you played / been playing rugby (in years)?

____

Question 17

On average over these years, how often do / did you train and play matches per week?

- ☐ Once a week
- ☐ 2 - 3 times a week
- ☐ 4 - 5 times a week
- ☐ 6 -7 times a week

10.2 Appendix B: Calibration study

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Pressure sensor calibration for measuring stud-player impacts

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Abstract

In rugby union, laceration injuries can occur from players stamping on opponents in the ruck. To measure the stud-skin interaction during stamping movements, pressure sensors can be used. Pressure sensor calibration techniques have highlighted the need to perform calibrations using appropriate impact dynamics. A pilot study with seven rugby players informed the expected peak forces and loading rates of rugby stamps. Subsequently, a custom calibration procedure was developed, using a drop hammer and force platform to replicate the experimentally observed forces and loading rates. The conventional calibration of the pressure sensor system, supplied by the manufacturer, overestimated total force by 132%. The method described in this paper resulted in a mean error of 7.5%. This study describes a simple and effective calibration procedure for using pressure sensors when measuring the peak force from stud-player impacts. The method has potential to be used as an improved calibration protocol when the expected peak force range of the measured event is between 1800 and 3000 N. The calibrated pressure sensors will be used to obtain kinetic data from stamping events in the ruck in rugby union.

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Keywords: shoe-skin interaction; pressure sensors; calibration; laceration injury; rugby ruck; peak force

1. Introduction

Lacerations account for approximately 6% of the professional match injuries in rugby union [1–3]. Laceration injuries in rugby union can be caused by a number of factors, one of them being studs on players' boots. Studs are regulated by World Rugby (the world governing body for rugby union). Regulation 12 [4] describes 'performance tests' for assessing laceration injury risk of new stud designs. Compliance with Regulation 12 is currently optional for manufacturers. The performance tests aim to simulate stamping - a perpendicular movement of the foot onto a player on the ground; and raking - a similar movement which also includes a horizontal component. Stamping has been previously identified as the most common mechanism causing stud laceration injuries in rugby union [5]. The recommended impact energy stated in the regulation for a stamp test is 4.2 J (50mm, 1m/s, with 8.5 kg). There is no recommended impact energy for the raking test. The British Standards for football shin guards (BS EN 13061:2009) require 14.6 J impact energy (1.0 kg impactor at 5.4 m/s) for testing stud impacts. Payne *et al.* (2013) [6] reviewed the design of human surrogates for the study of biomechanical injury and used an estimate of the lower leg and foot mass (4.58 kg) and in-play impact velocities for a football tackle (1.2-2.5 m/s) to calculate stud-player impact energy (3.3-14.3 J). The actual impact energy of foot-player contact is rarely measured and research using impact tests typically use estimates of body segment mass and inbound speed [7,8]. Ankrah and Mills (2003) [9] reviewed multiple shin guard test methods for their stud-impact specific test and found no other studies had considered stud impacts before. They estimated the stud impact energy at 10-13 J based on an effective foot mass of 0.62 kg and inbound speed of 6 m/s.

The wide variety of impact parameters used in previous research suggests a need for accurate measurements of foot-player contacts if test methods are to be representative and repeatable. The two main variables required for a simulated stamping impact are the inbound speed and impact force. Previously, other sports impacts have been quantified using pressure sensor technology [10–12]. Resistance-based force distribution sensors, such as the piezo-resistive pressure sensors made by Tekscan Inc (Boston, USA), are thin, flexible and give time-based results. Some problems with commercially available pressure sensors exist; mainly the influence of the calibration method on measurement accuracy [10,13,14]. First, the sensors are resistive and their response

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over their total capable measuring range is non-linear. When testing over a smaller loading range, sensor response within this range can appear linear [14]. Calibrating for the expected loading range is therefore important. Second, the rate at which the force is applied to the sensor, i.e. the loading rate, changes the behaviour of the sensor. When testing sports impacts which are dynamic in nature, it follows that a dynamic calibration procedure is recommended [10,15]. Finally, when applying constant force, piezo-resistive pressure sensors exhibit positive logarithmic drift [13]. When measuring pressure of force after initial impact, the output of the sensors needs to be corrected for this drift. After comparing standard calibrations from Tekscan Inc to custom calibrations, Brimacombe *et al.* (2009) [14] recommend user-defined calibration protocols to improve the measurement accuracy of the sensors. They also recommend to calibrate each sensor individually, since calibration equations can differ.

Obtaining kinetic and kinematic data from stud-player impact scenarios is non-trivial. It needs a lightweight, flexible system which is able to measure force and will be safe for participants to impact. Previous studies have successfully used pressure sensors for measuring in-play impact force in basketball [11] and rugby [12], but custom calibration methods need to be developed in order to optimise the accuracy of the sensors. This paper reports on an effective and simple calibration method for pressure sensors to measure peak impact force of the magnitude likely to be experienced during stud-player impacts in rugby union. The method is also applicable to other sport impact events.

2. Methods

Ethical approval for this study was obtained from the Health and Wellbeing ethical committee of Sheffield Hallam University.

2.1. Obtaining loading range parameters

To ensure the sensors were calibrated within a representative loading range, a pilot test was initially undertaken in which seven rugby players were asked to replicate stamping another player in the ruck. Participants were between 19 and 28 years old, weighed 80 ± 8 kg and had at least two years of experience playing rugby. The impact force of participants stamping directly onto the force plate was collected by a floor mounted force plate (9281CA, Kistler Instrument Corp, Winterthur, Switzerland; sample frequency 1000 Hz). Each participant repeated the stamp three times, giving a total of 21 stamps. Peak force was identified for each trial. The mean derivative of the loading phase (start impact to peak force) of the force trace was defined as 'loading rate', reported in N/ms. A 95% confidence interval (CI) of peak force was calculated for all stamps, as to define a range of forces for which the pressure sensors needed to be calibrated. This provides three impact energies; 95% CI lower bound, 95% CI upper bound and the mean.

2.2. Experimental set-up

Eight unused Tekscan F-scan 3000E 'Sport' flexible pressure sensors were used for evaluating the accuracy of force reconstruction after a custom calibration procedure within selected force ranges. An unmodified version of the Tekscan F-scan system was used in this research. Each sensor consisted of 954 force sensing elements called 'sensels'. Each sensel measures 2.5 by 2.5 mm giving a density of 3.9 sensels / cm^2. Each sensor was mounted on top of a portable force plate (9286AA, Kistler Instrument Corp, Winterthur, Switzerland) and protected by a 40mm thick sample of Silastic 3483 (Dow Corning Ltd, Michigan, United States). The loading rate response of various materials were analysed during pilot testing, and the silicone material Silastic 3483 was found most similar to loading rates observed in the initial stamping study. The material has also previously been found similar to *in-vivo* human tissue response by Hrysomallis (2009) [7]. The pressure sensor - force plate combination was impacted with a hemispherical head of 1.68 kg. Drop height was set at 58, 71 and 84 cm to replicate peak forces within the 95% CI range that was found in the stamping study. Raw pressure sensor data and vertical force was concurrently recorded by the pressure sensor system at 750 Hz and force plate at 3000 Hz.

2.3. Calibration procedure

Raw pressure sensor data was first resampled to 3000 Hz (matching the force plate sample rate) using spline interpolation with Matlab (R2014a, the Mathworks Inc, Massachusetts, United States). Pressure sensor and force plate signals were synchronised based on the peak pressure and peak force. For each trial, the time of peak force was identified and the sample time of the preceding three data points was also recorded (Figure 1). Force plate values and raw pressure sensor values at the four data points were extracted for the first four repeats only. This resulted in 16 pressure sensor and force plate data points for each test condition. The correlation between force and pressure sensor values were assessed using a polynomial fit. The coefficients for first, second and third order polynomials were calculated by minimising the sum of the squares of the error (Gauss's least squares approach). The polynomial with the lowest error was selected as the calibration relationship. One repeat for each test condition, and thus four data points per test condition, was not used for obtaining polynomial coefficients. These four data points were considered 'blind' and subsequently used to estimate the error of the custom calibration by applying the resulting calibration relationship to the raw pressure sensor data. The estimated peak force derived from the calibration of the blind pressure data was then compared to the measured peak force and the percentage error calculated.

Two pressure sensors were also calibrated with the manufacturer's recommended 'step' calibration procedure for dynamic testing [16] and both raw and calibrated data was obtained for these sensors. This made possible a comparison between the new custom calibration technique and the conventional technique.

3. Results

3.1. Loading range parameters

Seven participants repeated the stamp three times. Mean peak force of 21 stamps was 2328 N, with 95% CI [1834 N, 2821 N]. Mean loading rate over 21 trials was 254 N/ms with 95% CI [201 N/ms, 307 N/ms].

3.2. Test conditions

The corresponding drop hammer test conditions to replicate the loading parameters identified in the pilot study are shown in Table 1.

Table 1. Overview of test conditions of each sensor. Peak forces match those found in the initial stamping study.

	Drop mass (*kg*)	Drop height (*cm*)	Resulting peak force (*N*) and corresponding stamp force loading parameter
Condition 1	1.68	58	1834 *lower bound 95% CI*
Condition 2	1.68	71	2328 *mean peak stamp force*
Condition 3	1.68	84	2821 *upper bound 95% CI*

3.3. Sensor calibration

Eight different sensors were tested five times at conditions 1 - 3 (Table 1). Figure 1 shows an example force trace and corresponding raw pressure values. For each test condition 4 data points were recorded for each repeat (total n = 16). Using the Gaussian least squares method, a first order polynomial was identified as the best fit of the correlation between force and raw pressure values. This indicated that the calibration relationship was linear. Figure 2 (a - h) show the resulting linear calibration factor for each sensor.

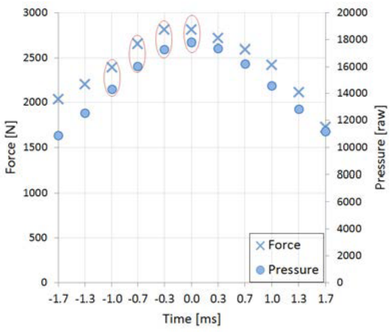

Fig. 1. Example force trace and raw pressure values. Peak force and three values leading up to peak force are highlighted in red circles

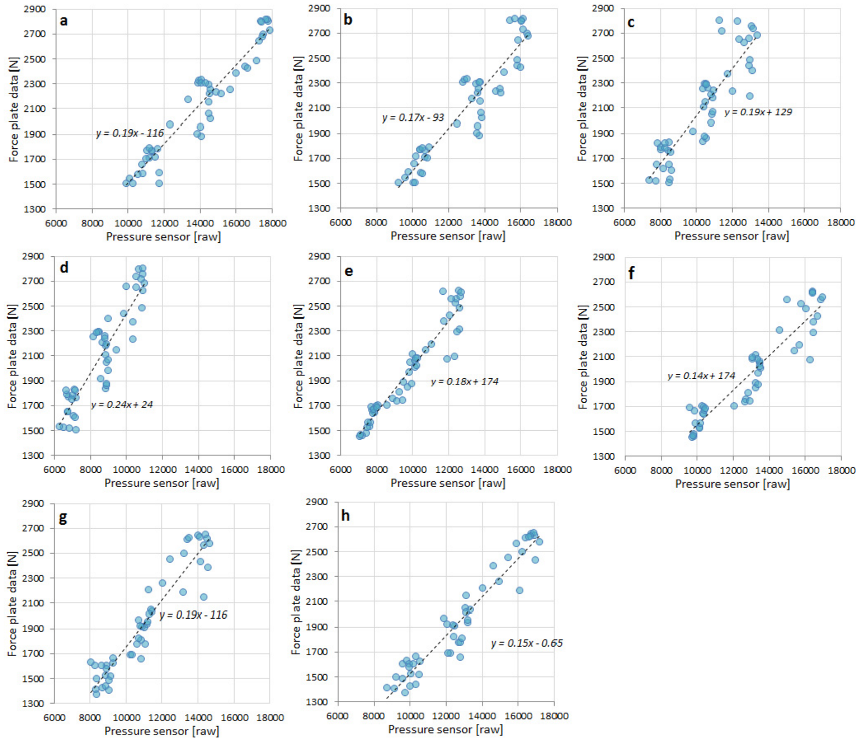

Fig. 2. Linear calibration within expected peak force range for (a) sensor 1; (b) sensor 2; (c) sensor 3; (d) sensor 4; (e) sensor 5; (f) sensor 6; (g) sensor 7; (h) sensor 8.

To assess the error of the resulting linear calibration factor, a blind set of raw pressure data collected during testing was used. The raw pressure values were calibrated using the linear calibration factor found for the corresponding sensor and the peak force values calculated. These were compared to the actual peak force measured on the force plate (Table 2). During the testing of sensor 1 and 2, Tekscan 'step' calibrated pressure sensor data was obtained next to the raw data. When comparing the Tekscan 'step' calibration results to the force plate data, the pressure sensors overestimated the peak force in both sensors (Table 2).

Table 2. Mean and standard deviation of the peak force measured from the force plate and compared to the custom and manufacturers calibration methods.

Condition	Peak force (*N*) from force plate	Loading rate (*N/ms*) from force plate	Error (%) of custom calibration process	Error (%) of manufacturers calibration process
1	1730 ± 75	278 ± 14	5.3 ± 1.7	134 ± 6.6
2	2169 ± 140	361 ± 26	7.5 ± 2.2	129 ± 3.8
3	2701 ± 87	460 ± 17	9.6 ± 7.6	134 ± 5.3
All	2200 (*n = 48*)	360 (*n = 48*)	7.5 ± 5.0 (*n = 96*)	132 ± 5.9 (*n = 30*)

4. Discussion

Piezo-resistive pressure sensor systems, such as Tekscan's F-scan pressure sensor system, are useful in biomechanical applications because they are lightweight, thin, flexible and safe to use. Previous research has highlighted the need for custom calibration methods with similar impact mechanics to its intended use, in order to improve the accuracy of measurement systems [10,13,15]. The expected force range of stamps in rugby was 1834 - 2821 N (95% CI). The 95% CI of the loading rate of simulated stamping was 201 N/ms - 307 N/ms. Conditions 1-3 of the calibration procedure aimed to replicate the range of peak forces found in the stamping test and established 2200N force. Loading rate was on average higher than the 95% CI at 360 N/ms. Lower drop heights were associated with lower loading rates (278 ± 14 N/ms for 58 cm).

The expected error of the measurement system is important for establishing confidence bounds and the interpretation of testing results. When minimising the sum of the squares of the errors with a first order polynomial, the error of the eight sensors was 7.5%. The error of the custom calibration is consistently lower than the error of Tekscan's standard calibration method, the 'step' calibration (mean ± SD: 132 ± 5.9%; n = 30).

Over time, pressure sensors can experience signal degradation, i.e., a loss of sensitivity. In dynamic loading situations, the effect of shear loads, loading between curved surfaces and small contact areas can all decrease measurement accuracy of pressure sensors [14] and it is difficult to predict this for future testing. Different types of data fitting were not tested in this study. Cazzola *et al.* (2013) [13] recommends a logarithmic regression to correct for drift when measuring longer contact times, as can be seen during rugby scrum impacts. In our pilot study we have seen that stud-player impact peak force can be expected at the first 40ms of the impact. Therefore no drift correction was applied to the signal. Non-linear fitting could prove itself useful when calibrating over a larger force range than was required for stud-player impact testing; it could be that the conductive response of the sensors is not directly proportional to the applied pressure. Calibrating each individual 'sensel' of the sensor separately has previously been shown to not have an effect on the magnitude of load measured [14].

Potential improvements for this calibration method would be to use a studded impactor instead of a hemispherical head, as to match the intended use more closely. Human-surface interaction is difficult to accurately simulate in a consistent way, and when aiming to match peak force and loading rate in our experimental set-up, a combination of silicone materials and a drop hammer machine presented the best results. Linear calibration methods produced error estimates of less than 10% within expected peak force loading ranges. It is currently unclear of a similar calibration approach would be equally effective for lower or higher force ranges than tested (1800 to 3000 N).

The findings of this study will be used in future experiments exploring the stud-player impacts which are thought to cause laceration injuries in rugby.

5. Conclusion

This study described a simple and effective calibration procedure for using pressure sensors when measuring stud-player impacts. Eight piezo-resistive pressure sensors were impacted with similar peak loads to rugby stamping impacts. The force range of the impacts was 1800 to 3000 N. The expected measurement error of the system after the standard manufacturers calibration was found to be 132 ± 6%; custom calibration reduced this error to 7.5 ± 5%. Linear calibration curves were found suitable for the application. This method shows that a calibration protocol simulating the expected peak force range - in this case 1800 to 3000 N - of the intended measure can reduce the error of a pressure sensor whilst using linear calibration coefficients.

6. Acknowledgements

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10.3 Appendix C: Pugh's 32 element model

1. Performance

- The finalised test rig should replicate the inbound velocities from the identified clusters (2.9 - 5.4 m/s).
- The test rig should replicate the inbound velocity angles from identified clusters (30°- 60°).
- The test rig should replicate the impact mass calculated from identified clusters (0.8 - 1.7 kg).
- The test rig should replicate the pitch angle from identified clusters (-10° - +30°).
- The test rig should incorporate a raking phase of the stud with end velocity of approximately 0.9 m/s.
- Sliding distance during the raking phase should be at least 52 mm.
- Stud force during this raking phase should be between 6 - 22 kg (95% CI).
- All impact parameter settings should be repeatable and accurate.
- The test rig needs to be able to mount a variety of studs by a simple method (e.g. a screw thread)
- The test rig needs to be able to mount a skin and soft tissue simulant in a sturdy, repeatable way.
- The mounting of the skin and soft tissue simulant should allow for these to be easily changed.

2. Environment

The equipment can expect to see temperature changes between 10 - 30° C. Pressure, humidity and vibrations will all be within normal, indoor levels. The equipment can be expected to only be used by trained individuals. It needs to be sturdy enough to handle some moving around, although not on a regular basis.

3. Life in service (performance)

The product life in service should be a minimum of five years, ideally up to 15 years. The intensity of product use will vary, but expected to average out between 1 - 3 impacts a week.

4. Maintenance

In the case that the product is build in-house, regular maintenance will be available during the lifespan of the product. If (part of) the total design is purchased from elsewhere, regular maintenance for this part is not desirable.

5. Target product cost

Material costs: Material costs cannot exceed £1500 due to the funding of this project.

Time commitment: The test rig will be built by the principal investigator (B Oudshoorn) in combination with our in-house design engineer (T Senior), employed by Sheffield Hallam University. The design engineer will have limited time available to spend on this project, thus time-consuming designs should be avoided where possible.

6. Competition

No direct competitors for this product are currently on the market. To increase the likelihood the developed test method will be accepted, basing its design on or around existing equipment is desirable, though not a requirement.

7. Shipping

The portability of the test method is not a main concern. However, in the event of moving research labs, the option for disassembling the test rig to a portable size would be required. No boat or aircraft shipping will likely be needed.

8. Packaging

Not applicable.

9. Quantity

The developed test method is likely to be a one-off design, used as a research tool. If the developed test method is implemented as a test standard, simplifications could be made to design to make it more easily reproducible. However, no investments should be made in specific tooling to make the manufacturing of multiple test rigs easier or cheaper.

10. Size

The height of the test rig cannot exceed normal ceiling height (2.4 m) and ideally should measure less than 3 x 3 m (width x length) for this is the currently available lab space.

11. Weight

No specific weight limits are applicable to this project. The finalised test rig is unlikely to require shipping within its lifespan.

13. Aesthetics, appearance and finish

Because the test rig is not designed for commercial resale, the aesthetics of the product are not deemed a priority. Where possible, wires and loose objects should be neatly concealed within the design to improve Health & Safety aspect of the test design.

14. Materials

No specific restrictions apply for the use of different materials. Other PDS considerations such as life in service, weight and costs determine the choice of material.

15. Product life span

The current studded footwear test methods have been implemented in 1990 and have not been changed since. Based on this, a product life span of 15 - 25 years can be expected to be needed, though replacement products within this time span could be manufactured as needed.

16. Standards and specifications

The developed test method has the potential to become a new standard in studded footwear regulations. The reproducibility of the design across other research centres or test houses around the world is therefore important.

17. Ergonomics

The test operator needs to be safe and comfortable when using the test rig. For acquiring the right inbound velocity, the expected height of the test operator needs to be considered. When utilising gravity to accelerate the impactor, drop height should not exceed the average person height, unless specific systems are put in place to re-set the test rig after each impact (ladder, step or pulley system). Re-setting the test rig for each impact should be safe and repeatable.

18. Customer

The first customer of the designed rig will be the designer (principal investigator, B Oudshoorn). Future users are likely to be other researchers, and potentially test house employees.

19. Quality and reliability

The quality and reliability should be maximised within the boundaries of the other design constraints on money and time. To ensure consistency in test outcomes, no test rig degradation can occur within the lifespan of the product and the settings of the inbound parameters of each test repetition should be repeatable to the highest possible standard.

20. Shelf life (storage)

Not applicable.

21. Processes

Not applicable.

22. Time-scales

The time scale for the whole project was three years. Assessing current laceration prevalence, identifying the dominant injury scenario and measuring the kinetics and kinematics associated with this type of injury need to be included in this time-scale. After the design and building of the test method has finished, a proof-of-concept of the developed product is needed. A design and building time of six months was thus allocated to this project.

23. Testing

The finished product should be tested to ensure it matches the performance criteria that were set out. This testing could be integrated with a proof-of-concept study.

24. Safety

Operating the machine should cause no harm to the operator in any circumstance. The final product should pass Sheffield Hallam University's Health and Safety requirements, including training for the operator and a risk assessment.

25. Company constraints

During the design phase, adapting or utilising currently available products within the Centre for Sports Engineering Research, or Sheffield Hallam University, is preferable. If current products do not suffice, a bespoke test rig can be developed. In this case, the design should take into account manufacturing methods available in-house.

26. Market constraints

Not applicable.

27. Patents, literature and product data

The current studded footwear test methods as specified in Regulation 12 (World Rugby 2015) are not patented and can be freely distributed. This project does not require commercial gains from the developed test method. Information about the design and final product will be distributed in the form of academic papers and an open-access thesis. The final design will most likely not be patented, in order not to hinder any potential adaptation of the test method for studded footwear regulations.

28. Political and social implications

If the developed test method is adapted as a new studded footwear standard, this could lead to a change in studded footwear sold and worn in rugby union.

29. Legal

Care should be taken that any stud 'passing' the developed test method, should not be seen as safe. Rather, its injury risk is comparable or less than the comparator stud, which universally has been accepted as an 'acceptable injury risk' (World Rugby 2015).

30. Installation

The installation of the product will be done by the principal investigator and design engineer, who are also responsible for the design and build of the test method.

31. Documentation

Documentation on the development and design need to be kept, as well as documentation on operating and servicing instructions. The design and development will also be offered for publication in academic journals.

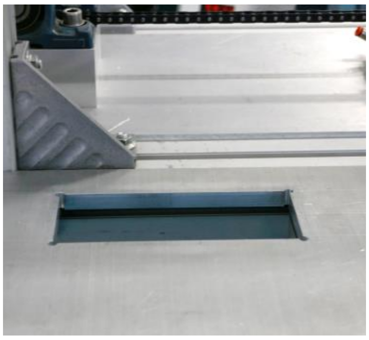
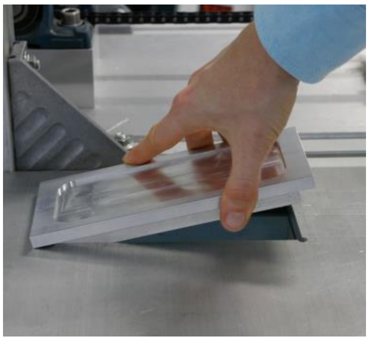
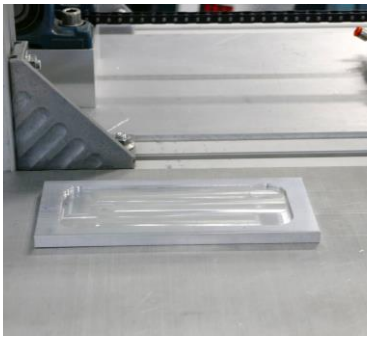
32. Disposal

Where possible, materials should be used which can easily be re-used or recycled if the product's life span comes to an end, or the product otherwise has to be dismantled.

10.4 Appendix D: Initial impact test procedure

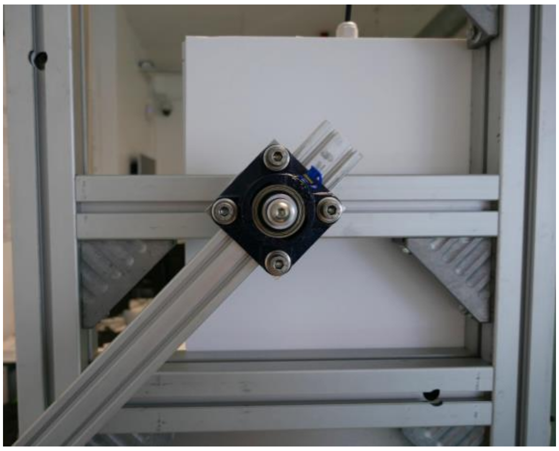
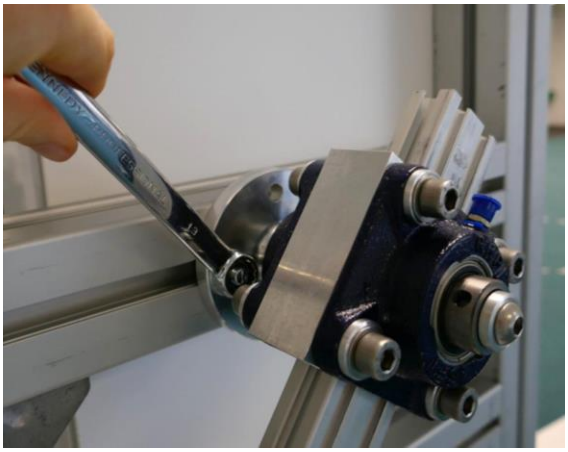
Step 1:

Place skin and soft tissue simulant tray in its position for the initial impact test.

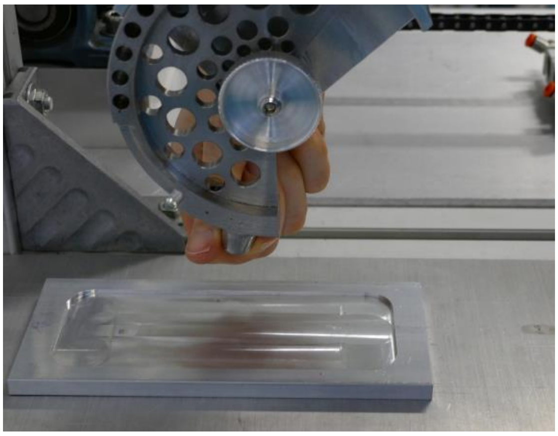
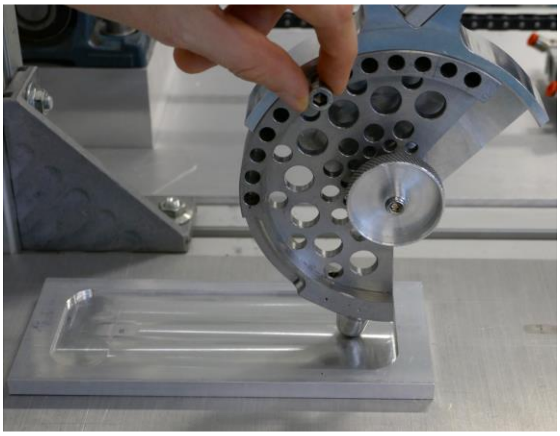
Step 2:

Set the angle of the inbound velocity at 30°, 45° or 60° by changing the pivot point of the pendulum arm.

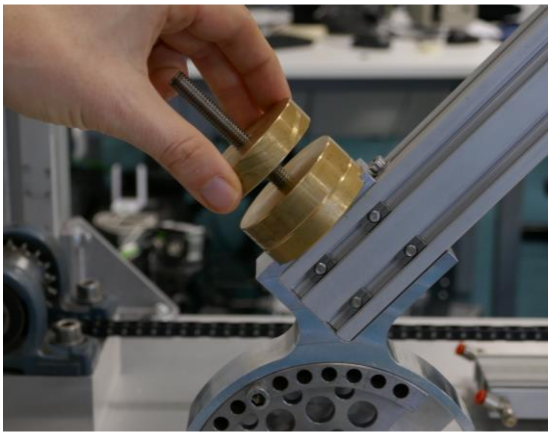
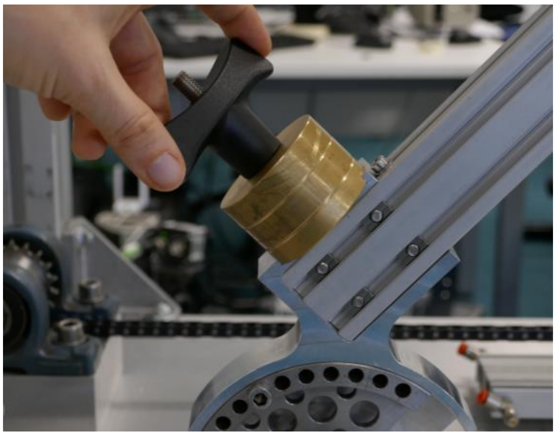
Step 3:

Screw the stud into the stud attachment. The impact angle of the stud can be set between -10° and 30°, at 5° interval options. Rotate the stud attachment until the required angle is achieved and lock in place with a bolt.

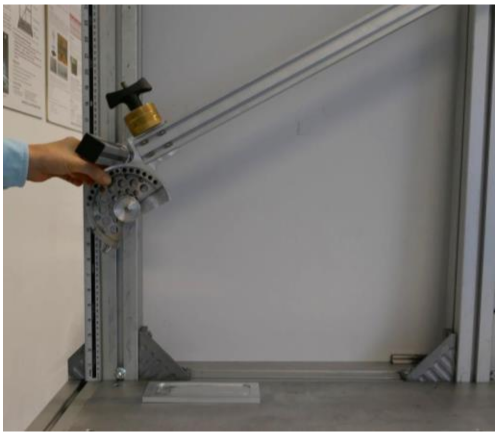
Step 4:

Set the impact mass required by adding weights to the stud attachment. The effective mass of the pendulum is 1.0 kg without any added weights, and can be increased to 2.2 kg in 0.2 kg intervals. Lock the added mass into place.

Step 5:

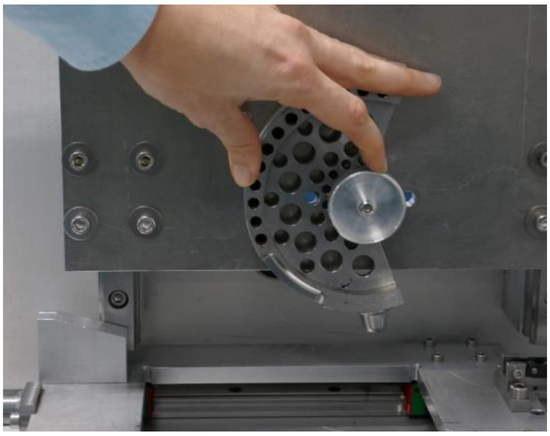
Set release height by adjusting the height of the bar. Hold the stud attachment against the bar and release when ready for the test.

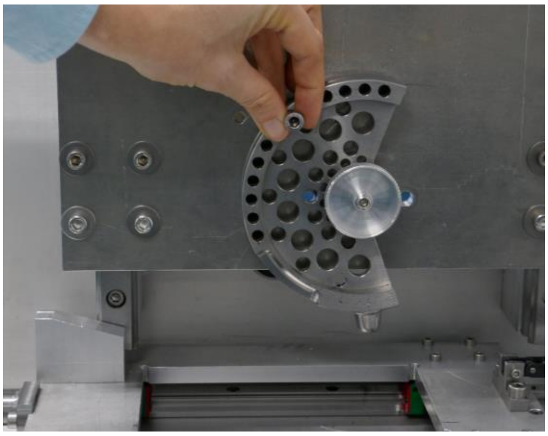


10.5 Appendix E: Raking phase test procedure

Step 1:

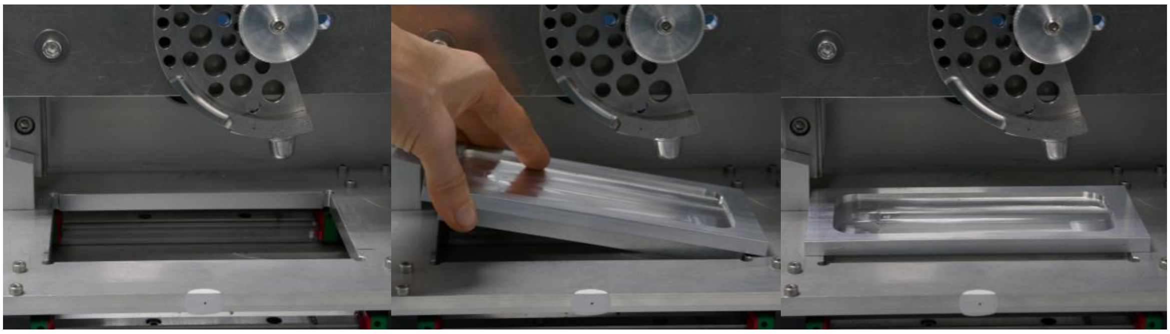
Remove stud attachment from the initial impact test and move to the raking test. Set the stud angle.





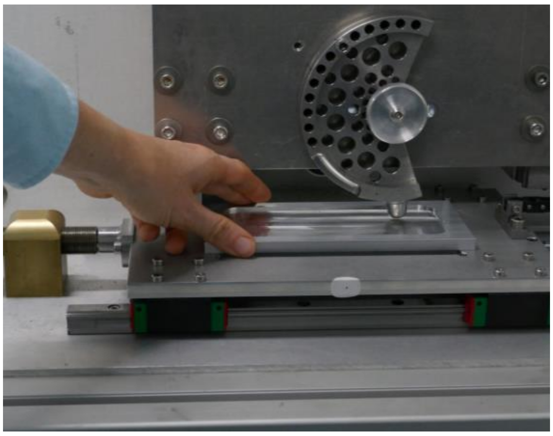
Step 2:

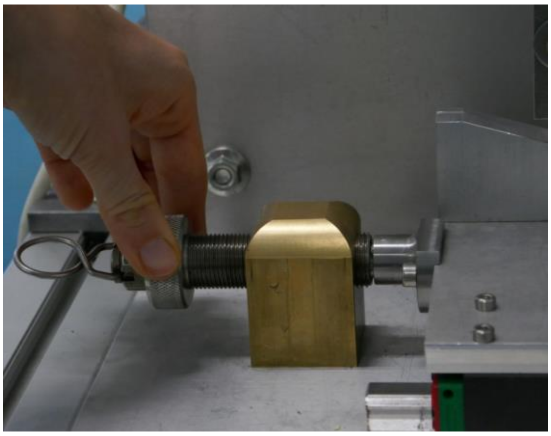
Remove skin simulant tray from the initial impact test and insert into the raking phase test.



Step 3:

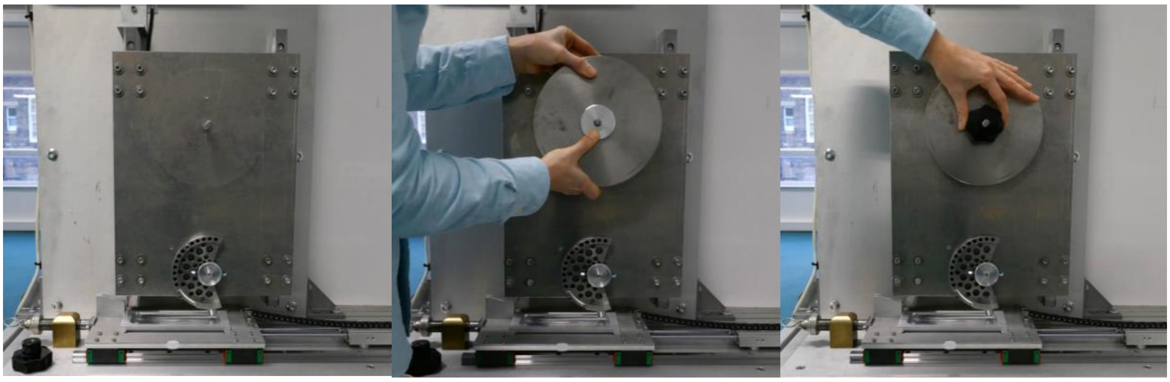
Align the stud with the initial impact damage by adjusting the position of the skin simulant tray. The tray can slide sideways (left picture) and the release mechanism can be adjusted forwards or backwards (right picture).





Step 4:

Add weights until the required raking mass is achieved. Without added weights, the raking mass is 6 kg. The mass can be increased at 4 kg intervals until 22 kg is reached. Secure mass by twisting knob.

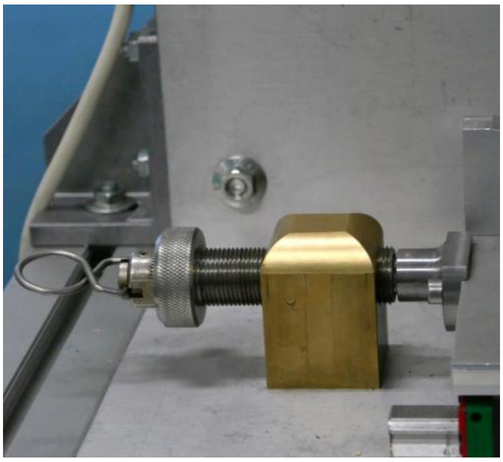


Step 5:

Adjust the pulling force by adding or removing springs and by increasing or decreasing the airflow of the damping unit. Test the resulting raking velocity with light gates or a high-speed camera.

Step 6:

Pre-tense the springs and bring simulant tray in its starting position. Secure the release pin. Lower the stud attachment and raking mass until the system is fully supported by the stud.



Step 7:

Pull the release pin when ready to start the test.